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The effects of topology and relative density of lattice liners on traumatic brain injury mitigation

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ABSTRACT

This paper evaluates the effects of topology and relative density of helmet lattice liners on mitigating Traumatic Brain Injury (TBI). Finite Element (FE) models of new lattice liners with prismatic and tetrahedral topologies were developed. A typical frontal head impact in motorcycle accidents was simulated, and linear and rotational accelerations of the head were recorded. A high-fidelity FE model of TBI was loaded with the accelerations to predict the brain response during the accident. The results show that prismatic lattices have better performance in preventing TBI than tetrahedral lattices and EPS that is typically used in helmets. Moreover, varying the cell size through the thickness of the liner improves its performance, but this effect was marginal. The relative density also has a significant effect, with lattices with lower relative densities providing better protection. Across different lattices studied here, the prismatic lattice with a relative density of 6% had the best performance and reduced the peak linear and rotational accelerations, Head Injury Criterion (HIC), brain strain and strain rate by 48%, 37%, 49%, 32% and 65% respectively, compared to the EPS liner. These results can be used to guide the design of lattice helmet liners for better mitigation of TBI.

1. Introduction

Helmets are the most effective item of personal protective equipment for mitigtion of the risk of head injuries (Fernandes and Alves de Sousa, 2013). However, they are mainly designed to protect the head against skull fracture and severe brain injuries (Marjoux et al., 2008; Zhang et al., 2009). Most helmets are designed to mitigate the transmitted translational forces to the head (ECE 22.05) even though it is known that the main mechanism of traumatic brain injuries is the rotational acceleration induced to the head due to oblique impacts (Holbourn, 1943; Krave et al., 2011; Post and Blaine Hoshizaki, 2015; Yoganandan et al., 2008). Despite the wide use of helmets, statistics show that head injuries are still the most common type of injury sustained by helmet users (Langlois et al., 2006; Piantini et al., 2016; Whyte et al., 2016). Hence, helmets should be further improved to reduce both rotational and translational forces for better mitigation of the risk of TBI. The helmet liner is the main energy absorbing part of the helmet, which is typically made of foams, such as expanded polystyrene (EPS). Some studies have shown that using an additively manufactured lattice liner, instead of EPS foams, can significantly improve the protection level of the helmets (Khosroshahi et al., 2018; Soe et al., 2015). Nonetheless, these lattices need to be designed for better TBI mitigation.

Recent advances in the field of additive manufacturing allow for a wide range of lattice structures, with unique mechanical properties, to be designed and manufactured (Liu and Hu, 2010; Mohsenizadeh et al., 2018; Sanami et al., 2014). Hitherto, most of the lattice structures introduced are designed to maximise stiffness and minimise mass (Du et al., 2017; Qiu et al., 2019; Song et al., 2019; Wang et al., 2019; Wu et al., 2019). A few studies have assessed whether lattice structures have better energy absorption when compared to conventional energy absorbing materials. These conventional materials typically have a stochastic cellular configuration, but lattice structures in contrast have an ordered configuration that can be designed according to required mechanical properties. This provides an opportunity to achieve a higher energy absorption efficiency than conventional foams (Schaedler et al., 2014). An experimental study showed that an additively manufactured lattice structure made of octet-truss unit cell has a comparable energy absorption to conventional materials such as EPS, but can recover almost completely after deforming by up to 70% strain

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(Mohsenizadeh et al., 2018). Moreover, compression tests of lattice structures at different loading speeds revealed that multilayer lattice structures respond differently at different loading rates (Ozdemir et al., 2017).

A few recent studies assessed the feasibility of the use of the lattice structures for head protection. Soe et al. (2015) proposed an additively manufactured cellular structure, with uniform cell sizes through the thickness, to improve the protection level of helmets. A recent study showed that an additively manufactured graded lattice liner reduced the risk of head injury compared to an EPS liner (Khosroshahi et al., 2018). In these studies, a specific type of unit cells was assessed and the effect of different toplogies was not investigated. Moreover, the performance of lattices as helmet liners for mitigation of traumatic brain injury (TBI) is still not fully understood.

Here we studied the effects of lattice topology and relative density on the TBI mitigation of lattice liners. We tested whether prismatic lattice liners can better protect the brain under oblique impacts than tetrahedral lattice liners, while also investigating whether lattices, as a whole, can outperform typical EPS liners. In addition to kinematics-based injury metrics, we used a high-fidelity model of TBI to determine strain and strain rate across the brain and used them to better assess the performance of the lattices in preventing TBI.

2. Methods

2.1. Lattice liners

Four different lattice topologies were designed to be incorporated into the helmet liner: prismatic and tetrahedral lattices, each with uniform and varying cell sizes (Fig. 1). For each lattice, we changed the number of cells through the thickness to achieve three different relative densities defined as $\frac{\rho^*}{\rho_s}$, where ρ^* is the density of the lattice and ρ_s refers to the density of the bulk material (Gibson and Ashby, 1999). In our previous study we found that Nylon has a better energy absorption than PLA (Khosroshahi et al., 2018). The lattices of current study were made of Nylon with a density of 1100 kg/m^3. In total, 12 different lattices were developed (Table 1).

A typical helmet has three main parts: liner, outer shell and the retention system. In this study, we considered the helmet as a hemispherical part to simplify its geometry. The inner radius of the liner and the thickness were 112 mm and 43 mm, respectively. The FE model of the liners was generated using beam elements in MATLAB using previously developed code (Khosroshahi et al., 2018). The Mat-24-Piecwise-Linear-Plasticity material model from LS-Dyna material library with Cowper-Symonds strain rate sensitivity formulation was used to simulate the Nylon lattice structure (LSTC, 2014; Moura et al., 2010). More information on the material properties, characterization and validation of the lattice FE models can be found in our previous work (Khosroshahi et al., 2018).

The helmet shell was modelled with 4-node quadrilateral shell elements and an elastic material model with Young's modulus of 7250 MPa, Poisson's ratio of 0.3 and density of 1200 kg/m^3 (Forero Rueda et al., 2009). The chin strap was modelled as an elastic band using 4-node quadrilateral shell elements with a Young's modulus of 1 GPa and a Poisson's ratio of 0.3. The helmet was fitted onto the THUMS human body model and surface to surface contact with a friction coefficient of 0.5 was defined at the head/liner and the liner/shell interfaces. These parameters are based on a previous study, which validated model predictions against experimental tests (Ghajari et al., 2011). Fig. 2 shows the helmeted human body model and different parts of the helmet. In the case of graded liners, the helmet was placed on the head in such a way that the smaller cells were close to the head and the larger cells were close to the outer shell of the helmet (Khosroshahi et al., 2018).

In order to compare the response of the lattice liner with conventional EPS liners, a model of the helmet with an EPS foam liner was also developed. The density of the liner was 50 gr/l, which is the EPS density usually used in commercial helmets, and its mass was 242 gr. The stress-strain curves of the EPS liner can be found in (Cui et al., 2009).

2.2. The total human body

The THUMS (Total HUman Model for Safety) V5 model was used in the present work (*THUMS Manual*) in order to include the effects of the body in helmeted head impacts. THUMS is a detailed human body finite element model, which incorporates details of the

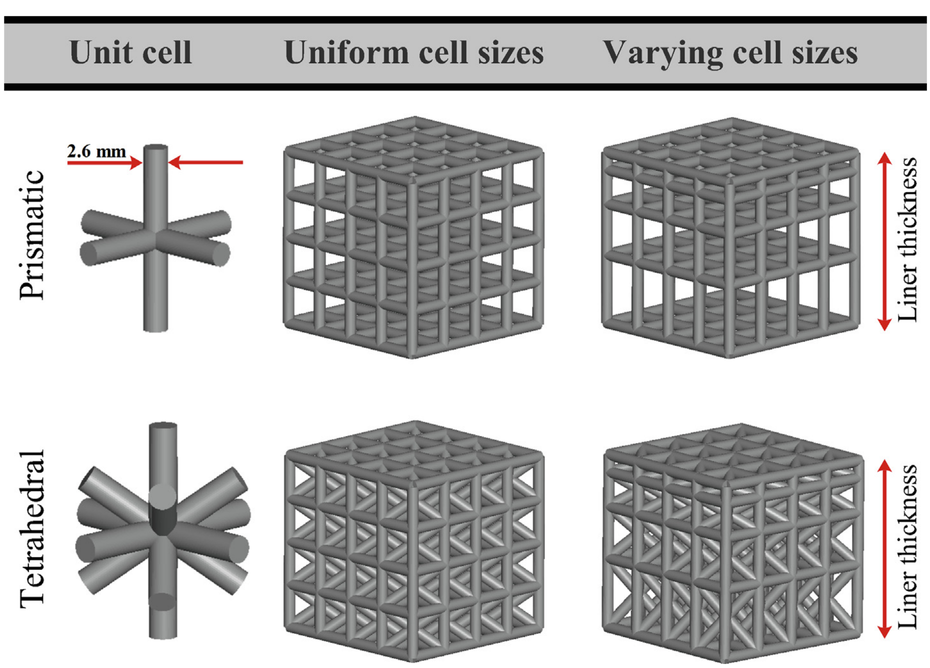

Fig. 1. Prismatic and tetrahedral lattice structures with four unit cells through the thickness.

Table 1
Relative density and the total mass of the lattice liners.

Prismatic unit cells			
Relative density [%] (and total mass [gr])			
No. of cells through thickness	4	5	6
Varying cell sizes	6.0 (320)	9.4 (500)	12.2 (650)
Uniform cell sizes	6.4 (343)	9.7 (516)	18.3 (696)
Tetrahedral unit cells			
Relative density [%] (and total mass [gr])			
No. of cells through thickness	3	4	5
Varying cell sizes	11.0 (586)	16.7 (890)	23.4 (1250)
Uniform cell sizes	11.3 (602)	17.2 (920)	24.2 (1294)

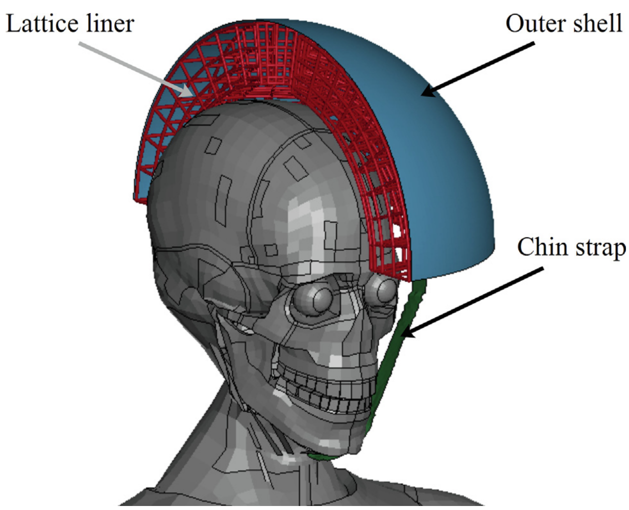

Fig. 2. The THUMS human body model equipped with a helmet model featuring a prismatic lattice liner (some parts are not shown).

cervical spine and muscles, and has been validated against cadaver and volunteer test results and used to represent the response of the human body in crash analyses (Chawla et al., 2005; *THUMS Manual*). The brain of THUMS, however, does not represent detailed anatomical features of the human brain, such as sulci and gyri. Therefore, in the present work, we used THUMS to determine the kinematics of the head during impacts and used the accelerations of the centre of gravity of the head to load a high fidelity model of TBI to determine strain and strain rate across the brain (Ghajari et al., 2017).

2.3. The traumatic brain injury model

The TBI model used in this study has 8 tissues, skin, skull, CSF, gray matter, white matter, ventricles, falx and tentorium (Fig. 3). Ogden hyperelastic constitutive model (MAT-77 from LS-Dyna material library) was used to model the response of the brain tissue (Ghajari et al., 2017). The Ogden model uses the following strain energy function:

$$\Psi^{\infty} = \sum_{p=1}^{n} \frac{\mu_p}{\alpha_p} \left(\lambda_1^{\alpha_p} + \lambda_2^{\alpha_p} + \lambda_3^{\alpha_p} - 3 \right) \tag{2}$$

in which $\lambda_{i(i=1,2 \text{ and } 3)}$ is principal stretch of the brain tissue. μ_p and α_p are material constants. The following formula was used to model the rate-dependent behavior of the brain tissue (Ghajari et al., 2017; Holzapfel, 2000):

$$S_{(t)} = S^{\infty} + \int_0^t G(t-T) \frac{\partial E(T)}{\partial T} dT \tag{3}$$

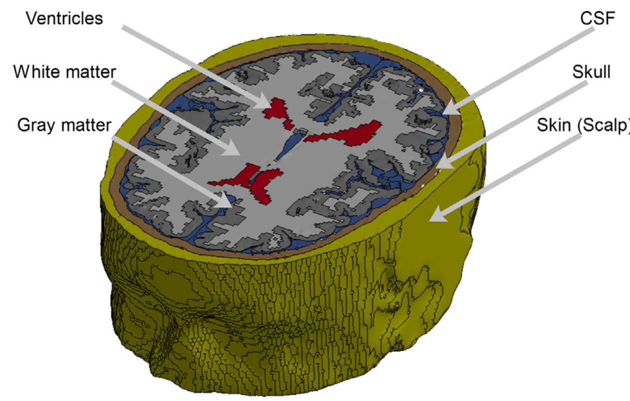

Fig. 3. The high-fidelity 3D FE model of traumatic brain injury (some parts of the model are not shown).

where S^{∞} is the long-term second Piola-Kirchhoff stress tensor and $G(t)$ is the relaxation function calculated as follows:

$$G(t) = \sum_{i=1}^{n} G_i \exp\left(\frac{-t}{\tau_i} \right) \tag{4}$$

where τ_i and G_i are material constants. The CSF, ventricles and membranes (i.e. pia, falx and tentorium) were modelled using hyperelastic and hyper-viscoelastic constitutive material models. All material constants for different tissues are presented in (Ghajari et al., 2017).

2.4. Oblique impacts

The helmets were tested virtually under oblique impacts. The helmeted THUMS was launched towards an inclined rigid anvil, forming a 45° angle to the horizon, at a speed of 7.5 m/s as shown in Fig. 4 (ECE 22.05). The same impact conditions were simulated for the helmets with the EPS and lattice liners. Simulations were performed on 6 cores of a 3.7 GHz processor using 32 GB RAM. The first 40 ms of the impacts were simulated, which took 10 h per simulation, and the linear and rotational accelerations of the centre of gravity of the head were recorded. All the simulations terminated without any stability problem. The TBI simulations were performed using 20 cores of a high-performance computer with 16 GB RAM, taking 3 h per simulation. Both kinematics and brain response measures were used to assess the performance of the helmet liners.

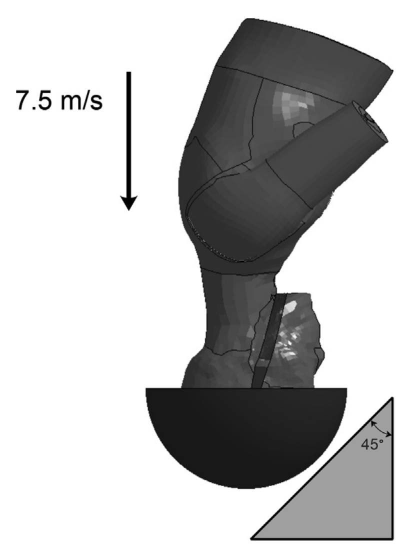

Fig. 4. Impact configuration (some parts of the model are not shown).

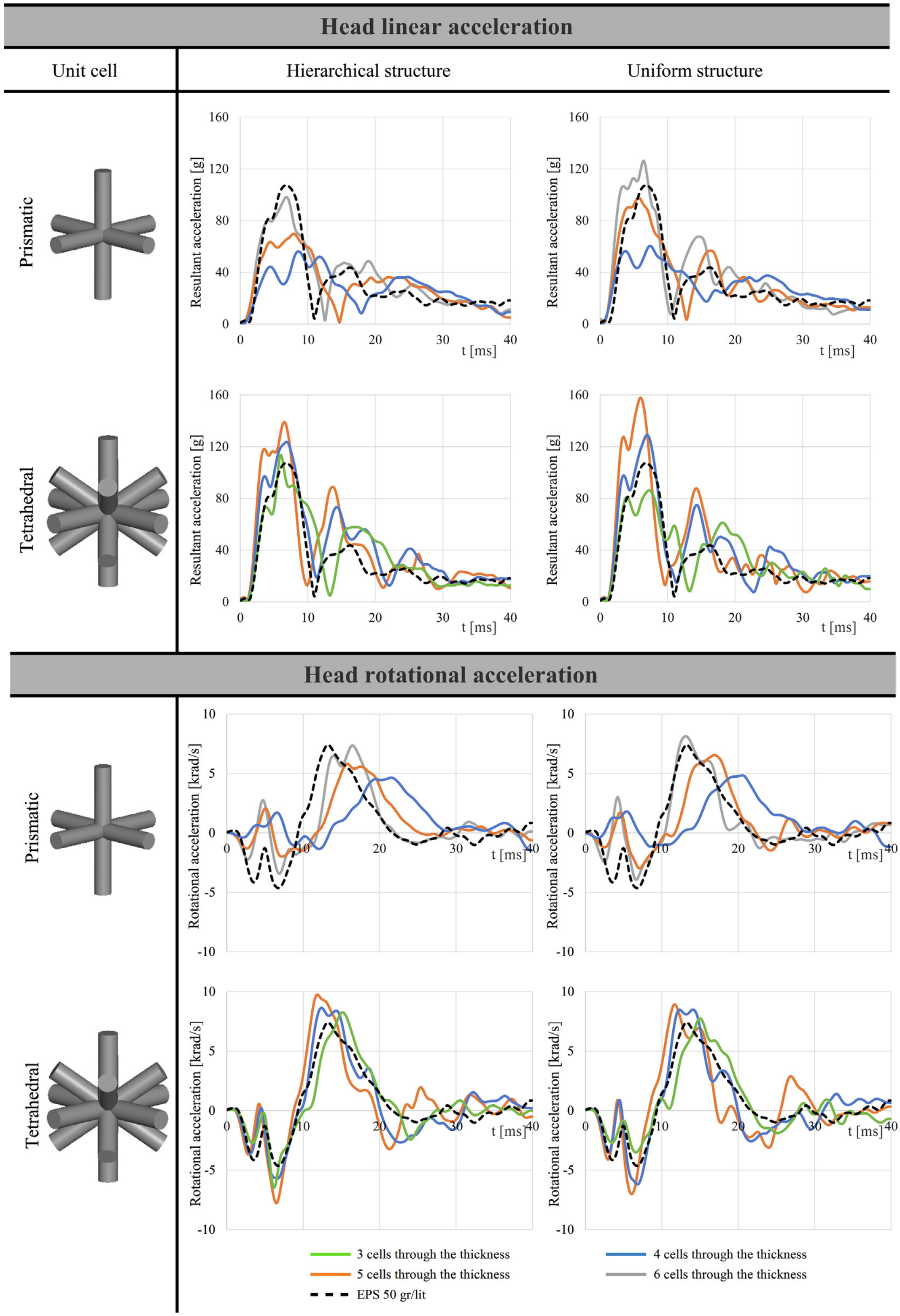

Fig. 5. Head linear and rotational acceleration for different helmet liners.

3. Results

All impacts led to a peak in linear accelerations within 10 ms of the impact's initiation (Fig. 5) which can be linked to the impact force (Mills, 2010). The rotational accelerations, in contrast, reached their peak value between 10 and 20 ms, which can be associated to the combined effects of the neck and the normal and tangential forces applied on the helmet by the anvil (Ghajari et al., 2013).

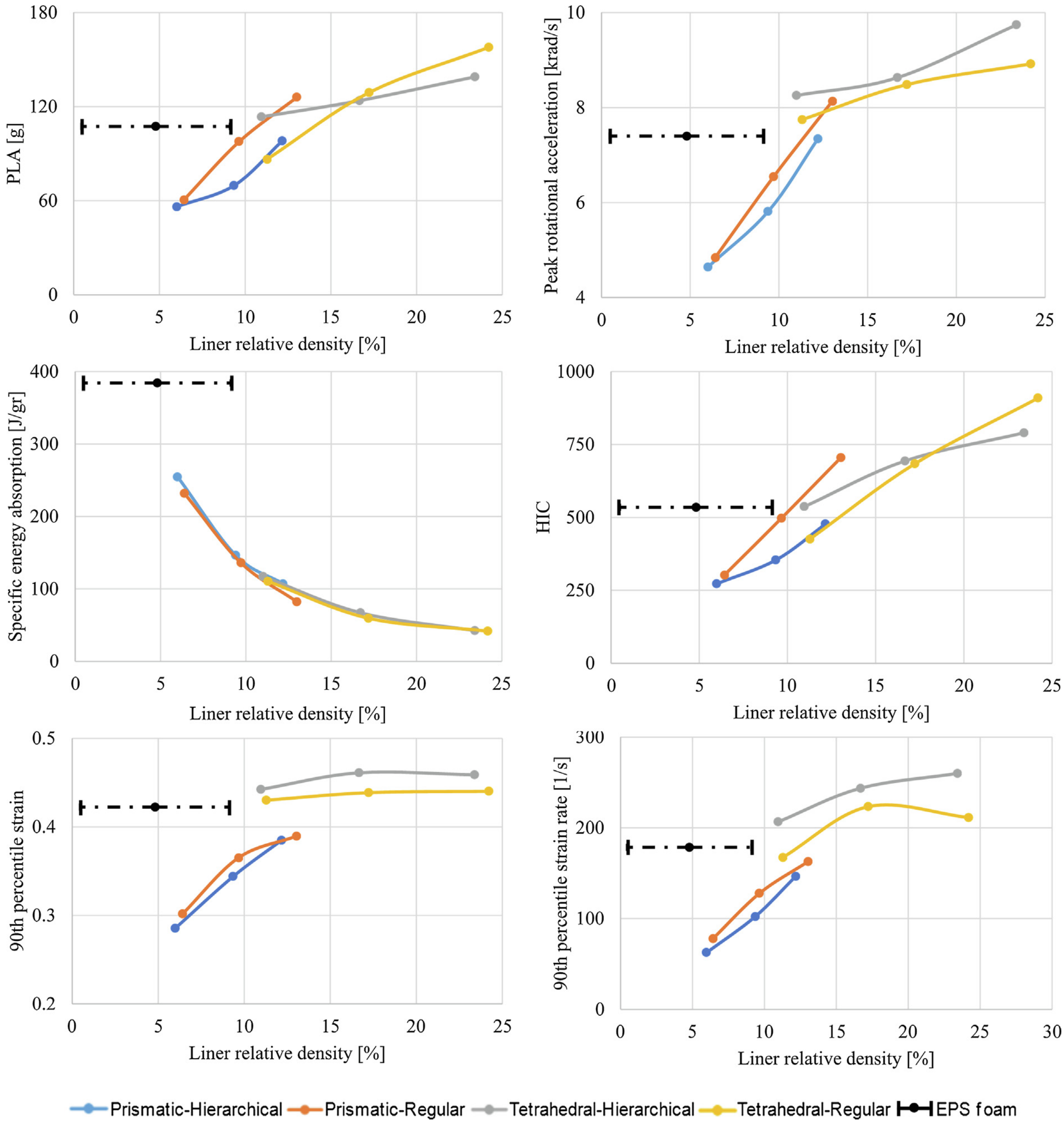

Fig. 6. Peak linear and rotational acceleration, HIC, energy absorption, brain 90th percentile strain and strain rate for different helmet liners.

3.1. Prismatic topology and lower relative densities decreased head accelerations

The results show that the lattice topology and relative density have large effects on accelerations. To better show these effects, we determined peak linear and rotational accelerations and HIC (Fig. 6). The helmets with prismatic lattice liners produced lower peak linear and rotational accelerations and HIC compared with the tetrahedral lattice liners. In addition, varying the cell density across the thickness decreased the head accelerations in all prismatic lattice liners, but not in all tetrahedral lattice liners. Our results also show that for each lattice topology, liners with lower relative densities had better impact performance. The helmet with the best performance was made of a graded lattice with four prismatic unit cells through the thickness (relative density = 6%). This liner reduced peak linear acceleration (PLA), peak rotational accelerarion (PRA) and HIC by 48%, 37%, and 49% respectively, when compared with the EPS liner.

3.2. Prismatic topology and lower relative densities also decreased brain strain and strain rate

Kinematics-based measures of TBI do not account for different tissues and details of brain anatomy. Hence, we used the FE model of TBI to predict the distribution of the maximum Green-Lagrange strain and strain rate across the brain for different liner designs. The contour plots (Fig. 7) show that with the EPS liner and most lattice liners, the strain and strain rate reached values larger than 40% and 200 s^{-1} respectively, which are well beyond brain injury threshold values suggested in previous studies (Ghajari et al., 2017). Interestingly, changing the lattice topology from tetrahedral

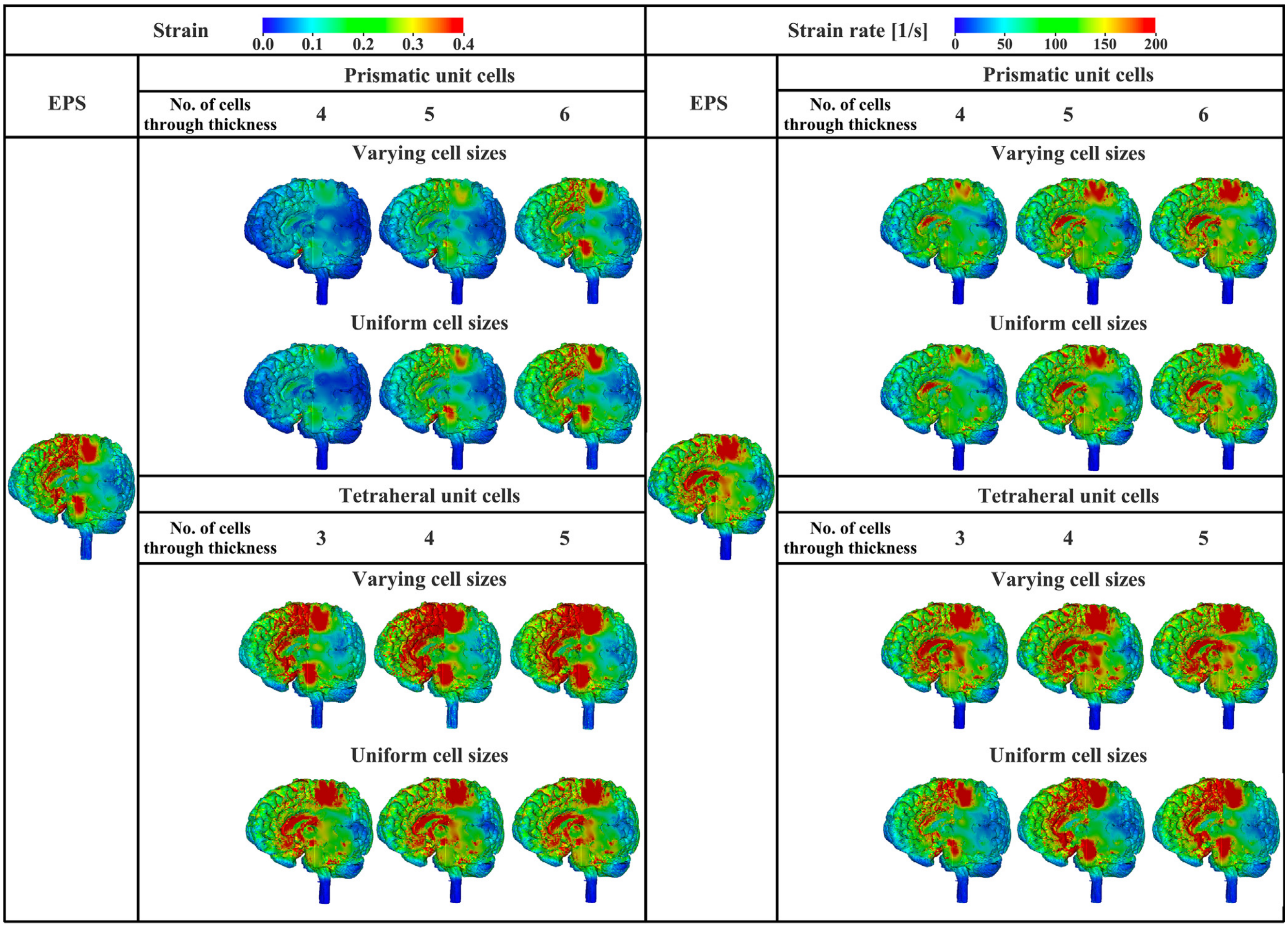

Fig. 7. Strain and strain rate distribution in the brain.

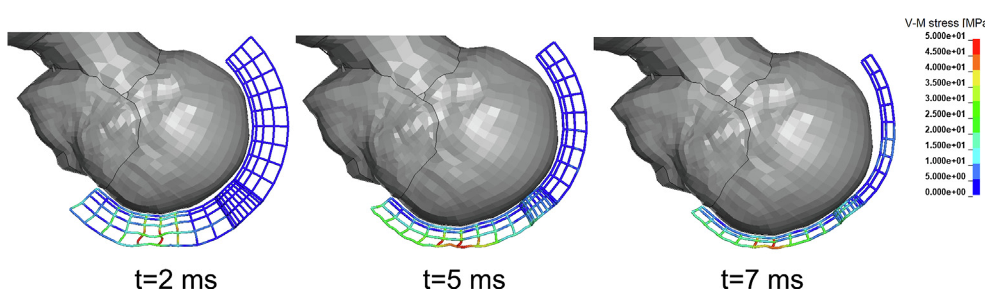

Fig. 8. Buckling initiation of the first (left), second (middle) and the third (right) layer of the liner (the collapsed layers are not shown).

to prismatic and decreasing the relative density significantly reduced the areas with large strains and strain rates.

To quantify the effects of the lattice topology and relative density on brain strain and strain rate, we determined the 90th percentile strain and strain rate across the brain for each liner (Fig. 6). The data confirm that the prismatic lattices produced lower strains and strains rates than tetrahedral lattices (22.5% reduction in strain and 48.2% reduction in strain rate). They also show that decreasing the relative density for each lattice topology decreased the strain and strain rate (10.9% reduction in strain and 26.8% reduction in strain rate for prismatic lattices and 2.9% reduction in strain and 20.6% reduction in strain rate for tetrahedral lattices). Varying the cell size across the thickness in prismatic lattices led to a small improvement in strain (1.0% Reduction) and strain rate (7.0% reduction). In tetrahedral lattices uniform cell sizes led to a small improvement in strain (4.1% reduction) and a large improvement in strain rate (18% reduction). The liner with the graded prismatic lattice liner and four cells through the thickness resulted in the lowest strain and strain rate in the brain, decreasing these measures of injury by 32% and 65% respectively compared with the EPS liner.

3.3. Larger specific energy absorption in prismatic lattices and lattices with lower relative density

We determined the specific energy absorption, defined as the energy by the liner divided by its mass. Our results show that tetrahedral lattices have lower specific energy absorption than

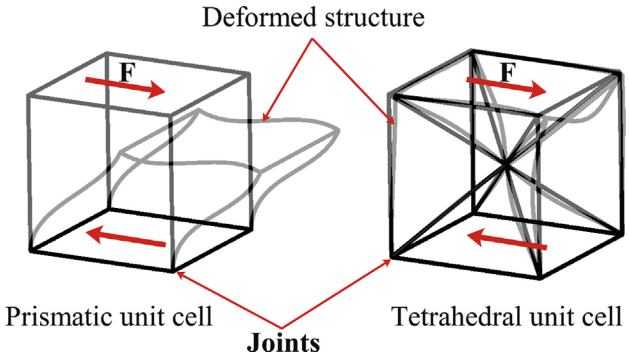

Fig. 9. Schematic deformation of stretching-dominated (left) and bending-dominated (right) lattices under the same load.

prismatic lattices, possibly due to their lower compliance and larger mass (Fig. 6). We also found that decreasing the relative density for each lattice topology increased the specific energy absorption. Varying the cell size across the liner thickness led to a marginal increase in the specific energy absorption in both prismatic and tetrahedral lattices. The specific energy absorption of the liner with 4 prismatic cells and varying cell size was the highest among the lattice liners, but interestingly it was still lower than the specific energy absorption of the EPS liner (33.7%).

4. Discussion

The results of our simulations show that the lattice liners with lower relative densities had larger specific energy absorption. This indicates that a larger portion of the lattice was involved in energy absorption when the relative density was decreased. The cellular structures with lower relative densities have lower yield stress thus they can crush at lower levels of the stress and dissipate the energy of the impact by transmitting lower forces to the head (Di Landro et al., 2002; Gibson and Ashby, 1999), providing higher level of protection.

According to our simulations the graded lattice liner with four cell sizes through the thickness can reduce the risk of traumatic brain injuries more than other lattice liners and the conventional EPS liner. This is due to the different bucking capacity of the struts in each layer, which let the liner crush from the most outer layer of the liner as shown in Fig. 8. This finding is in agreement with previous experimental and computational studies (Cui et al., 2009; Di Landro et al., 2002). In a graded-structure helmet liner, the liner reaches its maximum level of energy absorption by crushing almost equally in all layers from the most outer layer of the liner to the most inner layer. This is different to a regular structure where the level of energy absorption varies from the outer layer to the inner (Cui et al., 2009), which means not all layers of the liner reach their maximum level of energy absorption. Therefore, a graded structure helmet liner can dissipate the energy of the impact more than a regular structure liner.

Even though the protection level of the EPS liner, measured with PLA, PRA, HIC, strain and strain rate, is lower than that of the prismatic lattice liners, its specific energy absorption is higher. This can be explained by the negligible Poisson's ratio of the EPS. EPS foam can absorb the energy of the impact due to crushing but they cannot distribute it laterally. The EPS foam deforms in a small area, therefore, despite its large deformation and significant local energy absorption, it cannot distribute the energy of the impact in a wide area. In contrast, the lattice structures can better distribute the energy of the impact by engaging a larger portion of the liner in large deformation and energy absorption.

We found a significant decrease in strain and strain rate across the brain when using prismatic unit cells than tetrahedral (Figs. 6 and 7). This is attributed to the higher stiffness of lattice liners with tetrahedral unit cells due to their stretching-dominated response (Deshpande et al., 2001). Considering lattice structures as a network of struts connected to each other by pin joints, as shown in Fig. 9, helps to better understand their deformation mechanism (Deshpande et al., 2001). The lattice structure with prismatic unit cells (Fig. 9) responds as a mechanism when the cell collapses under loading, however, the lattice with the tetrahedral unit cells behaves as a structure and undergoes small deformation under loading. In the real lattice, there are no pin joints; the struts are fixed at their joints. As a result, the struts of the lattice without diagonal members (prismatic lattice) bend under loading leading to their buckling. This phenomenon does not occur in tetrahedral lattices due to their diagonal members. This is the main reason as to why, in general, lattice liners with prismatic unit cells (bending-dominated structure) can better absorb the energy of the impact and lower strain and strain rates across the brain than tetrahedral unit cells (Fig. 7).

In summary, the results of our simulations show that using a graded lattice liner with four cells through the thickness and without diagonal struts can significantly reduce the strain and strain rate across the brain compared to conventional EPS foams. This is driven by better capability of the lattice liners in both energy absorption and energy distribution. The advances in the field of additive manufacturing can allow such lattice structures to be produced on demand and with a greater level of simplicity and has the potential for more efficient substitutes for the conventional EPS liners in the near future. Although this work has shown the benefit of different lattice topologies in improving TBI prevention, the level of energy absorption and TBI prevention of the liners can still be improved and be proved with experimental tests. This is the subject of future studies.

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Declaration of Competing Interest

The authors have no conflicts of interest to disclose.

References

Chawla, A., Mukherjee, S., Mohan, D., Jain, S., 2005. Validation of the cervical spine model in Thums. Proc. 19th Int Tech. Conf. Enhanc. Saf. Veh., 1–11

Cui, L., Forero Rueda, M.A., Gilchrist, M.D., 2009. Optimisation of energy absorbing liner for equestrian helmets. Part II: Functionally graded foam liner. Mater. Des. 30, 3414–3419. https://doi.org/10.1016/j.matdes.2009.03.044.

Deshpande, V.S., Ashby, M.F., Fleck, N.A., 2001. Foam topology: bending versus stretching dominated architectures. Acta Mater. 49, 1035–1040. https://doi.org/10.1016/S1359-6454(00)00379-7.

Di Landro, L., Sala, G., Olivieri, D., 2002. Deformation mechanisms and energy absorption of polystyrene foams for protective helmets. Polym. Test. 21, 217–228. https://doi.org/10.1016/S0142-9418(01)00073-3.

Du, Y., Li, H., Luo, Z., Tian, Q., 2017. Topological design optimization of lattice structures to maximize shear stiffness. Adv. Eng. Softw. 112, 211–221. https://doi.org/10.1016/J.ADVENGSOFT.2017.04.011.

ECE 22.05, Uniform provisions concerning the approval of protective helmets and of their visors for drivers and passengers of motorcycles and mopeds, E/ECE/324 & E/ECE/TRANS/505 Addendum 21, Regulation No. 22, Revision 4, 2002. Geneva.

Fernandes, F.A.O., Alves de Sousa, R.J., 2013. Motorcycle helmets—a state of the art review. Accid. Anal. Prev. 56, 1–21. https://doi.org/10.1016/j.aap.2013.03.011.

Forero Rueda, M.A., Cui, L., Gilchrist, M.D., 2009. Optimisation of energy absorbing liner for equestrian helmets. Part I: Layered foam liner. Mater. Des. 30, 3405–3413. https://doi.org/10.1016/j.matdes.2009.03.037.

Ghajari, M., Hellyer, P.J., Sharp, D.J., 2017. Computational modelling of traumatic brain injury predicts the location of chronic traumatic encephalopathy pathology. Brain 140, 333–343. https://doi.org/10.1093/brain/aww317.

Ghajari, M., Peldschus, S., Galvanetto, U., Iannucci, L., 2013. Effects of the presence of the body in helmet oblique impacts. Accid. Anal. Prev. 50, 263–271. https://doi.org/10.1016/j.aap.2012.04.016.

Ghajari, M., Peldschus, S., Galvanetto, U., Iannucci, L., 2011. Evaluation of the effective mass of the body for helmet impacts. Int. J. Crashworthiness 16, 621–631. https://doi.org/10.1080/13588265.2011.616078.

Gibson, L.J., Ashby, M.F., 1999. Cellular Solids, Structure and Properties. Solid State Science Series, Cambridge.

Holbourn, A.H.S., 1943. Mechanics of head injuries. Lancet 242, 438–441. https://doi.org/10.1016/S0140-6736(00)87453-X.

Holzapfel, G.A., 2000. Nonlinear Solid Mechanics: A Continuum Approach for Engineering. Wiley.

Khosroshahi, S.F., Tsampas, S.A., Galvanetto, U., 2018. Feasibility study on the use of a hierarchical lattice architecture for helmet liners. Mater. Today Commun. 14, 312–323. https://doi.org/10.1016/j.mtcomm.2018.02.002.

Krave, U., Al-Olama, M., Hansson, H.A., 2011. Rotational acceleration closed head flexion trauma generates more extensive diffuse brain injury than extension trauma. J. Neurotrauma 28, 57–70. https://doi.org/10.1089/neu.2010.1431.

Langlois, J.A., Rutland-Brown, W., Wald, M.M., 2006. The epidemiology and impact of traumatic brain injury: a brief overview. J. Head Trauma Rehabil. 21, 375–378.

Liu, Y., Hu, H., 2010. A review on auxetic structures and polymeric materials 5, 1052–1063.

LSTC, 2014. LS-Dyna keyword user's manual.

Marjoux, D., Baumgartner, D., Deck, C., Willinger, R., 2008. Head injury prediction capability of the HIC, HIP, SIMon and ULP criteria. Accid. Anal. Prev. 40, 1135–1148. https://doi.org/10.1016/j.aap.2007.12.006.

Mills, N.J., 2010. Critical evaluation of the SHARP motorcycle helmet rating. Int. J. Crashworthiness 15, 331–342. https://doi.org/10.1080/13588260903251000.

Mohsenizadeh, M., Gasbarri, F., Munther, M., Beheshti, A., Davami, K., 2018. Additively-manufactured lightweight metamaterials for energy absorption. Mater. Des. 139, 521–530. https://doi.org/10.1016/j.matdes.2017.11.037.

Moura, R.T., Mazzariol, L.M., Clausen, A.H., Langseth, M., Alves, M., 2010. Axial Loading of Cylinders Made of Nylon with Nano-particles. Proc. IMPLAST 2010 Conf.

Ozdemir, Z., Tyas, A., Goodall, R., Askes, H., 2017. Energy absorption in lattice structures in dynamics: nonlinear FE simulations. Int. J. Impact Eng. 102, 1–15. https://doi.org/10.1016/j.ijimpeng.2016.11.016.

Piantini, S., Pierini, M., Delogu, M., Baldanzini, N., Franci, A., Mangini, M., Peris, A., 2016. Injury analysis of powered two-wheeler versus other-vehicle urban accidents, in: IRCOBI Conference. Malaga, Spain.

Post, A., Blaine Hoshizaki, T., 2015. Rotational acceleration, brain tissue strain, and the relationship to concussion. J. Biomech. Eng. 137. https://doi.org/10.1115/1.4028983.

Qiu, W., Lu, F., Wang, G., Huang, G., Zhang, H., Zhang, Z., Gong, C., 2019. Evaluation of mechanical performance and optimization design for lattice girders. Tunn. Undergr. Sp. Technol. 87, 100–111. https://doi.org/10.1016/J.TUST.2019.02.008.

Sanami, M., Ravirala, N., Alderson, K., Alderson, A., 2014. Auxetic materials for sports applications. Procedia Eng. 72, 453–458. https://doi.org/10.1016/j.proeng.2014.06.079.

Schaedler, B.T.A., Ro, C.J., Sorensen, A.E., Eckel, Z., Yang, S.S., Carter, W.B., Jacobsen, A.J., 2014. Designing Metallic Microlattices for Energy Absorber Applications 276–283. https://doi.org/10.1002/adem.201300206.

Soe, S.P., Martin, P., Jones, M., Robinson, M., Theobald, P., 2015. Feasibility of optimising bicycle helmet design safety through the use of additive manufactured TPE cellular structures. Int. J. Adv. Manuf. Technol. 79, 1975–1982. https://doi.org/10.1007/s00170-015-6972-y.

Song, J., Wang, Y., Zhou, W., Fan, R., Yu, B., Lu, Y., Li, L., 2019. Topology optimization-guided lattice composites and their mechanical characterizations. Compos. Part B Eng. 160, 402–411. https://doi.org/10.1016/J.COMPOSITESB.2018.12.027.

THUMS, AM50 Occupant Model, Academic Version 5.0, 2015.

Wang, Y., Groen, J.P., Sigmund, O., 2019. Simple optimal lattice structures for arbitrary loadings. Extrem. Mech. Lett. 29, 100447. https://doi.org/10.1016/J.EML.2019.03.004.

Whyte, T., Gibson, T., Anderson, R., Eager, D., Milthorpe, B., 2016. Mechanisms of head and neck injuries sustained by helmeted motorcyclists in fatal real-world crashes: analysis of 47 in-depth cases. J. Neurotrauma. https://doi.org/10.1089/neu.2015.4208.

Wu, Z., Xia, L., Wang, S., Shi, T., 2019. Topology optimization of hierarchical lattice structures with substructuring. Comput. Methods Appl. Mech. Eng. 345, 602–617. https://doi.org/10.1016/J.CMA.2018.11.003.

Yoganandan, N., Li, J., Zhang, J., Pintar, F.A., Gennarelli, T.A., 2008. Influence of angular acceleration-deceleration pulse shapes on regional brain strains. J. Biomech. 41, 2253–2262. https://doi.org/10.1016/j.jbiomech.2008.04.019.

Zhang, J., Yoganandan, N., Pintar, F.A., 2009. Dynamic biomechanics of the human head in lateral impacts. Ann. Adv. Automot. Med. Assoc. Adv. Automot. Med. Annu. Sci. Conf. 53, 249–256.