

Investigation of biomechanics of skull structures damages caused by dynamic loads

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Purpose: The aim of the study was to examine the influence of cranial sutures on the crack behaviour of a human skull after the impact. The authors focused on the assessment of skull breaking nature, based on a real-world vehicle-to-bicyclist accident. In the state of the art, there is still no consensus about sutures mechanical properties. Currently, most of the numerical head models do not have distinguished cranial sutures. *Methods:* The authors compared different elastic properties for cranial sutures and their influence on the nature of the skull fracture. The mathematical and numerical modelling have been applied to mimic the nature of the skull fracture. The LS-DYNA explicit code with material models featuring the erosion of finite elements was used. The models of the skull with different cranial sutures properties were impacted against a validated front-end of a vehicle. *Results:* Various fracture patterns were obtained for different material properties of the sutures and the results were compared to a model without the cranial sutures. Based on the results, a graph was plotted to indicate differences in sutures energy absorption capabilities. The numerical results were supported by the mathematical modelling. The developed diagram may enable better understanding of the complex mechanical phenomena on the suture interface. *Conclusions:* Biomechanical evidence was provided for the important role of the sutures in numerical models as well as their significant influence on the biomechanics of skull fractures caused by dynamic loads.

Key words: skull fracture, biomechanics, cranial sutures, vulnerable road user, finite element method

1. Introduction

Traumatic brain injury (TBI) is one of the most severe injuries in the world [23]. The annual TBI hospitalization rate is 235–275 per 100 000 people [18]. The problem of cranio-cerebral injuries is significant and has long-term consequences. The main causes of brain tissue damage are mechanical overloads, which occur to the head during traffic accidents, sports injuries and in the acts of violence. WHO predicts that by the year 2020 road accidents will become the leading cause of premature death. Currently, the total number of fatalities is estimated to be 2 million every year. 75% of this number are cyclists, motorcyclists and pe-

destrians [21]. The percentage of cyclists killed correlates strongly with countries' bicycle tradition and proper infrastructure, which encourages inhabitants to choose bicycle as a mean of transport. In Poland in 2017, there were 31 106 road accidents in which 42 297 people were injured or killed [16]. The head injuries are one of the main causes of fatalities in bicycle accidents. Cyclists who died as a result of an accident usually did not wear helmets [13]. In addition, the most common head injuries include skull fractures [12].

Therefore, the explanation of the biomechanics of the cranial structure damage is crucial in the aspect of safety devices, forensic analysis and treatment of head injuries. Nevertheless, to establish the countermeasures such as proper bicycle helmets, we need to rec-

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ognise and comprehend the mechanism of the occurrence of head injuries, including skull fractures. However, until now, the cranial fracture mechanism resulting from mechanical overloads has not been fully understood [24], in particular, the role of cranial sutures in the context of biomechanical response to physical forces [20], [25].

The main building component of the bone tissue is mineralized collagen fibril (*ossein*). The collagen fibres in the periosteum and endosteum are arranged in a hierarchical way. On the contrary, in the diploë, collagen fibrils are oriented irregularly and the structure is spongy. The position of fibres in the bone affects its mechanical properties. Nevertheless, the microstructure of the arrangement of these fibres and its mechanical characteristics are not well known. The bone structure undergoes constant remodelling in order to repair micro-damages and to adapt to its mechanical function [15]. All skull bones are connected by suture joints – the cranial sutures are composed mainly of collagen. Three cranial sutures intersect the structure of the skull: coronal, sagittal, lambdoid (Fig. 1). The histological architecture of the sutures are variable and complex [22].

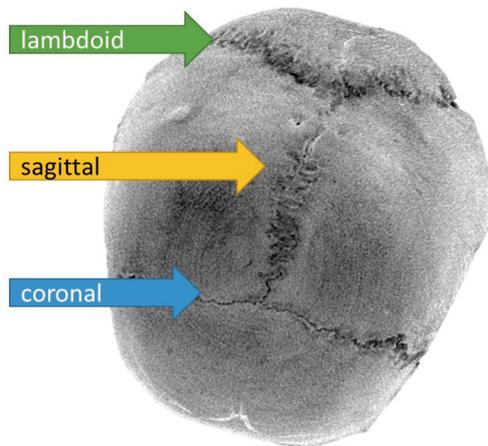


Fig. 1. Skull sutures: view of the convexity with coronal (blue arrow), sagittal (orange arrow) and lambdoid (green arrow) sutures

The literature lacks the consensus on the mechanical properties of cranial sutures. This is due to their different spatial distribution of collagen fibres, which results in different geometries and mechanical properties. However, researchers are limited to the amount of the samples taken from human cadavers. Therefore, the mechanical properties of cranial sutures significantly differ in the literature.

Measuring strain or damage during an impact, especially *in vivo*, is a challenge, which also implies ethical issues. Thus, one of the most robust methods to simulate

and address fractures or delamination in multiphase biomechanical systems, is the finite element method (FEM) [14]. It should be emphasized that with the rapid development of the computing power the number of numerical head models in the literature has significantly increased. Nowadays, head models contribute expressively to TBI research. Nevertheless, there is still limited research concentrated on cranial trauma in the context of road traffic participants, in particular the collision of vehicles with cyclists. At the same time, despite the efforts made to understand the global functionality and mechanics of the head, little attention has been paid to the understanding of the biomechanical features of the skull and, in particular, the cranial sutures, especially in the context of sutures under dynamic loading conditions. Currently, it is still not explained how energy dissipation occurs in cranial sutures under mechanical load.

Overall, this novel study aims to investigate on how dynamic loading affects the mechanics of cranial sutures and surrounding bones – the case is based on a real-world car accident with a vulnerable road user.

2. Materials and methods

2.1. Finite element head model

The skull model was created based on computed tomography (CT) scan of an adult male head – a cyclist, who was struck from the left side by a motor vehicle. As a result of the impact, the skull was fractured – the physical examination upon arriving at the hospital revealed Glasgow Coma Scale score of 3 (E1V1M1). Figure 2 depicts the CT medical image and a 3D visualisation of the patient's skull.

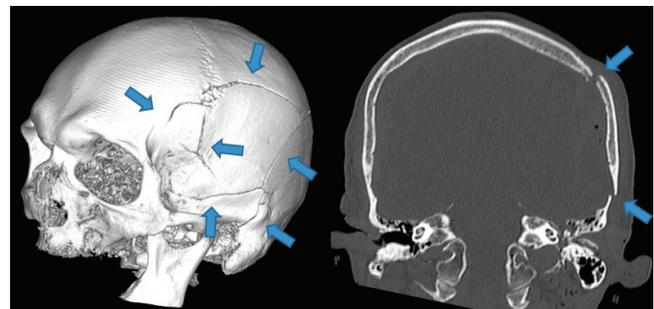


Fig. 2. A patient with a skull fracture. Fracture sites are marked with arrows

The dedicated software 3D Slicer enabled the authors to visualise the skull separately from the other

head structures. Due to its complex geometry, the facial bones, more specifically the ethmoid bone and the segment of the sphenoid bone, were removed. As a result, the stereolithographic (STL) model of the region of interest was obtained. The 3D STL object created enabled the authors to proceed with digital processing using computer-aided software such as CATIA v5 and MeshLab. In CATIA v5, the irregular areas of the head were simplified – especially within the sphenoid bone and the remains of the ethmoid bone.

The external and internal surfaces – periosteum and endosteum – were assigned with the same thickness of 1 mm and meshed with *penta* finite elements. According to [11], [19], the thicknesses of the compacted layers are approximately 1 mm. Thus, the thickness of each of the layers in the model fits this range. The diploë layer was further meshed with *tetra* finite elements. Therefore, its thickness, and, in consequence, the total thickness of the cerebral part of the skull is variable. This approach is in line with the real skull geometry, whose thickness is not constant [4]. Additionally, according to [11], there is no significant correlation between skull thickness and subject's age, which also supports the carried out discretization process. Overall, the FE model of the skull consists of 117 670 finite elements. The procedure for creating a numeric model of the skull is depicted in Fig. 3.

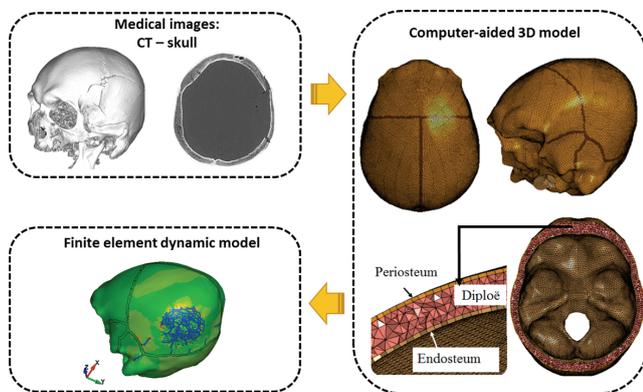


Fig. 3. The procedure for creating a numerical model of the skull

The skull material properties are in accordance with the Global Human Body Model Consortium (GHMBC) model [5] using *piecewise linear plasticity* material model (MAT 24) in LS-DYNA. Periosteum and endosteum were modelled with the same mechanical properties. The overall applied material model includes the feature of eroding individual finite elements by exceeding specific thresholds. For periosteum and endosteum, the maximum principal strain equals 0.42%, while for diploë the maximum principal

stress is 20 MPa [5]. Mechanical properties such as density (δ), Young's modulus (E) and Poisson's ratio (ν) are summarized in Table 1.

Table 1. Mechanical properties for individual layers of the skull

Skull layer	δ [t/mm ³]	E [MPa]	ν [-]
Periosteum and endosteum	2.1e-9	0.1e5	0.25
Diploë	1.0e-9	0.6e4	0.3

The numerical model includes cranial sutures. The literature review has indicated that the material properties of cranial sutures for an adult are not fully tested. Currently, more attention is focused on the study of mechanical properties of sutures in children. This is due to a change in the geometry of the skull during the children growth [1]. Furthermore, the test outcome between the authors do not agree with the mechanical properties of cranial sutures in adults [7], [10], [25]. The discrepancy between these results may result from individual mechanical properties in human or are related to the age. In the state of the art, Young's modulus for sutures is given in the range 50–500 MPa with the Poisson's ratio of 0.3. However, depending on the stiffness of cranial sutures, the biomechanical response of the skull to mechanical loads may be different. Due to the lack of consensus within literature, the different stiffness values of cranial sutures were taken into account in the presented research. The cranial sutures were modelled as isotropic structures. Mechanical properties for cranial sutures are summarized in Table 2.

Table 2. Mechanical properties of cranial sutures basing on various references

Reference #	Reference	δ [t/mm ³]	E [MPa]	ν [-]
[6]	Jasinowski et al., 2010	2.1e-9	50	0.3
[9]	Li et al., 2011	2.1e-9	100	0.3
[12], [25]	Margulies and Thibault, 2000; Zhang and Yang, 2015	2.1e-9	200	0.3
[25]	Zhang and Yang, 2015	2.1e-9	300	0.3
	Zhang and Yang, 2015	2.1e-9	400	0.3
	Zhang and Yang, 2015	2.1e-9	500	0.3

2.2. Finite element model of vehicle and impact simulation

As it was mentioned before, the numerical test was carried out for a male cyclist who was severely injured in a traffic accident with a sports car. Figure 4

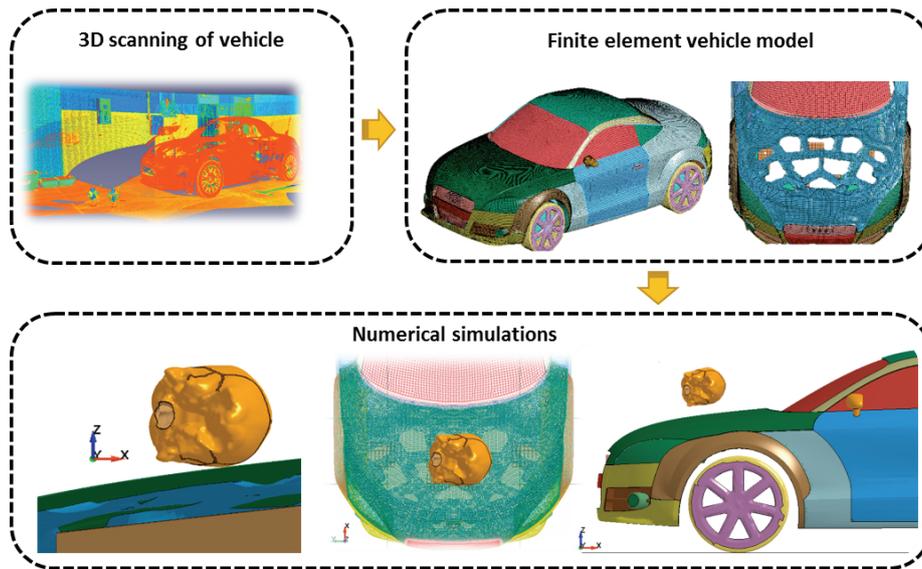


Fig. 4. Methodology for the investigation of damage to skull structures

depicts the vehicle geometry, which was obtained as a result out of 3D laser scanning. The result of the scan was an assembled point cloud. Based on the developed geometric model of the vehicle, a FE model in LS-DYNA explicit code was created – using the methodology, which was earlier validated for pedestrian accidents and described by Ptak and Fernandes et al. [2], [17]. For the sake of brevity, the material models for the vehicle are fully referred in [3]. The boundary conditions, such as the velocity and impact position used in this study, were based on real accident data provided by the Neurosurgery Department at Legnica Hospital. The initial velocity of 80 km/h was assigned to the skull and the horizontal approach angle was set to 65° basing on the forensic investigation. The bicyclist did not wear a helmet at the time of the impact – consequently, this safety device is dismissed by the authors in this study. The methodology for the biomechanical investigation of skull structures damage is shown in Fig. 4.

3. Results

In this study, six skull models with different stiffness parameters (Table 2) of cranial sutures were analysed and further compared to a model without the sutures. As a result of the numerical analyses, the authors obtained maps of principal stress on the skull (Fig. 5). The results showed that the stiffness of cranial sutures has a significant impact on the skull bone fracture path. The greater the stiffness of cranial sutures, the smaller the number of cracks on periosteum

near the cranial suture. This phenomenon is confirmed by the plot of the internal energy in the cranial sutures, which shows that energy decreases, when the Young's modulus increases (Fig. 7). This explains that the value of the Young's modulus of the cranial sutures is closer to the stiffness of the cranial bones, while the distribution of dynamic stresses on periosteum rapidly achieve uniformity. It should be noted here, that in the skull model without cranial sutures the stress concentrates in the region where the skull impacts the vehicle's bonnet.

Figure 5 depicts the maps of the maximum principal stress for all cases, at $t = 5$ ms after the initial contact with the bonnet. The left part shows the outside view of the skull, whereas the right presents a cross-section view focusing on the endosteum. For the maps of maximum principal stress, it should be noted that the region of direct contact between bonnet and skull shall not be analyzed in detail. The mentioned erosion of finite elements creates a coarse stress state close to the region of the direct impact. However, the energy dissipation by diploë layer is relevant. The stress distribution on the skull without sutures is smooth across the whole skull and next to the region of the direct contact. The erosion elements algorithm does not show any long fracture.

While increasing the value of $E = 50$ MPa to $E = 500$ MPa, it becomes visible that number and length of the cracks are decreasing. Most of the eroded finite elements occurred in periosteum which is subjected to the direct contact with the vehicle front-end. However, most of the long cracks occurred on the endosteum. Generally, it should be mentioned, that the angle between the direction of the fracture

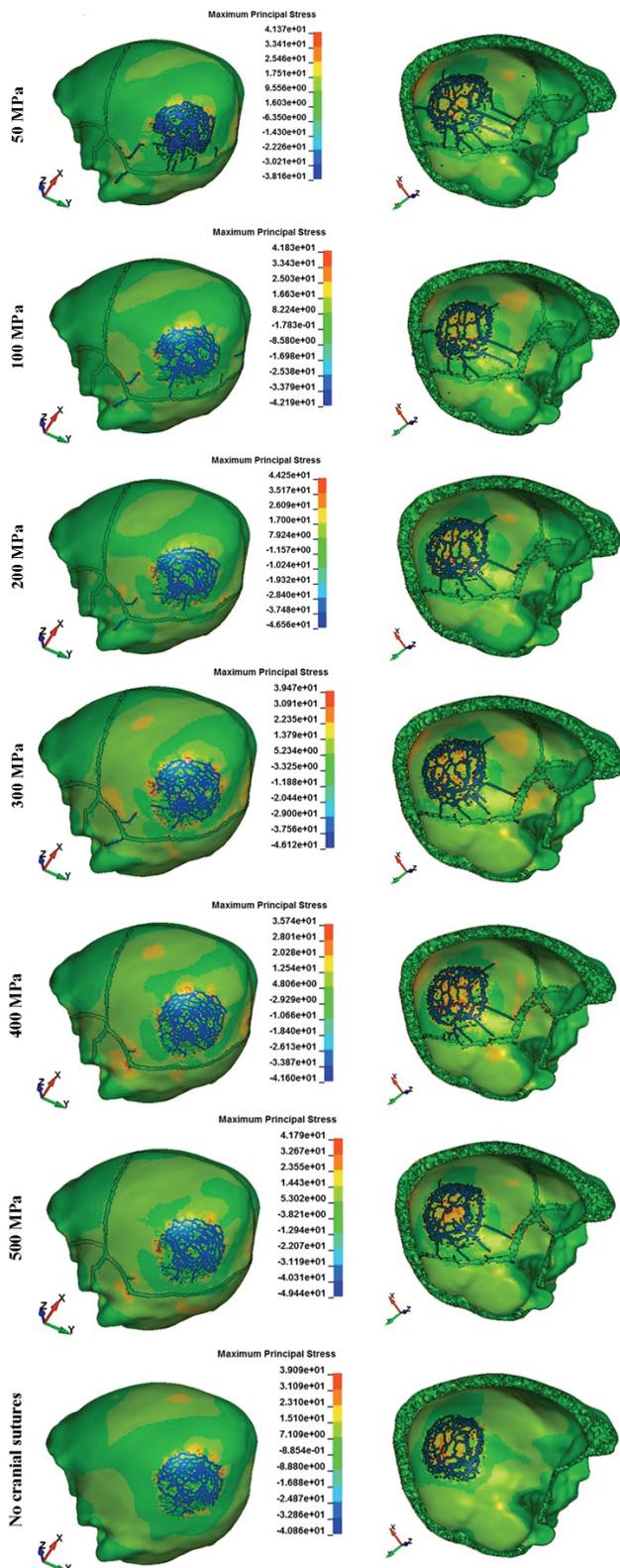


Fig. 5. Distribution of maximum principal stress and fracture pattern on periosteum and endosteum for cranial suture with Young's modulus of: 50–500 MPa and for a skull without cranial sutures; 5 ms after the impact

and the tangent line to the sutures in the starting point of the crack is tending towards 90° . The nature

of the crack is related to the material structure and properties of the affected bone. The ultimate tensile stress of endosteum or periosteum is lower than the ultimate compressible stress. This situation leads to a crack initiating in a notch on the tensile side of the bended bone. The subsequent erosion of the affected finite elements, which is also responsible for the further crack propagation is performed according to the tensile failure criterion. This is applicable for both periosteum and endosteum. Overall, the observed behavior leads to an initial cracking of the endosteum layer, followed by the periosteum layer. Concentrating on the boundary of the skull toward the sutures, the Young's modulus of the suture is lower than the bones Young's modulus. The increased capability to deform means also that the suture is not able to prevent the skull bone from bending. As a result, the fracture occurs at the tensile side of the bone perpendicular to the suture. The resultant effect of the crack growth is shown in Fig. 6. The element erosion used for crack modelling allowed the authors to simulate the nature of skull fracture. In the remaining cases, cracks might occur on both sides of suture. With time, the crack is growing towards the center of the skull tables. The blue color shows finite elements, which met failure criteria.

The internal energy stored in sutures for each considered Young's modulus is shown on Fig. 7. In the case when a lower Young's modulus is assigned to a suture, the energy is increasingly transferred to the suture. Up from $t = 10$ ms, the internal energy shown in Fig. 7 is building a plateau, the elastic energy entrapped in the sutures due to the plastic deformation of the diploë layer. Moreover, the peak of internal energy stored in sutures is over four times greater for the highest-to-lowest considered Young's modulus. Higher energy absorption is directly connected to the strain. Consequently, it seems reasonable to continue this approach due to the proved high influence concerning stress distribution and the according failure nature for the skull.

4. Discussion

The systematic review of the literature revealed that there is very limited knowledge about the biomechanical response of cranial sutures under mechanical loading. The research presented in the literature shows that Young's modulus for cranial sutures is in the range of 50–500 MPa. Nevertheless, the cranial sutures have a much smaller Young's modulus than skull bones.

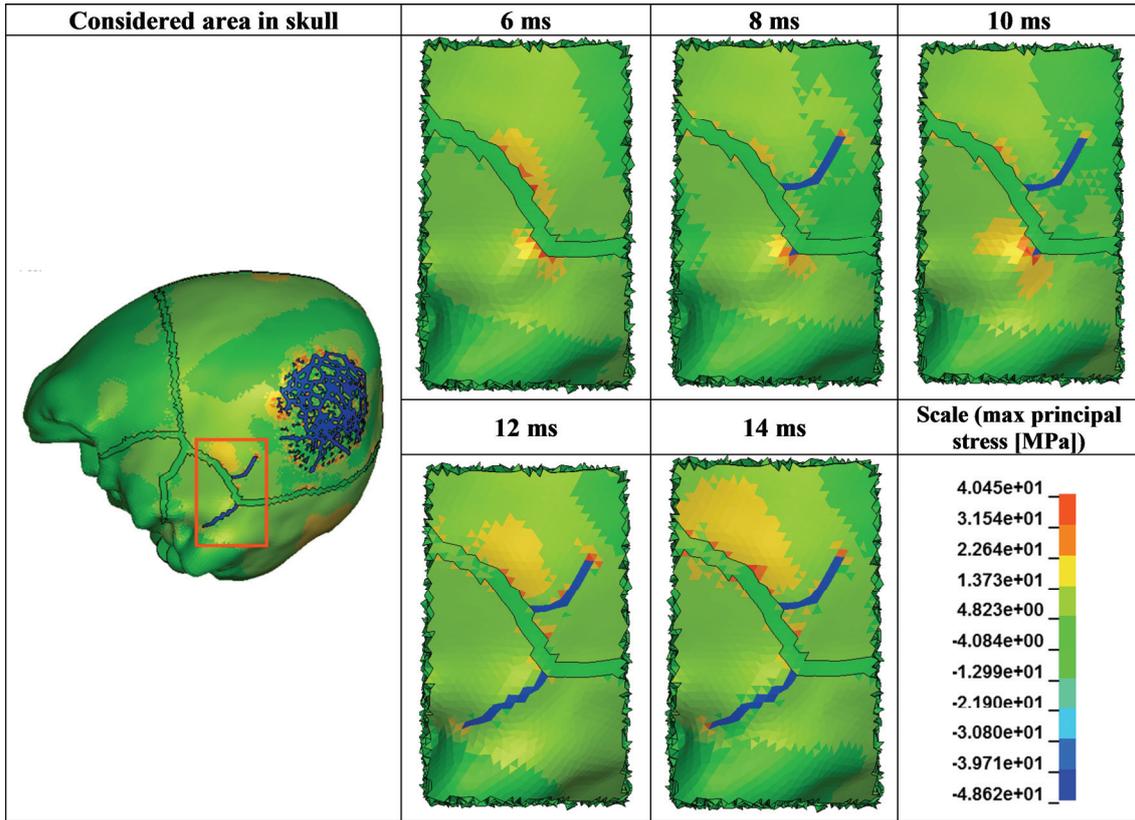


Fig. 6. Crack growth in periosteum layer for $E = 300$ MPa suture configuration

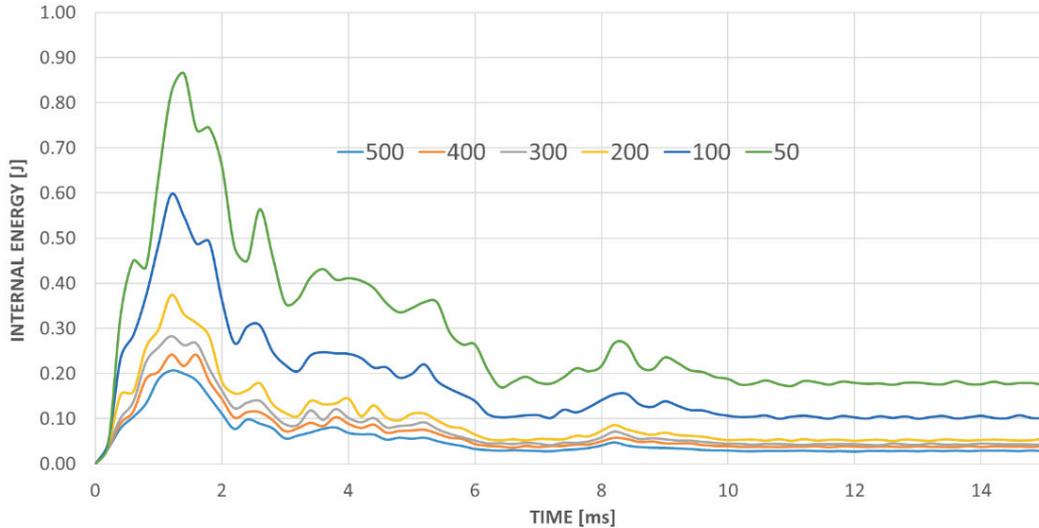


Fig. 7. Internal energy for cranial sutures with assigned Young's modulus within 50–500 MPa

Hence, cranial sutures have higher energy storage capabilities than skull bones [8]. As a result, this effect allows the cranial suture to act as a cushion in the skull [6]. However, a few studies are devoted to the effect of stiffness of cranial sutures on the amount of absorbed energy by them and the distribution of stresses in the bones of the skull. Therefore, in this research, we examined the stiffness of the sutures to

understand better the skull response under mechanical overload.

Moreover, the authors created a material model diagram for a better understanding of the complex mechanical phenomena on the suture interface. The model is depicted in Fig. 8, where m_i refers to mass of the particular concentrated mass, E_0 is the Young's modulus for elastic part (linear-elastic characteristic), E_{tan} , de-

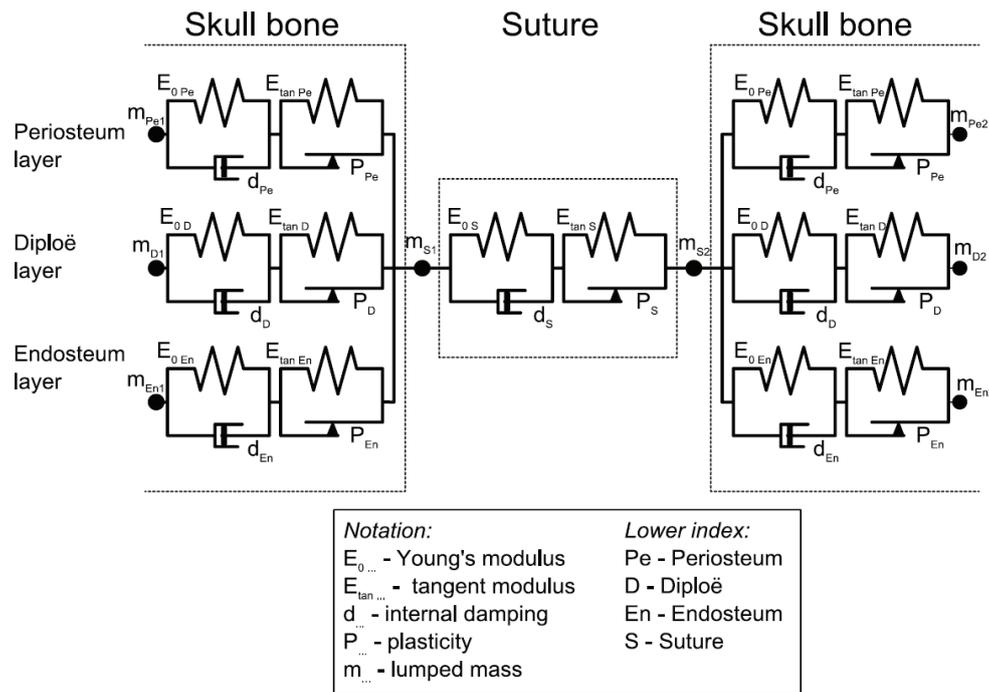


Fig. 8. Material model used in LS-DYNA for modelling the bones of the skull – the suture interface

describes the relation between stress and strain above yield stress, P is an ideal plastic model, whereas d is internal damping. This diagram shows adequately that for all tissues of skull a bilinear material model was used. For this study, the authors propose the model as shown in Fig. 8.

The results of our research have shown that the stiffness of cranial sutures plays an important role in the stress distribution and energy absorption. The increase in stiffness of cranial sutures caused more equal stress distribution across the parts of the skull. This effect can be seen on principal stress distributions for the periosteum. The analysis shows fracture initiation in the endosteum layer. It might be connected to different ultimate stresses for tension and compression. However, it should be noted that the model did not encompass the brain tissues. In fact, in humans, the brain can act as a springy base, which can affect the type of fracture in the endosteum. However, due to the complexity of the computational model, adding a highly deformable element – the brain – would significantly extend the computational time. On the other hand, a potential validation process would involve a prepared skull – which shall make the process more robust and feasible. This research was focused on simulating the proper crack initiation and growth in the skull bone. The element deletion implemented in LS-DYNA has a great potential, as strain rate can be taken into consideration. Moreover, XFEM – a finite element method for crack growth without remeshing,

which is available in Abaqus Standard module, is very convenient to study the crack growth. However, implementing the XFEM framework is regarded by authors as the future research.

5. Conclusions

The objective of this study was to examine the effect of cranial sutures on the skull's dynamic response. The authors focused on the assessment of skull breaking nature based on a vehicle-to-bicyclist traffic accident. However, in the state of the art there is still no consensus about the suture mechanical properties. In addition, most of the numerical head models do not have distinguished cranial sutures. Thus, the authors try to address these issues through comparison of the influence of different elastic properties of cranial sutures on the nature of skull fracture.

The research results showed that as the stiffness of cranial sutures increases, the energy absorbed by them decreases. Storing the largest energy in the cranial sutures with the smallest Young's modulus affected the largest amount of concentration of the principal stresses located perpendicular to the cranial sutures in the periosteum. Consequently, the higher the stiffness of the cranial sutures, the lower the risk of periosteum fractures around the cranial sutures. All of this points to the fact that the cranial sutures are a kind of dila-

tion. This finding has been confirmed in the literature [8], [9], which describes that cranial sutures are bio-composite structures that transmit loads and absorb the impact energy. Thus, the hypothesis of this paper – that mechanical properties of the suture may play a significant role in skull fracture nature – has been verified and confirmed.

The progress in the biomechanical research, combined with the use of advanced algorithms and numerical models, has helped the authors to improve current knowledge about cranial sutures. Nevertheless, in order to comprehensively understand the biomechanics of the skull structure as a result of the interaction of external forces, there is a further need to address this scientific problem, in particular, with the inclusion of computer models in micro- and meso-scales, which take into account the directionality of bone adhesions. At the same time, it is important to test the geometrical complexity of cranial sutures. The results obtained showed an important role of the sutures in numerical models as well as their significant influence on the biomechanics of skull fractures caused by dynamic loads. This knowledge is the foundation for the development of head protection systems and can also be important for neurosurgery and biomedical areas.

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