Biomedical Engineering for Medical Assistance “Stress and Acceleration Analysis for Damage Half-skull as A Consequence of Concrete Impact on a Protective Helmet” (Mechanical and Computer Engineering)

Houneida Sakly1,*, Mourad Said2, Moncef Tagina1
1COSMOS Laboratory -National Institute of Computer Sciences - Campus University of Mannouba –Tunisia
2Radiology and Medical Imaging Unit, International Center Carthage Medical, Tourist Area "JINEN EL OUEST"-5000 Monastir-Tunisia

*Corresponding author’s email: houneida.sakly@esiee.fr

Abstract. The dynamic analysis of fluid mechanics CFD applied to the medical field is one of the most important structural problems. They impose additional mechanical constraints on the construction of the 3D model to be processed and reduce its performance. As the acceleration forces of the concrete mainly excite these contacts between the skull and the Helmet, they could taken into consideration in this proposed system (helmet, concrete, and skull). Our aim is to evaluate the stress measurements in an accidental half-skull 3D with the computer fluids dynamics (CFD) tools basing on the mechanical and physical properties of the wall the skull. The second contribution is to analyze the acceleration of a concrete and the impact in the stress evaluation on half skull with and without helmet.

Keywords. CFD tools; skull; helmet; concrete; stress measurement

1. Introduction
The cranial structure is a complex region, with many vessels and vital structures of the cervix, which must be preserved during our scenario [1-3]. The three-dimensional reconstruction of cranium serves to highlight the main anatomical relationships with vital structures in all dimensions [4]. The reconstruction of the 3D skull model has become a fundamental task for the practical [5-7]. Occupational accidents are considered among the major public health issues and evaluated as one of the leading causes of mortality and morbidity. The total number of fatal occupational accidents registered a sharp decrease of 12.2% compared to 2015 and it is still the construction sector that has the highest rate of fatal accidents since they represent 30.7% of all fatal occupational accidents in Tunisia [8]. The contribution of this paper is to demonstrate the impact of CFD analysis which serves to offer considerable potential for studying stress on the body of the skull. In particular, the problem of designing a protective helmet is designed to protect the skull from a direct hit. At this preliminary stage, the reconstruction of the geometry for the crash is modeled for the skull [9-10] is considered sufficient for the immediate purpose.
Furthermore, this concept aims to analyze what properties of the helmet are needed to prevent skull fracture [11] as well as an estimate of concrete acceleration.

However, a more refined but more accurate modeling that could be done was needed. This analysis makes it possible to examine and test the effect on the internal organs as well as on the skeleton and to underline the cases where the impact was not normal on the surface of the helmet. The modern helmet design is composed of a variety of shapes combined with mixed materials. This choice of shape depends primarily on aesthetic reasons, but the reasonable choice of materials with the appropriate properties is crucial to achieve attenuation of stress wave transmission after impact. At this stage, the protective helmet measures must be included in the standards concerning the unpaid rights that must be established. Its success could be examined in detail. The benefits brought to the operation of CFD modules is that the design may be in a state of dysfunction but that the reasons for its success can be examined in detail. In addition, the method could be extended so that modeling contains a combination of the helmet of the skull and the concrete so that the experimental attempts will be useful [12-14]. In this paper, a model of anisotropic behavior of the helmet was proposed, skull and concrete to simulate the acceleration of contact and stress and measurements.

2. Methods and materials

2.1 Methods

Two 3D models was performed:

-Skull without and with Helmet: a Tet-Dominant algorithm was executed with Automatic grading Mesh, moderate Fitness and 8 processors for parallel processing. The skull point was fixed according to the three axes: x = 0.02017m, y = 0.02516m, z = 0.583m. The physical contact is conFigd for Newtonian nonlinearity and 0.0001 iteration and no friction criterion. The Convergence stabilization and smoothing contact are checked in standards. The wall and skull for both of two model contact are resolved by the Augmented Lagrange method with a coefficient = 100. Cranial structures can be identified in augmented reality. The purpose of this study was to analyze the adapted mechanical behavior with the isotropic linear elastic law for a biomechanical model of skull with and without Helmet. A segmented 3D model of (skull, Helmet, Concrete) was generated. The quantitative evaluation of the deformation between the reconstructed model and the deformed model was assessed by the Hausdorff distance and qualitatively by the identification of 3 anatomical landmarks. The presentation of a constituent material law for linear viscoelasticity requires a relaxation formulation as a function of time for complete anisotropy. Both the elastic and viscous properties are anisotropic, and each element has a relaxation tensor.

2.2 Materiels

The description of the material is summarized in table 1, which follows for both models:

Table 1. Material proposal

| Isotropic linear Elastic behavior, Isotropic directional dependency | Young’s modulus (Pa) | Poisson’s ratio | Density (kg / m³) | Description |
|---|---|---|---|---|
| Skull | 40000000000 | 0.2 | 2240 |
| Concrete | 60000000000 | 0.2 | 3000 |

Second Model with Helmet

| Isotropic linear Elastic behavior, Isotropic directional dependency | Young’s modulus/Pa | Poisson’s ratio | Density (kg / m³) | Description |
|---|---|---|---|---|
3. Results and discussion
The helmet has been designed with a simplified form. The requirement of symmetry must be verified. The design of the half answers the localized cause for the model limited to a quarter of the helmet. The model consists of three layers: the outer surface of the hard helmet, the quasi-foam filling material and the skull. In this preliminary study, the mesh for both of two model: skull without and with Helmet in Figure 1 were made around the deformation reflection to have a classification introduced in a coherent model.

![Figure 1. (A) Mesh of the skull 3D without Helmet (B) Mesh of the skull 3D with Helmet](image)

The generation of the Mesh for the first model without Helmet was refined with local element size technique for the skull inner and outer surfaces and the whole wall with global elapsed time 4.9 seconds so such the second model takes 2.7 seconds. The initial conditions are configd as follows: displacement, acceleration, velocity, and constraints. Global conditions are applied to fix the wall. The inner and outer surfaces of the filling material are constrained with the inner and outer surfaces of the skull, respectively.
The distribution of the direct stress component in the Y direction and the effective stress values of Von Mises is shown in the Figure 2 and 3 for both of models.

**Figure 2.** Von Mises Stress for the model of skull without Helmet

![Von Mises Stress for the model of skull without Helmet](image1)

**Figure 3.** Von Mises Stress for the model of skull with Helmet

![Von Mises Stress for the model of skull with Helmet](image2)

A comparison of the actual stress values with the static yield strength for the helmet material in both models should be established. Considerable performance was seen with the peak level on the outer surface away from the point of impact with $7 \times 10^8$ (N/m²) for the helmet-free skull model and maximum stress of $1.75 \times 10^8$ (N/m²) for the model with Helmet.
The examination of the difference in the level of effective stress at the level of the skull compared to that of the external surface in the two designs. Figure 4 shows the impact of acceleration on contours for the skull surface at the time of $2 \times 10^{-3}$ (s) and $1.25 \times 10^{-3}$ (s). In this case, it is significant that the behavior of the helmet presents a safety element and protection: the stress decreases for the model with helmet and increases in the opposite case and remains completely relative to the acceleration of concrete what would be desirable from the point of view of the prevention of skull fracture.

From a parallel processing point of view, the first skull model without Helmet requires a total CPU time = 1346.29 s and the second model requires a total CPU time = 11427.33 s. The results simulated in Table 2 show that the first design is greedier in terms of execution time and memory consumption.

Table 2. Details of parallel processing and the cost of memory

| Skull without Helmet | |
|----------------------|------------------|
| Statistics on all transient time with CPU | Dynamic memory allocation and storage |
| Total CPU Time 1346.29 s | Percentage use of the directory: 59% |
| Total CPU user time 1203.41 s | Percentage of use of the directory: 57% |
| CPU total time 142.88 s | The memory requested at launch is overestimated: 13920.00 MB. |
| Time remaining CPU 35994153.71 s | The peak used memory: 4354.23 MB |

| Skull with Helmet | |
|-------------------|------------------|
| Statistics on all transient time with CPU | Dynamic memory allocation and storage |
| Total CPU time 11427.33 s | Percentage use of the directory: 83% |
| Time CPU user total 10907.88 s | Percentage of use of the directory: 10% |
| CPU time total system 519.45 s | The memory requested at launch is overestimated: 29200.00 MB. |
| Time remaining CPU 35984072.67 s | The peak used memory: 4957.89 MB |

As predicted in advance, the direction of impact occurs via the maximum displacements after the outer layer has lost the acceleration imparted immediately after the impact. The movements of the concrete or the node in the middle of Helmet and the node located at the center of the skull are perfectly

Figure 4. (A) Simulation of the acceleration for the skull without Helmet (B) Simulation of the acceleration for the skull with Helmet
adapted during the monitoring period and which indicates consequently that the outer part of the filling layer absorbs the energy of the contact impact.

Subsequently, the behavior at longer times manifests in Figure 5 and 6 the residual vibration resulting from the elasticity of the materials.

![Figure 5. Simulation of the residuum of the model of skull without Helmet](image)

![Figure 6. Simulation of the residuum of the model of skull with Helmet](image)

In particular, it is important to emphasize the role of reinforcements in the anti-shock helmet and skull layers of varying orientation. When a stress wave coming from a contact object (concrete) considered an impact point meets one of these layers, its propagation will be deflected [15].

4. Conclusion

In conclusion, the design of the protective helmet is considered at this stage satisfactory while taking into consideration the accuracy of the reality of the prevailing conditions. Therefore, a proposition of a perspective of our research method to extend this vision to include the brain with other organs, such as the heart, kidneys or lungs or other parts of the skeleton that could be modeled by an equivalent approach.

References

[1] Opperman L A, 2000 Cranial sutures as intramembranous bone growth sites vol 219 no 4 (Dev. Dyn) pp 472–485.

[2] Zhang Z Q and Yang J L, 2015 Biomechanical dynamics of cranial sutures during simulated impulsive loading vol 2015 (Appl. Bionics Biomech).
[3] Herring S W and Ochareon P, 2005 Bone – special problems of the craniofacial region vol 8 no 3 (Orthod. Craniofac) pp 174–182.

[4] Bonne N X, Dubrulle F, Risoud M, and Vincent C, 2017 How to perform 3D reconstruction of skull base tumours vol 134 no 2 (Eur. Ann. Otorhinolaryngol. Head Neck Dis) pp 117–120.

[5] Jain A K, Klare B, and Park U, 2012 Face matching and retrieval in forensics applications vol 19 (IEEE Multimed) p 20.

[6] Gao Y, Wang M, Tao D, Ji R, and Dai Q, 2012 3-D Object retrieval and recognition with hypergraph analysis vol 21 no 9 (IEEE Trans. Image Process) pp 4290–4303.

[7] Song M, Tao D, Huang X, Chen C, and Bu J, 2012 Three-dimensional face reconstruction from a single image by a coupled RBF network vol 21 no 5 (IEEE Trans. Image Process) pp 2887–2897.

[8] SST E, 2018 Tunisia: work accidents, occupational accidents & occupational diseases - year 2016 - the worker’s gazette (CNAM via ISST, Tunisia, Statistical Investigation).

[9] Libby J, Marghoub A, Johnson D, Khonsari R H, Fagan M J, and Moazen M, 2017 Modelling human skull growth: a validated computational model vol 14 no 130 (J. R. Soc. Interface).

[10] Rafferty K L and Herring S W, 1999 Craniofacial sutures: morphology, growth, and in vivo masticatory strains vol 242 no 2 (J. Morphol) pp 167–179.

[11] Hajiaghamemar M, Lan I S, Christian C W, Coats B, and Margulies S S, 2019 Infant skull fracture risk for low height falls vol 133 no 3 (Int. J. Legal Med) pp 847–862.

[12] Kormi K and Etheridge R A, 1992 Application of the finite-element method to simulation of damage to the human skull as a consequence of missile impact on a multi-layered composite crash helmet vol 14 no 3 (J. Biomed) pp 203–208.

[13] Mall G, Hubig M, Koebske J, and Steinbuch R, 1997 Finite-element modeling of the human neurocranium under functional anatomical aspects vol 179 no 4 (Ann. Anat. - Anat. Anz) pp 303–309.

[14] Smith T, Lenkeit J, and Boughton J, 2005 Application of finite element analysis to helmet design (IUTAM Symposium on Impact Biomechanics: From Fundamental Insights to Applications) pp 255–262.

[15] Li B W, Zhao H P, Qin Q H, Feng X Q, and Yu S W, 2012 Numerical study on the effects of hierarchical wavy interface morphology on fracture toughness vol 57 (Comput Mater) pp 14–22.