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# COMPARATIVE ANALYSIS OF IMPACT ABSORPTION CAPACITY OF FOOTBALL HELMETS

**Viezzer, L. R.[1], Ferreira, A. P. C. S. [2], Mendonça, C. J. A.[3], Olanyk, L. Z.[4]**

[1] Federal University of Technology – Parana (UTFPR), lucasviezzer@gmail.com

[2] Federal University of Technology – Parana (UTFPR), apaula@utfpr.edu.br

[3] Federal University of Technology – Parana (UTFPR), cjamendonca@yahoo.com.br

[4] Federal University of Technology – Parana (UTFPR), olanyk@utfpr.edu.br

*Abstract: The sports industry and medicine are constantly seeking to develop new impact-absorbing technologies in order to reduce the incidence of head injuries. The traditional football helmet models, widely used, have the concept of an external hard shell, with a padding of polymers, focused on energy absorption. An American startup from Seattle developed a completely new concept that consists of an internal hard shell, an external soft shell and an interface of columnar elements. In situations of impact, the outer layer deforms elastically at the point of contact, while the interface structure deforms through buckling and bending mechanisms, and then return to their original shape, dissipating a larger portion of energy and decreasing the peak impact force. Experimental tests conclude that the new concept is the safest to prevent injury. In this context, the present research seeks to qualitatively, through finite element models, validate the results obtained. Initially, the properties of the materials (polycarbonates, foam and rubber) and other important information for each helmet are defined, based on publicly available models. The software Abaqus Explicit is used, developing a numerical model from the experimental system used by the Virginia Tech University. The resulting acceleration data is taken from nodes on the innermost surface of the helmet model, in the region where impact occurs. The literature presents similar graphs for the resulting acceleration, and it can be concluded that there is a qualitative validation for the results obtained in this work. It is verified that the new helmet concept presents a resulting acceleration 70% lower when compared to the traditional concept. The values for the Brain Injury Criterion (BIC) calculated are 0.0047 for the traditional model and 0.0018 for the new model. Thus, the new model has a significantly lower criterion than the traditional one, proving to be less susceptible to brain injury.*

***Keywords:*** ***football helmet, head injury, impact, energy absorption, finite elements***

## INTRODUCTION

Sports helmets, especially football helmets, are designed to reduce the likelihood of brain injuries resulting from impacts to the head region, dissipating and distributing the applied energy and protecting the head from possible fractures and/or neurological damage. This growing concern to reduce the number of concussions and other brain injuries in American football athletes constantly contributes to the development of new technologies applied to the structure of helmets. In this work a comparative analysis is made between a traditional helmet model and one with an innovative design.

The traditional model considered in this work is the Riddell™ (Fig. 1), consisting of an outer shell, preferably made of plastic material, which meets the requirements for the resistance to tension and durability characteristics suitable for sports helmets, such as polycarbonates (Lexan™), and an impact absorbing coating, fixed to the inner wall of the shell, normally through the Velcro® system. This system increases the duration of the impact pulse and, thus, decreases the mechanical load felt by the brain (Viano, Casson and Pellman, 2007). Generally, each lining element has one or two inner layers of padded material, and is encapsulated by plastic material, and may or may not be inflated with pressurized air. The padding materials typically used are vinyl-nitrile (VN) foams, expanded polypropylene (EPP) or thermoplastic polyurethane (TPU) (Patzin, 2014). When more than one layer is used in the structure, it is recommended that the layer closest to the inner surface of the shell be a stiffer foam to attenuate the energy, while the layer closest to the athlete's head be a more malleable foam to provide comfort. The combination of these materials with pressurized air contributes to the best fit of the helmet on the user's head. For the forehead, a padded material made entirely of energy-absorbing foam is used, preferably vinyl-nitrile or polyurethane.

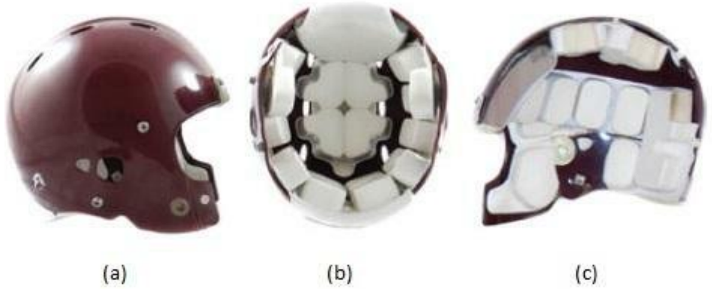


**Figure 1 – Riddell's shell and padding in views (a) lateral; (b) bottom and (c) sectional. Adapted from Rowson et al (2014)**

The startup Vicis, with the aim of creating a football helmet safer in relation to traumatic injuries, mainly concussions, created an innovative model of helmet. The previous or traditional technologies aim to absorb linear incident forces, transmitting the peak force of the impact to the user's head, while the new model is associated with helmets with non-linear deformation elements, represented by a columnar structure, which deform according to the force incident on impact (Browd et al., 2016). This structure is positioned at the interface between two shells, one external and one internal, with different properties (Fig. 2). Both the outer and inner layers are relatively rigid, made of plastic material. The outer layer, made of polycarbonate, allows local deformations when subjected to impact loads, while the inner layer is made of a polycarbonate up to five times stiffer than the outer layer, in order to prevent intense impacts capable of causing fractures and bruises in the athlete's head.


**Figure 2 – Vicis helmet: outer shell (Lode Shell®), interface (Vicis Rflx®) e inner shell (Arch Shell®).**

**Source: Vicis (2018)**

The interface structure between shells consists of thin, elongated, columnar elements, which deform non-linearly and act against the incident forces on the helmet. These structures can have high aspect ratios, from 3:1 to 1000:1. This non-linear deformation aims to provide an improvement in protection against the resulting linear accelerations or decelerations, but also against rotational ones. These elements are configured to buckle when a load is applied, that is, a sudden failure of the elements when subjected to a compressive stress, this stress being less than the maximum compressive stress allowed by the material, deforming elastically and returning to its initial condition after the load application. The flexibility of the filaments also allows the outer shell to move relatively to the inner shell. These filaments can be manufactured with foams, elastomers, polymers and a combination of these materials.

## BIOMECHANICS OF IMPACT

Research developed regarding the biomechanics of injuries and other risks associated with impact loads on the human body are conducted in order to determine their severity. In order to study the tolerance of the human body to these loads, criteria are established to analyze shocks and establish the limit supported by vital parts, such as the head and trunk. This paper aims to study impacts in the head region, and therefore, addresses the criterion for brain injuries. These injuries are recognized as the most harmful in crash and accident situations. Human tolerance for these impacts is represented by a graph called the Wayne State Tolerance Curve and is used to define the level at which acceleration, or deceleration of the head, results in concussions or skull fractures (Johnson and Mamalis, 1978).

The Wayne State Tolerance Curve is used as the basis for several injury severity injuries. For brain injuries, the best known is the Gadd Severity Index (GSI) (Johnson, Skorecki and Wells, 1975). The effective acceleration ($a$) as a function of the gravity acceleration ($g$), is integrated with respect to time ($t$) in microseconds (ms), where $T$ is the total time of impact duration. This duration must be between 0.25 and 50 milliseconds, the interval for which the equation is developed and validated. The value of 1000 represents the threshold for serious internal injuries in a frontal impact. Therefore, if the GSI is less than this value, the impact is tolerated by the body without permanent damage to the brain or skull of a normal adult.

$$GSI \equiv \int_0^T \quad a^{2,5} dt < 1000 \tag{1}$$

Subsequently, the GSI is replaced by the Brain Injury Criterion (BIC), considered the best indicator for brain injury available (Chou, Howell, and Chang, 1988; Zhou, Thomas and Stibich, 1998). The BIC also considers a threshold value of 1000 for non-permanent injuries, but the index is analyzed for a time interval between $t_1$ and $t_2$, where the index has a maximum value and its respective resulting acceleration ($a(t)$).

$$BIC \equiv (t_2 - t_1) \left(\frac{1}{t_2 - t_1} \int_{t_1}^{t_2} \quad a(t) dt\right)^{2,5} < 1000 \tag{2}$$

In football, three types of impacts are considered: helmet-to-helmet, helmet-to-ground and helmet-to-body. Viano, Casson and Pellman (2007) analyzes approximately two hundred impacts among professional football athletes, concluding that 60% of impacts are of the first type, and that 70% occur in the helmet shell region. For cases where brain injuries are found, 88% of them occurred in helmet-to-helmet impacts.

For the present work, it is necessary to define performance comparison criteria between the analyzed helmets. In this way, the numerical simulation will provide graphs of the maximum resulting acceleration in the internal part of each model and the BIC (Brain Injury Criterion) will also be calculated.

## NUMERICAL MODELS OF HELMETS

### Material properties

The properties of the materials that make up each helmet and other important information for the present model are defined from publicly available models to serve as a basis for studies regarding the prevention of injuries and the development of better helmets.

Two materials are used in the proposed structure for the first helmet model studied. The shell material is polycarbonate (PC) and the padding is foam. In addition to basic properties for defining the elastic behavior of the material (Tab. 1) specific elastic properties are needed for the foam, resulting from a stress-strain curve (Fig. 3) of the material for a specific strain rate. The colored lines represent the result of each experimental test (compression, tension and shear) performed to define the foam materials used in a real helmet.

**Table 1 – Material properties. Adapted from Fahlstedt et al. (2018)**

| Property | Riddell® Polycarbonate (PC) | Foam |
|---|---|---|
| Density (kg/m$^3$) | 1095 | 170.5 |
| Young's Modulus (GPa) | 1.565 | 0.003 |
| Poison Coefficient | 0.4 | 0 |

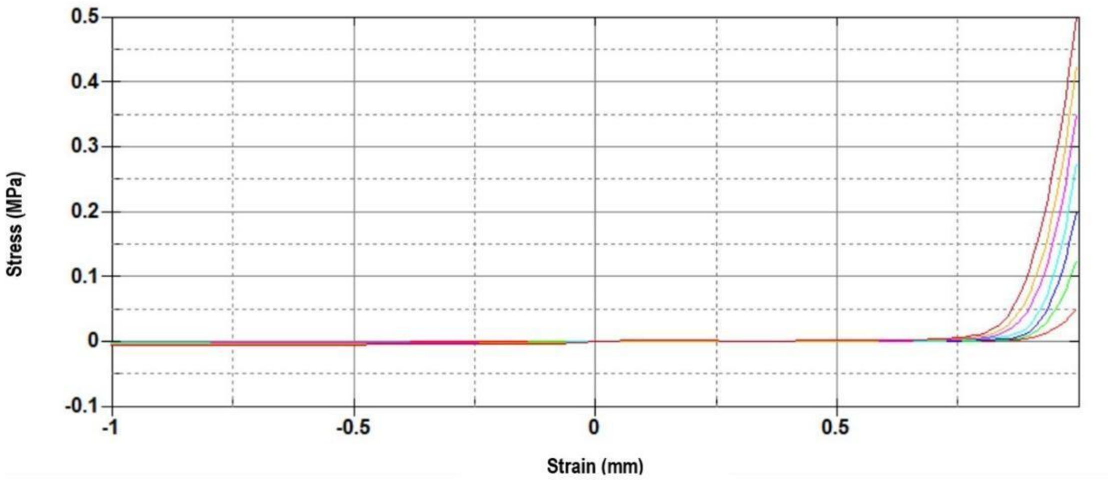


**Figure 3 – Stress-strain curve for a strain rate of $10^{-6}$ $s^{-1}$. Adapted from Fahlstedt et al. (2018)**

For the second model, three materials are defined. The two shells are made of polycarbonate with different properties (Tab. 2). The interface material between the shells has some specific properties (Tab. 3). In order to be well characterized in the simulation, shear test data is needed to determine the dimensionless shear stress relaxation modulus of the material (*gR*) with respect to time (Fig. 4). This modulus is obtained from the ratio between the shear stress relaxation modulus at a given instant *t* (*Gr(t)*) and the instantaneous shear modulus ($G_0$) of the material. Data is adapted based on Abaqus commercial software user manuals and tutorials.

**Table 2 – External/internal shell PC properties: Adapted from Giudice et al. (2018)**

| Property | External shell polycarbonate (PC) | Internal shell polycarbonate (PC) |
|---|---|---|
| Density ($kg/m^3$) | 969.6 | 991.9 |
| Young's Modulus (GPa) | 0.55 | 2.75 |
| Poison Coefficient | 0.4 | 0.4 |

**Table 3 – Interface columns properties. Adapted from Giudice et al. (2018)**

| Property | Value | Unit |
|---|---|---|
| Density ($kg/m^3$) | 1160 | kg/m3 |
| Volumetric module (GPa) | 0.05 | GPa |
| Instantaneous shear modulus ($G_0$) (GPa) | 0.028 | GPa |
| Infinitesimal shear modulus ($G_\infty$) (GPa) | 0.0105 | GPa |

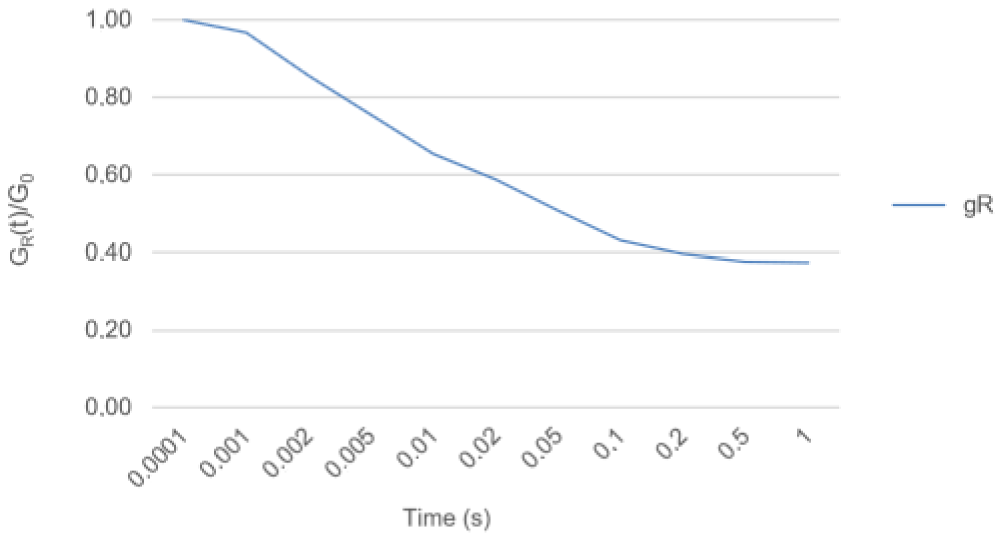


**Figure 4 – Shear-stress relaxation test curve for a viscoelastic material. Adapted from Dassault Systèmes (2013)**

In both models, it is necessary to understand how materials behave elastically, in order to define the necessary properties and the best way to insert them into the software used. Due to the complexity of the materials used and their relevance to obtain coherent and significant results, a greater detail of this step can be found in Viezzer (2019).

## Impact test simulation

In order to simulate real experiments, a numerical test model is developed from the system used by the Virginia Tech University in 2018 (Fig. 5), as it is a more reliable method for reproducing and repeating the tests in relation to other impact tests (Pellman et al., 2006).

The experimental system consists of a pendulum with a length of 1905 mm and a total mass of 37 kg, including a 15.5 kg spherical calotte at the end of the pendulum, with a diameter of 203 mm, and a radius of curvature equivalent to 127 mm and is positioned at 15º to the vertical axis. The pendulum strikes the helmet fixed below its axis of rotation, with an inclination of approximately 10º in relation to the horizontal plane and positioned at 25 mm in the negative direction of the y-axis of the coordinate system. The total moment of inertia of the pendulum is 72 kg.m$^2$, of which the spherical calotte accounts for 78% of the value. The pendulum is only allowed to move with angular velocity, without translation in the axis of rotation, and it hits the helmet with an equivalent linear velocity of 3 m/s. The coefficient of friction considered is 0.1 and the simulation duration is 200 ms.

**Figure 5 – Impact pendulum used in experimental tests by Virginia Tech University. Source: Rowson; Tyson (2018)**

The spherical calotte is considered a rigid object, so only the mass and the moment of inertia are necessary for its characterization. To simplify the model, only the spherical calotte is considered in the system since it is the only structure of the pendulum that participates in the contact with the helmet. Rectangular discrete rigid elements of 4 nodes (R3D4) are used. The number of nodes per element is suggested by the software based on the element type considered for each structure. The mesh is then defined according to the approximate global size of the element (AGS), a dimensionless parameter used by Abaqus.

## Traditional helmet model

The traditional helmet model is represented by an outer shell 4 mm thick and 150 mm in diameter and a solid inner shell 25 mm thick (Fig. 6). These dimensions are extracted from a real helmet model. The structure is positioned at the origin of the software Cartesian system, so that it can be reached by the pendulum spherical calotte. The shell material is characterized as elastic and isotropic, and its properties refer to the polycarbonate described in Tab. 1. The padding foam is characterized as low-density foam, in addition to being elastic and isotropic, and therefore, the properties of Tab. 1 and the uniaxial stress and compression test data correspondent to the Fig. 3 are used.

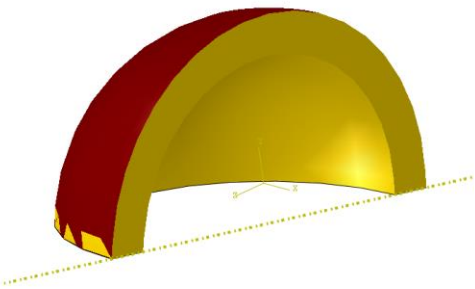

**Figure 6 – Sectional view of traditional helmet model**

The boundary conditions applied to the helmet structure is the fixation of the inner perimeter of the padding, and it remains fixed during the analysis. The outer surface of the padding in yellow (closest to the pendulum) is attached to the inner surface of the shell in red through the tie constraint, imposing the same displacement to both structures after the load is applied. The model creation is carried out through Abaqus/Explicit. The helmet shell is formed by S4R shell elements, recommended for thin thickness geometries. This element is defined by an average surface, composed of 4 nodes with 6 degrees of freedom. Solid elements (C3D10M) are used for the padding, in the form of a modified tetrahedron with 10 nodes, as it is considered a solid. Both parts are simulated with AGS of 10.

## Helmet model with columnar structures interface

The model with columnar structures interface is represented by an outer shell of 260 mm in diameter, an inner shell of 200 mm in diameter, both 4 mm thick, and an interface layer of 30 mm thick, represented by a series of 5 mm diameter columns evenly distributed between two thin shells (approximately 0.5 mm) of the same material (Fig. 7). These dimensions are extracted from a real helmet model.

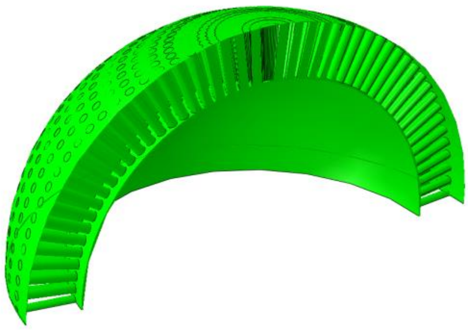

**Figure 7 – Section view of interface layer**

The shells material is characterized as elastic and isotropic, with the properties described in Tab. 2. As it is a material with complex characterization and aiming at a qualitative validation of the numerical model, a simplified viscoelasticity analysis is performed for the interface, through the properties presented in Tab. 3 and in Fig. 4. The infinitesimal shear stress relaxation modulus is used as a limiting value in the simulation, obtained from the ratio between $G_\infty$ and $G_0$, also dimensionless.

The inner shell perimeter is fixed and remains in this way during load application. The interface layer is attached to the outer and inner shells through the tie constraint, imposing the same displacement to all structures during the simulation. The model creation is carried out through Abaqus/Explicit for the shells, however, due to the complexity of the interface structure, SolidWorks software is used for its modeling. The mesh is formed by S4R shell elements for both helmet shells and 3-D tension elements (C3D8R), represented by a parallelepiped with 8 nodes, for the columns. All structures are defined with an AGS of 10.

The resulting acceleration data is taken from nodes on the innermost surface of the helmet model, in the region where the pendulum impact occurs. From these graphs, the node that has the greatest acceleration is chosen to represent the magnitude of the maximum acceleration that reaches the athlete's skull in a real impact situation. From the data extracted from the simulations, it is possible to conclude, in a qualitative way, which structure is more efficient in absorbing energy and reducing the resulting acceleration.

## RESULTS

From the simulations carried out with both helmet structures, it is possible to obtain the graph of maximum resulting acceleration in the inner part of each model, for the minimum time interval available in Abaqus of 0.01 seconds. It is observed in the acceleration-time graphs (Fig. 8) that the new structure has a resulting acceleration with a value 70% lower than the traditional structure. This difference is largely due to the difference in design between structures: while the traditional one seeks to distribute the force applied at the point of impact to the entire shell and then use the padding to reduce this energy that causes the sudden acceleration of the brain, the new model seeks to absorb most of the energy applied directly at the point of impact, deforming the outer shell (more malleable) and the interface layer.

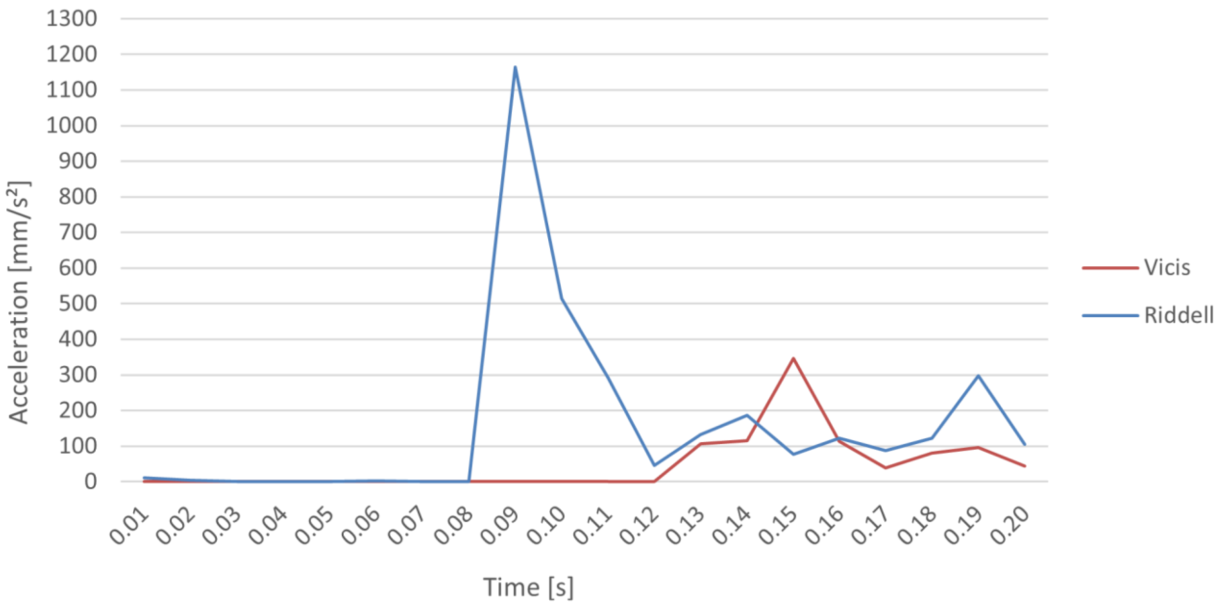


**Figure 8 – Acceleration-time curve of both helmets for a 3 m/s impact velocity**

Both models are simulated under the same conditions, so the stress to which the structures are subjected is the same. According to Hooke's Law applied to materials, stress is directly proportional to strain, related through the modulus of elasticity. For the same stress applied to two different materials, the one with the lowest modulus of elasticity will allow greater deformations. According to Tab. 1 and Tab. 2, the outer shell of the new model has a lower modulus of elasticity compared to the Riddell®'s shell, therefore allowing greater deformation. This deformation generates a displacement of the structure at the point of impact. As the displacement of the outer shell and the interface layer is greater compared to the displacement of the traditional shell, the portion of energy absorbed is, consequently, greater for the same force. In this way, the peak force reaches the inner shell (more rigid) with a lower value than the initial one, resulting in a low resulting acceleration, reducing the risk of injury.

Another relevant point is the time lag between the peak acceleration between the models. Note that for the second model, the peak acceleration takes 75 ms longer than the traditional model. This difference is due to the longer time the pulse needs to reach the inner surface of the helmet after having lost energy in the previous layers.

The values for the brain injury criterion (BIC) calculated are 0.0047 for the traditional model and 0.0018 for the new model. Both are far below the threshold of 1000 for permanent injuries, but it is noted that the new model has a criterion 60% lower than the traditional one, suggesting to be less susceptible to brain injuries.

For these reasons, it is considered the new model the safest to be used in the practice of sport, in relation to the prevention of possible injuries. Zhang et al. (2003) presents similar graphs for resulting acceleration in comparative studies between experimental tests and numerical models for different impact tests. Rowson and Tyson (2018) reach similar conclusions in experimental tests with the helmets and the pendulum used in the numerical model of this paper.

For model validation, the experimental study Rowson and Tyson (2018) is used. The method consists of obtaining data from pendulum impacts in different helmet models and concluding about the safest helmet based on the STAR index. It is possible to observe that the methodology used in the present work is sufficient to represent the test performed, since the conclusion about the best helmet model is similar to the conclusion obtained by Virginia Tech University and the STAR index.

After the validation of the model, it is necessary to understand how the impact velocity influences the behavior of structures. The resulting acceleration results for a pendulum of 4.6 m/s are shown in Fig. 9. The speed influences the acceleration that reaches the inner part of the helmet in both cases. Note that when the speed is increased by approximately 50%, the resulting acceleration has an increase of only 15% on average, for both structures, with the new model maintaining a 70% lower acceleration compared to the traditional one. In addition, for the same duration of impact, the peak acceleration on the inside occurs between 30 and 40 ms earlier, compared to the speed of 3 m/s, and after this peak, the new model structure ends up being more stable, with the acceleration gradually reducing, while the traditional structure ends up having other peaks of lesser intensity. For the BIC, the traditional structure has a value of 0.0041, while the new model reaches a value of 0.0016, remaining the helmet with the lowest risk of brain injury for the user.

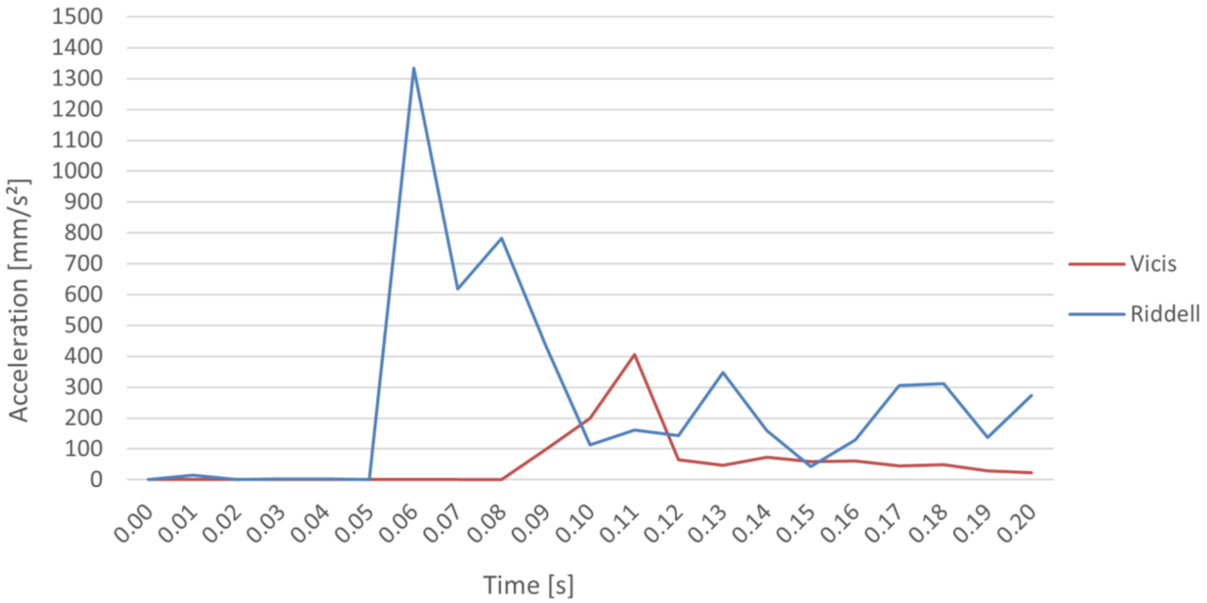


**Figure 9 - Acceleration-time curve of both helmets for a 4.6 m/s impact velocity**

The results for the highest impact velocity of 6.1 m/s are shown in Fig. 10. The resulting acceleration for the traditional model undergoes a considerable increase, of 82%, in relation to the results for the lowest impact velocity. The acceleration peak occurs 10 ms earlier compared to the 4.6 m/s impact and then behaves similarly to the previous case, with the resulting acceleration value oscillating with the lower amplitude. The BIC of the structure reaches 0.0054, the highest among the three cases, demonstrating that the impact velocity has an influence on the probability of brain injury.

The new model obtained an increase in the resulting acceleration, but less significant, with a 22% increase in relation to the simulations for the lowest speed and 80% less than the traditional model for 6.1 m/s. It can be said that the impact velocity has very little influence on the resulting acceleration for this model, compared to the influence for the traditional one. The peak duration period is similar to that of the acceleration peak for the speed of 4.6 m/s, but occurs almost 50 ms in advance. The curve starts to oscillate more after this instant. The BIC calculated for the impact at a speed of 6.1 m/s is 0.0017 and the helmet remains the safest model in prevention of brain injuries.

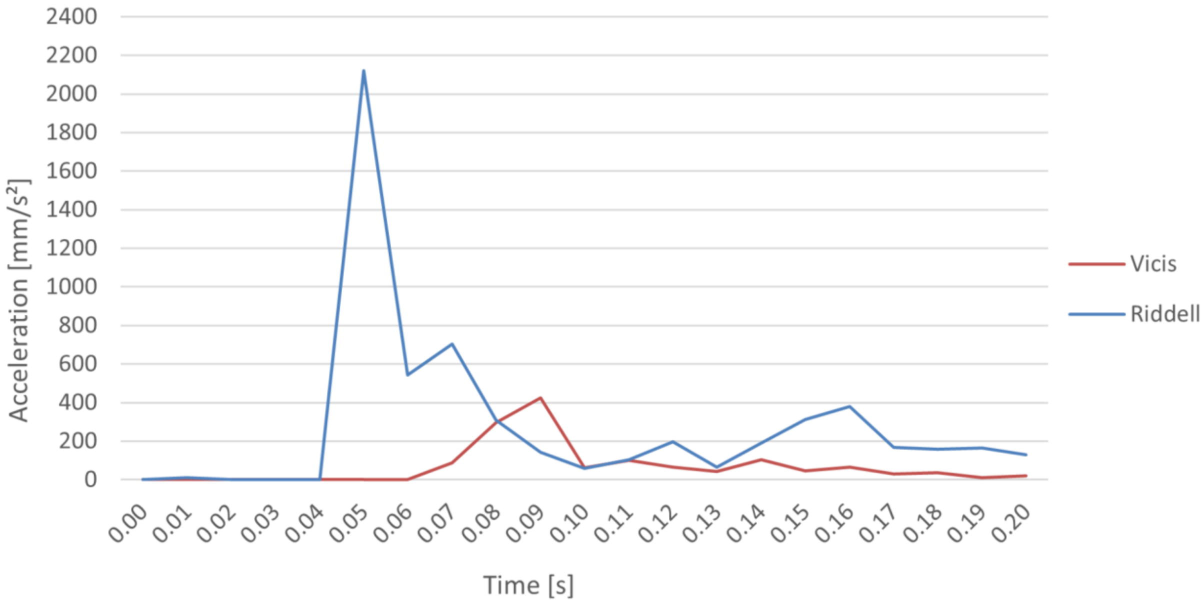


**Figure 10 - Acceleration-time curve of both helmets for a 6.1 m/s impact velocity**

## CONCLUSIONS

Football helmets are used to prevent brain injuries, using a characteristic shock-absorbing structure consisting of a rigid shell and an impact-absorbing padding. This combination aims to reduce as much as possible the peak force that reaches the athlete's skull during the impact, as well as the linear and angular accelerations, which are responsible for the occurrence of concussions and trauma.

A new model of structure is developed, modifying the traditional concept and placing the rigid shell on the inside of the helmet, and a more malleable shell on the outside, separated by an interface of columnar structures that deform in a non-linear way, which aims to check better impact absorption. In this context, this paper evaluates the energy absorption capacity and the resulting acceleration of the impact through numerical simulations of both structures, involving geometries close to the real model.

It suggests that the new helmet model is more efficient than the traditional one for shock absorption and reduction of the resulting acceleration in the athlete's brain. It is visible in the simulations that this is possible due to a new concept that applies a rigid shell on the inside of the helmet, and a malleable shell on the outside, allowing greater deformations at the point of impact, resulting in a greater displacement of the structure and, consequently, more energy absorbed. The BIC of both models is calculated, and the new model is shown to carry less risk in relation to brain injury, confirming previous studies of Rowson and Tyson, 2018. For higher impact velocities, it is observed that the behavior of both structures remains similar in relation to lower velocities, proving that Vicis model is more efficient.

For future work, it is suggested to carry out experimental analysis to improve the numerical results. In addition, the numerical model can be improved by more complete definition of properties, such as experimental analysis of the materials to be used. The study can also be expanded to a new concept of structures, more complex geometries and even other materials that have impact absorption characteristics.

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