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A Peak Virtual Power Concept to Compute Brain Injuries Associated with Concussion

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Abstract

Concussion can result from various sports collisions and can lead to the disruption of neuronal cell membranes and axonal stretching, leading to a neuro-metabolic cascade of molecular changes in the brain. There is currently no agreement on which numerical method can assess such low-level injuries. This paper demonstrates for the first time that Peak Virtual Power (PVP), based on the Clausius-Duhem inequality, assuming that the injury is represented by the irreversible work in a human body, could be a candidate to capture brain distortion related to concussion. Three NFL helmet-to-helmet impacts were reconstructed with finite element analysis, using validated helmet computer models fitted with calibrated Hybrid III headforms against linear and angular acceleration impact corridors. The Hybrid III head form was then replaced with a THUMS 4.02 human head model, in which the PVP was computed at the corpus callosum and midbrain locations. The results indicate that mild and severe concussions could be prevented for lateral collisions and frontal impacts with PVP values lower than 0.928mW and 9.405mW, respectively, and no concussion would happen in the head vertical impact direction for a PVP value less than 1.184mW. This innovative method proposes a new paradigm to assess brain injuries and helmet design.

Keywords: Concussion; Brain distortion; Peak Virtual Power; THUMS 4.02

Abbreviations: PVP: Peak Virtual Power; CTE: Chronic Traumatic Encephalopathy; NFL: National Football League; FE HBM: Finite Element Human Computer Models; MPS: Maximum Principal Strain; PMHS: Post-Mortem Human Subjects; DOE: Design of Experiment; LH: Latin Hypercube; RSM: Response Surface Method; GCS: Glasgow Coma Scale; BIRC: Brain Rotation Injury Criteria

Introduction

Concussion is a type of traumatic brain injury which results in stretching and damage in brain cells, tissue, and structures. Concussion can result from various events in everyday life including falls, sports collisions and motor vehicle accidents. An impact to the head could disrupt neuronal cell membranes, leading to a neuro-metabolic cascade of molecular changes in the brain, which increases vulnerability to a repeat in-jury [1]. A high incidence of concussions has been reported in sports: there are approximately 3.8 million sports-related concussions per year in the USA [2] and it has been estimated that about 19% of participants in contact sports, such as American football, suffer at least one concussion during a competitive season [3]. However, its recognition is still difficult and concussive events often go untreated. In recent years, the correlation between repeated concussive events and the development of diseases such as dementia, depression, Alzheimer's and chronic traumatic encephalopathy (CTE) is becoming increasingly evident, after studies conducted on the brains of retired players of the National Football League (NFL) [4,5]. Hence, this underlines the importance of improving concussion recognition and developing effective strategies against repeated injury in sports.

In contact sports, the majority of concussive events occur after player-to-player collisions, in particular head-to-head impacts resulting in the highest rate of concussions. In American football head-to-head contact takes place when the helmets of two players contact each other. Intentionally causing such types of collisions is banned in most leagues, due to the elevated risk of injury it presents.

In the literature two different kinds of parameters are analyzed in order to evaluate concussion:

a) Global kinematic parameters: maximum linear and rotational acceleration and their duration, like HIC or BRiC, which consider a global motion, but cannot pro-vide any deterministic information about brain distortion.

b) Intracerebral parameters: maximum axonal strain, maximum strain energy, maximum von Mises stress, maximum von Mises strain, maximum shear stress, maxi-mum shear strain, maximum principal stress, maximum principal strain, mini-mum and maximum pressure [6-9].

However, these parameters can only approximate a mechanical threshold for concussion (often not even unambiguous), estimating a tolerance level for a 50% risk of injury [6-9] when used in finite element human computer models (FE HBM). Furthermore, using strains is not useful to capture the injury severity range, as well as trauma location, which has been proven to be too diffused to be of any relevance [10]. There is no link between strain and injury severity, as strain on its own does not include strain rate effects. The Peak Virtual Power (PVP) method, on the other hand, can overcome these prediction shortcomings. The PVP is derived from the rate-dependent form of the 2nd law of thermodynamics using the Clausius-Duhem inequality, assuming that the injury is represented by the irreversible work in a human body [10]. PVP coincide with peak entropy production and entropy is irreversible. This method is scientifically rigorous and based on theoretical physics (Equation 1) and has been proven to compute accurately the severity of brain white and grey matter injury, and more particularly the location of the brain damage [10,11]. PVP (which is power per unit volume) relates to specific power which also been described by Lemaitre and Chaboche [11].

Fundamentally, PVP can be formulated as:

$$VP = \int_V \sigma \cdot (\dot{D}) \cdot dv \qquad (1)$$

Where $\dot{D}$ is the rate of deformation tensor.

For an elastic-plastic material, PVP is formulated as:

$$PVP \alpha \max(\sigma \cdot \dot{\varepsilon}_p) \alpha AIS \qquad (2)$$

Where ($\dot{\varepsilon}_p$) is the rate of plastic strain.

However, the THUMS human brain computer model, used in this study, is based on an incompressible Kelvin–Maxwell viscoelastic material behaviour [10], consequently there is no plastic region. Therefore, the authors propose a dimensionally equivalent relationship from Equation (2), which considers the total strain (equation 3).

$$PVP \alpha \max(\sigma \cdot \dot{\varepsilon}_{total}) \alpha AIS \qquad (3)$$

This proposal is comparable with eminent research on brain injuries [12], which relate brain injuries with the product of strain and strain rate.

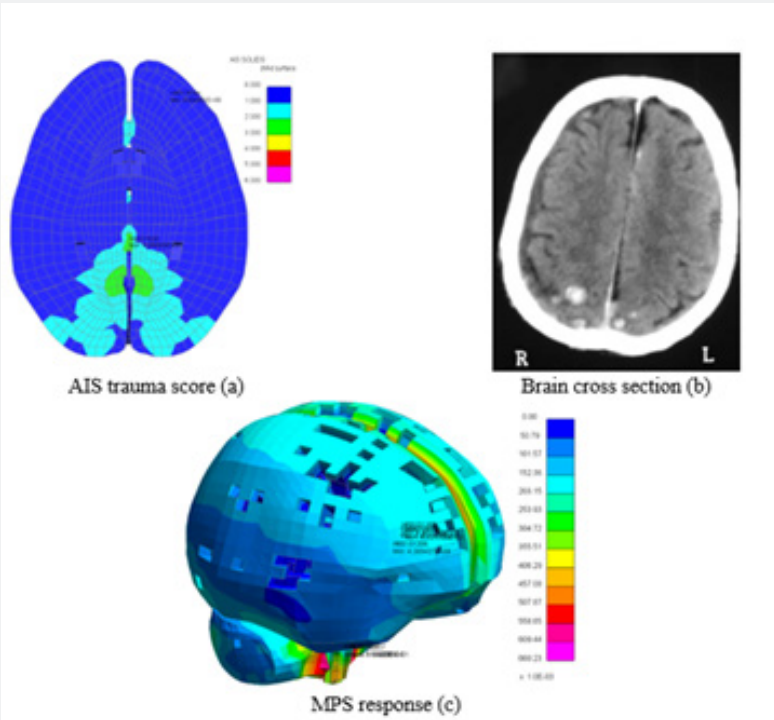

Figure 1: Grey Matter trauma result in a pedestrian collision: PVP (a), brain cross section (b), MPS (c).

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As an illustration, in the case of an occipital fall, the PVP method was utilized to compute the trauma severity level, as well as its location, and compared it to standard CAE output, like maximum principal strain (MPS) [10]. As PVP and AIS are proportional, the PVP values per element are converted into AIS levels, as per (Figure 1a), showing an AIS 3, which relates to some brain damage on the CT-Scan in (Figure 1(b)).

In the case studied [10], in (Figure 1(a)), the PVP method was suggesting a trauma location comparable to the post-mortem (Figure 1(b)), while the MPS trauma plot was not able to suggest any relevant trauma location. Indeed, the THUMS user manual stipulates an AIS for MPS values greater than 25%, hence in the case of (Figure 1(c)), all the brain would have an AIS greater than 4 which is not logical for a simple occipital fall, confirming that the MPS trauma method is inadequate.

In FE HBM models, PVP is computed by extracting the maximum value of the product between Von Mises Stress by Von Mises strain rate, for each element. The Von Mises strain rate was computed by differentiating the Von Mises strain against time. Research [13] impacted rat brains and recorded the oedema in 5 segments, as il-lustrated in (Figure 2), measuring strain and strain rate in each segment (Figure 2). The research highlighted that the maximum oedema occurred at segment 3 (ipsilateral).

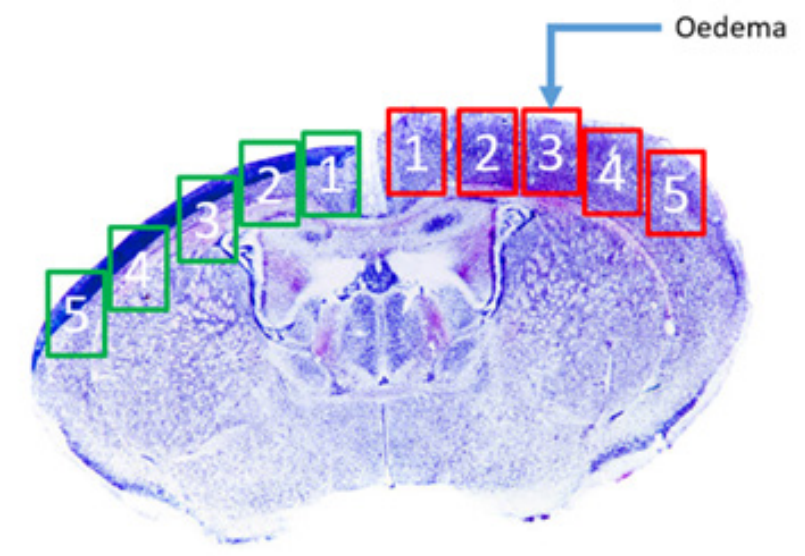

Figure 2: Rat brain segments location illustration, inspired from [13].

In the case studied [10], in (Figure 1(a)), the PVP method was suggesting a trauma location comparable to the post-mortem (Figure 1(b)), while the MPS trauma plot was not able to suggest any relevant trauma location. Indeed, the THUMS user manual stipulates an AIS for MPS values greater than 25%, hence in the case of (Figure 1(c)), all the brain would have an AIS greater than 4 which is not logical for a simple occipital fall, confirming that the MPS trauma method is inadequate. In FE HBM models, PVP is computed by extracting the maximum value of the product between Von Mises Stress by Von Mises strain rate, for each element. The Von Mises strain rate was computed by differentiating the Von Mises strain against time. Research [13] impacted rat brains and recorded the oedema in 5 segments, as il-lustrated in (Figure 2), measuring strain and strain rate in each segment (Figure 2). The research highlighted that the maximum oedema occurred at segment 3 (ipsilateral).

It is impossible to determine which part of the brain contusion has occurred (damage) when strain and strain rate are considered independently. Using PVP, which is the product of strain AND strain rate, it is possible to compute which section sustains the most injury. In the case of (Figure 3), segment 3 has the highest PVP, hence the highest injury location (circled in Figure 3), which had been observed. Therefore, PVP will be used in this study to assess brain deformation that results in concussion. In this paper, we will propose a PVP tolerance, in the mid-brain, for each impact direction that causes mild concussion.

Methodology

This paper calculates the PVP in the brain centre in real concussive head-to-head impacts in American Football. To measure and analyse the effects of the impact on the brain, the THUMS FE model Version 4.02 AM50 was chosen, because of the detailed modelling of internal organs compared to previous versions [14]. THUMS is validated against 38 tests with post-mortem human subjects (PMHS) loading tests both to body components and whole body. For the purpose of this paper, only the head with its

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associated organs are extracted from the complete THUMS model. The head model does not include blood vessels and fluid-structure interaction details, having discretized the brain with a 5mm average element size. All the elements are Lagrangian and the contact is a global sliding contact. The methodology will be based on the following steps illustrated in (Figure 4) and will consider NFL head collisions, using a Riddell VSR-4 helmet. Such protective device is worn by professionals and includes a polycarbonate shell, an inflatable liner with 2 inflation points, a steel polyvinyl-coated quarter-back/running back style facemask and a 4-point chinstrap. Note that the Hybrid III, THUMS4.02 and the Riddell VSR-4 helmet computer models have already been calibrated against physical tests.

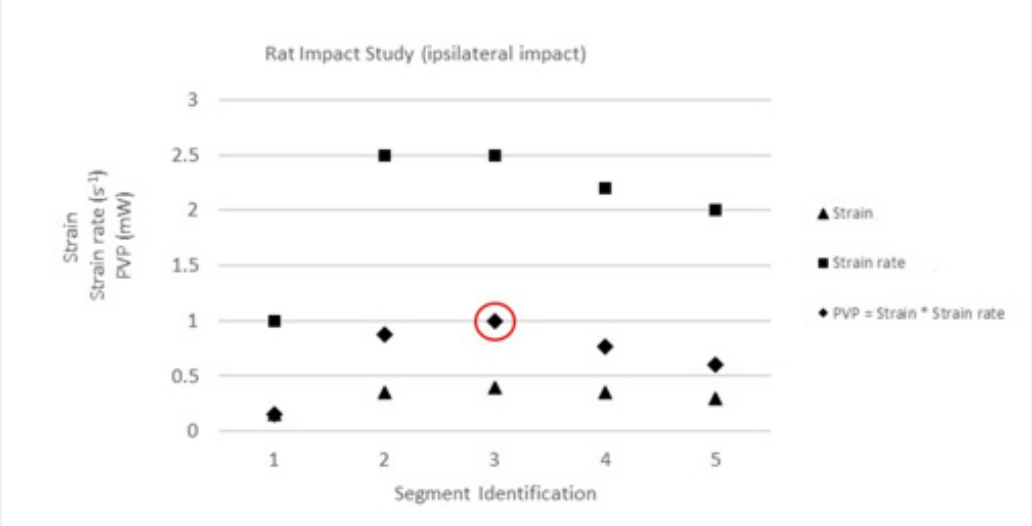

Figure 3: Rat Brain impact study results [13]. Maximum PVP (circled) in segment 3.

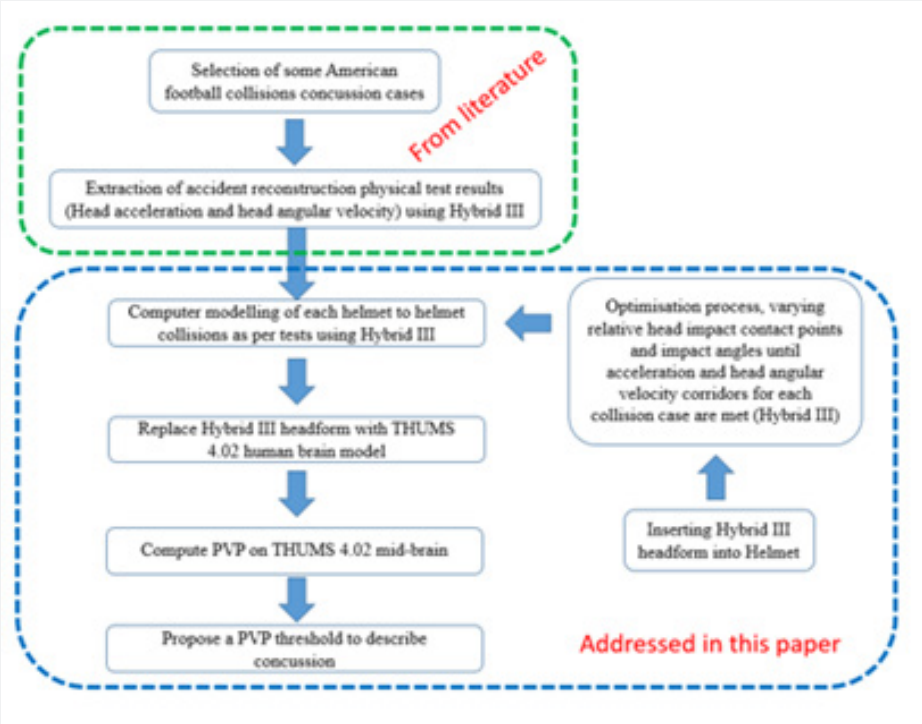

Figure 4: Methodology flowchart.

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Selection of American football collisions concussion cases

The data from the laboratory reconstruction of three incidents of helmet-to-helmet impact resulting in mild concussion during National Football League games was used to replicate real concussive impacts in virtual simulations [15-18]. The concussions have been rated as "mild" as the players carried on playing the game. The experimental tests were carried out using helmeted 50th percentile adult male Hybrid III rigid dummies (head-neck complex), simulating the struck and the striking players, with the same impact velocity, direction, and head kinematics as in the game. The kinematic parameters were determined by a video analysis of the impact. The Hybrid III heads were instrumented with nine linear accelerometers to measure both linear and rotational accelerations [15]. The Hybrid III heads were equipped with Riddell VSR-4 helmets [19]. A helmeted head/neck assembly (simulating the struck player) was guided in freefall from a height sufficient to achieve the same impact velocity as that determined from the video analysis of the game impact. Each accident reconstruction test was repeated; a mean of these tests was then extracted. This was followed by the estimation of test spread, which was calculated using a 2-standard deviation about the mean, leading to the creation of head linear and angular acceleration corridors for each test. These corridors are illustrated in the 3.0 Results section.

The test results are listed in (Table 1). The linear and angular acceleration traces, including test variability corridors have also been extracted from literature and used in the 3.0 Results section.

Table 1: Impact location and reconstruction results for the struck (concussed) and the striking players.

Case	Initial Velocity	Player (Condition)	Linear Acceleration Peak (g)	Rotational Acceleration Peak(rad/s^2)	HIC	Mild Concussion
1	9.5	Struck	118.5	9678	554	Yes
		Striking	59.9	5205	127	No
2	9.9	Struck	80.3	5148	185	Yes
		Striking	34.9	2714	53	No
3	10.8	Struck	123.7	9590	370	Yes
		Striking	88.7	6136	202	No

Fitting the Hybrid III head form in the Helmet

The Hybrid III head form is positioned inside the helmet by using a FE pre-processor. An FE contact is then applied between the head skin and the interior of the helmet. The positioning ensured that no mesh interferences were present during this phase and that any interferences between the helmet foam padding and the skull were adjusted. This was achieved by translating the nodes from the padding. A cross-section through the mesh of the computer model, including the helmet and the head, as illustrated in (Figure 5). It can be noted that the head form is not a free motion head form, but it is connected to a standard Hybrid III neck. The accelerometer in the head will be used to extract the linear acceleration, while the loadcell in the neck will be used to ex-tract the angular acceleration. Both heads are not restrained and are free to move.

Figure 5. Fitting the Hybrid III headform to the helmet (Left: helmet; right: cross section Hybrid III head in helmet).

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Computer modelling of each helmet-to-helmet collision as per tests using Hybrid II

In order to recreate the three collisions using computer modelling an optimisation phase was undertaken to find the most plausible impact point and angle for each case. The relative position and impact angle of each head was parametrised using the LS-Dyna *Include_translate and *Include_rotate. The parametric model was fed into the optimizer LS-OPT to define the initial position of the three heads in the three cases that best fit the experimental results in terms of linear and angular acceleration (Figure 6).

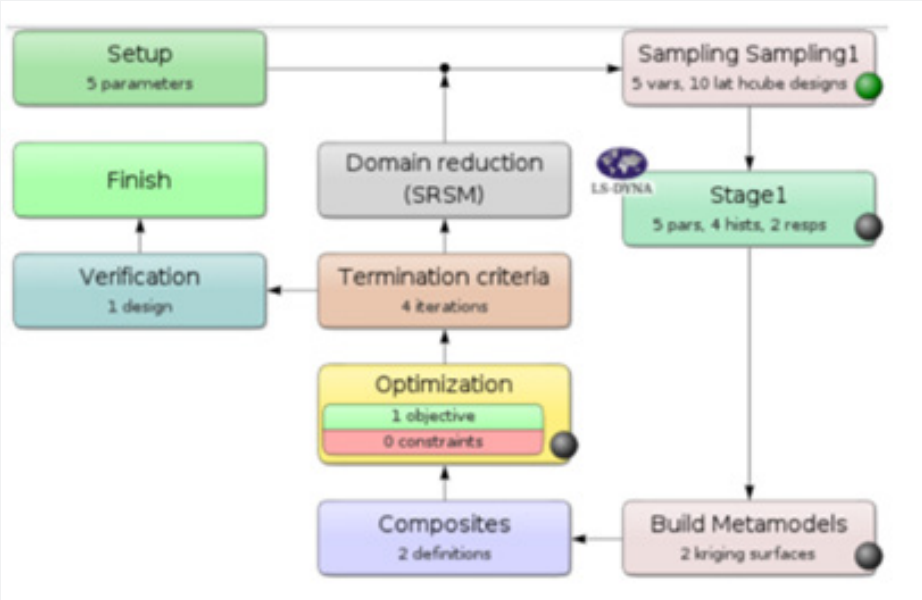

Figure 6: Optimisation flowchart.

In the first instance, before the optimisation process takes place, and for each collision, the two headforms with the helmet were positioned visually relative to each other, as per the physical test setup. The optimisation requires a starting point, and this is how it was defined. There may be a level of subjectivity in this step, however, the process which follow will remove this bias. In a second step, a design of experiment (DOE), using 5 variables, was setup, considering what could be the Y position, Z position, X rotation, Y rotation and Z rotation of the striking head (Figure 7). The aim of this DOE is to capture the possible head acceleration responses across the collision de-sign domain. In order to scan through this domain, a Latin Hypercube (LH) sampling method is chosen to create the sample points for the five design variables discussed. The LH method is a stochastic DOE algorithm, which generates random numbers con-forming to a uniform distribution. It is particularly suited for optimization with genetic algorithms and Response Surface Method (RSM) training. To find a balance between accuracy and computational resources usage, the number of sample points has been chosen to be n = 50. For each impact, the respective linear head and angular accelerations are recorded for both heads. Following the DOE, two kriging response surfaces (1 for each output) were built ("Build Metamodel in Figure 6). The first kriging response surface represented the linear acceleration, while the second one to the angular acceleration, and this for each collision. Finally, an optimisation objective was created aiming to achieve for peak linear and angular acceleration values of (Table 1). The kriging response surfaces were then used, with an optimisation algorithm to extract the best combination of the 5 variables to reach these maximum values for each impact. In order to accept the results from the optimisation, the optimized responses are overlaid against the test corridors. To confirm the success of the optimisation, the signal has to lay within the test corridors.

Following the optimisation phase, the initial impact positions for the three cases have been found and illustrated in (Figure 7).

Replacement of Hybrid III head model by THUMS 4.02 head for concussion assessment

The same process as 2.2 was applied with the THUMS 4.02 head model, as shown in (Figure 8). As the Hybrid III headform and the THUMS 4.02 outer head geometry slightly differ, a verification step was undertaken to prevent, again, any initial penetrations and cross-edge, which may cause artificial contact forces in the

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head and skew the computer trauma computation outputs. Few nodal penetrations were present in the chin-strap area. These nodes were manually moved by few millimeters away from the THUMS headform. The same orientation and impact velocities were applied to the helmeted THUMS head model.

Figure 7: Actual head positions optimised for each collision case.

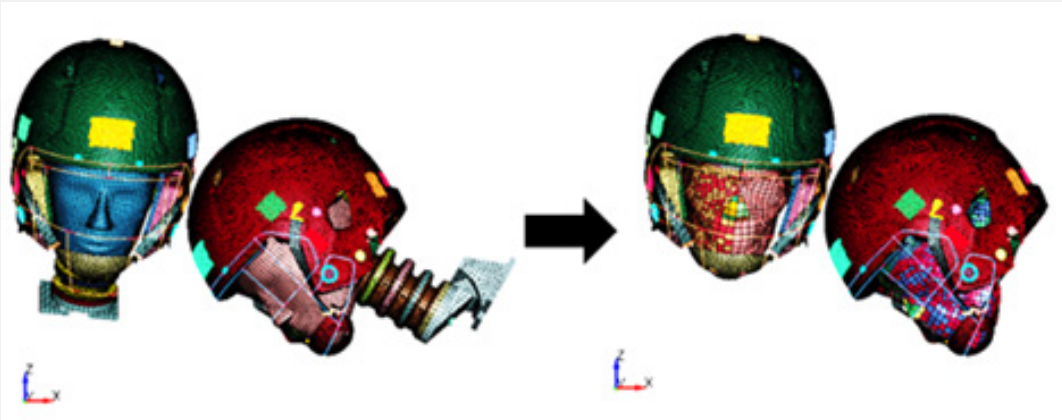

Figure 8: Replacement of the Hybrid III FE model with the THUMS FE model. Case 1: Hybrid III FE model on the left and THUMS FE model on the right.

Defining the zone in brain for concussion

The mid-brain is responsible for the motor movement, particularly movements of the eyes and to the auditory and the visual processing. The corpus callosum is, on the other hand is responsible for connecting the axon fibres between the two central hemispheres. Previous research has proven that both of these areas are important zones for concussion [20-22]. As such, a new part division of the brain was defined, starting from the white matter of the THUMS FE model, enclosing both of these regions of the brain (Figure 9). This division was necessary to compute PVP, as per Equation 1, and evaluate how much power is necessary to cause concussion for this specific area of the brain.

This was performed by creating a new *Part in the THUMS 4.02 computer model and move the elements from the white matter, located in the mid-brain area, into this new part. The same material properties as well as element formulation were transferred, as no literature was found stipulating that that the mid-brain material properties were different from white matter.

Results

Results of Hybrid III correlation

The linear and angular acceleration correlation results (plotted with a standard CFC1000 cut-off filter) are provided in (Figure 10) and (Figure 11) for case 1, (Figure 12) and (Figure 13) for case 2, (Figure 14) and (Figure 15) for case 3. These results are laid over the test corridors defined in Section 2.1.

All cases mostly meet the test corridors, highlighting that the impact boundary conditions are a plausible representation of the collisions which are being reconstructed.

PVP Results using THUMS 4.02

Following the three previous successful applications which showed strong correlation, the Hybrid III headform was replaced

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by a THUMS 4.02 human head, as per the methodology. The models were then recomputed and the PVP values for the corpus callosum and the mid-brain for the struck and striking players extracted. The results are presented in (Table 2).

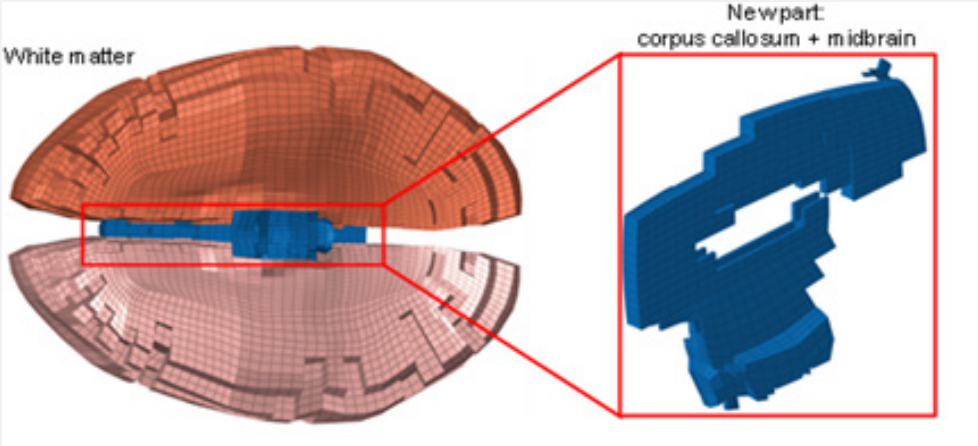

Figure 9: New division extracted from the White Matter of the THUMS FE model.

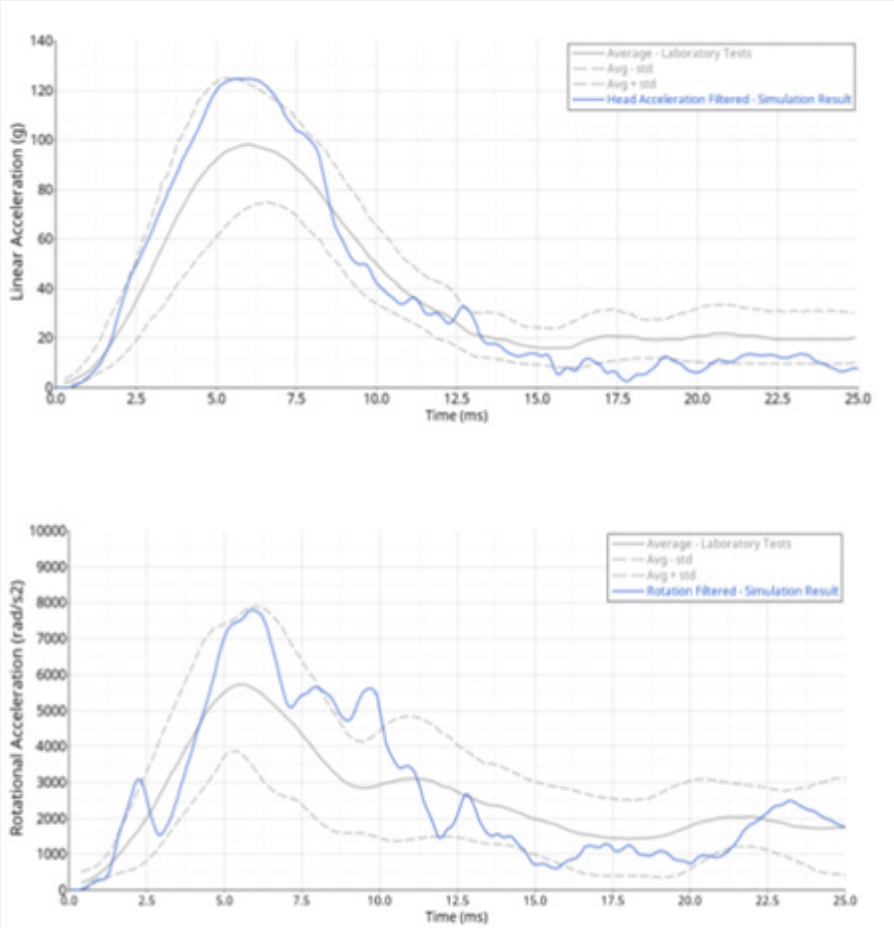

Figure 10: Case 1-Struck player Hybrid III model response vs tests.

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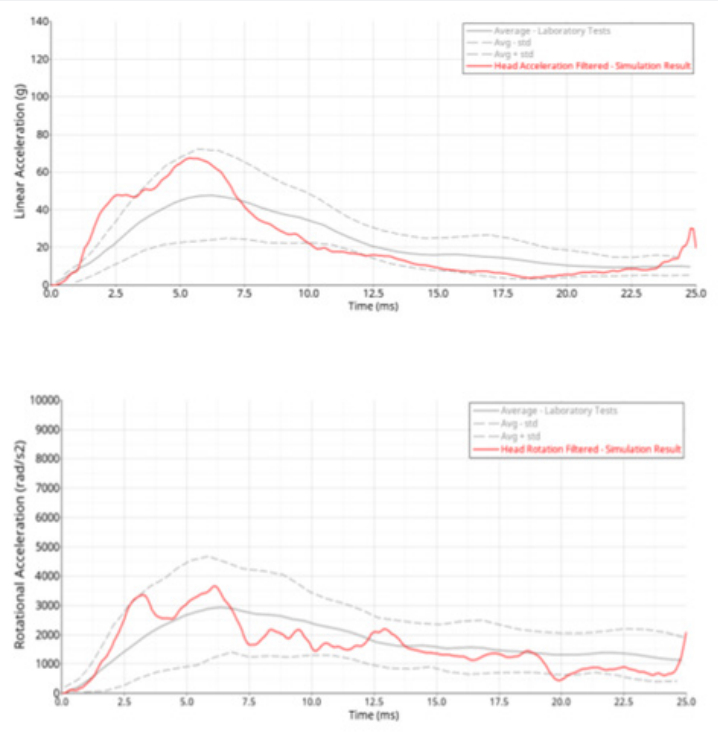

Figure 11: Case 1-Striking player Hybrid III model response vs tests.

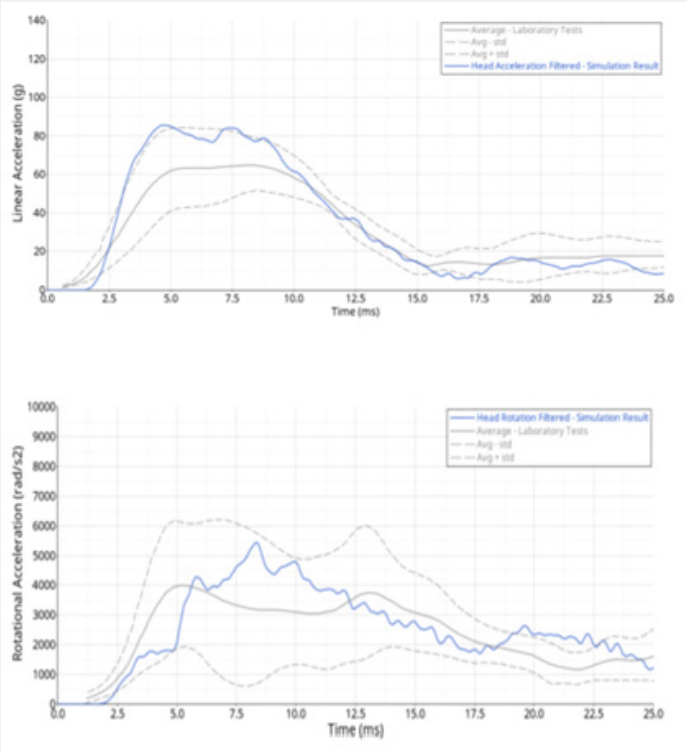

Figure 12: Case 2-Struck player Hybrid III model response vs tests.

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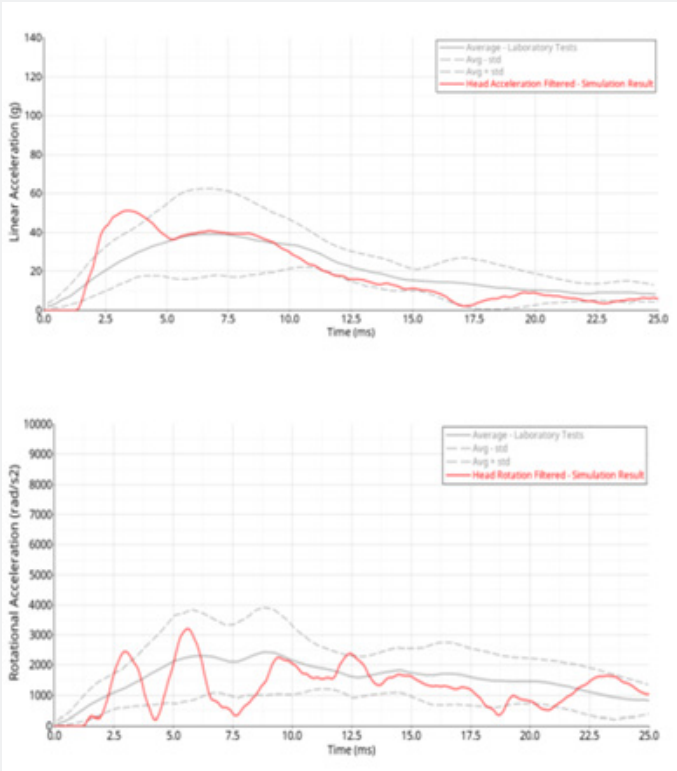

Figure 13: Case 2-Striking player Hybrid III model response vs tests.

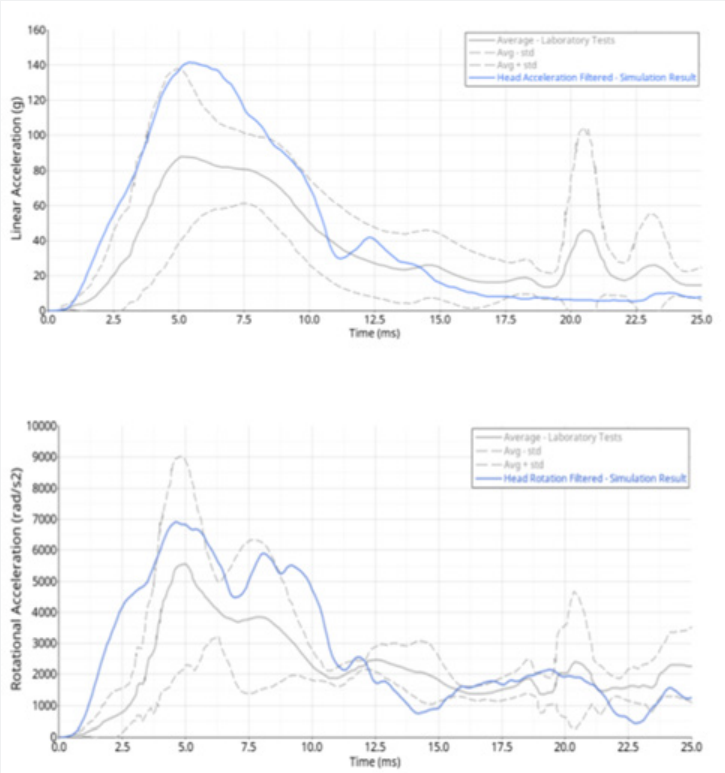

Figure 14: Case 3-Struck player Hybrid III model response vs tests.

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Table 2: PVP results from the three collisions studied.

			Impact Location	Impact Velocity (m/s)	PVP (mW)	Concussed (Y/N)
Striking	WM_Centre	Case 1	Lateral LHS	9.5	0.928	Y
Struck	WM_Centre	Case 1	Head Top	0.0	0.951	N
Striking	WM_Centre	Case 2	Head frontal	9.9	9.405	Y
Struck	WM_Centre	Case 2	Head Top	0.0	0.243	N
Striking	WM_Centre	Case 3	Lateral LHS	10.8	2.923	Y
Struck	WM_Centre	Case 3	Head Top	0.0	1.184	N

Figure 15: Case 3-Striking player Hybrid III model response vs tests.

Discussion

Concussion is a type or traumatic brain injury. Trauma in general can occur at the cell, tissue or organ level and in concussions it involves the stretching and damaging of brain elements and structures. Computational models, which are the end-product of applying tissue property behaviour in silico, can produce numerical solutions and translate the trauma scenarios to real world situations. As shown in the present study the three collisions' angular and linear accelerations measured in the FE impact simulation with the Hybrid III headform, mostly meet the upper and lower corridor test responses. Consequently, the initial boundary conditions (impact velocities, impact angles, and locations) defined with the optimization process can be considered realistic and offer a solid foundation for studying brain distortion by replacing the Hybrid III headform with a THUMS 4.02 human model.

The optimisation process was overall successful, as for each collision, however, in some cases, it can be noticed that the signals are slightly off and are sometimes more aligned with the upper bounds. This is a consequence of the optimisation process undertaken by using only the peak accelerations as the sole objective function. As such, there was no control on the shape of the acceleration responses, as well as its shape relative to the test spread. The facts that the acceleration signals are mostly within the corridors is due to the fact that the Hybrid III headform and the Riddell SV4 CAE model have been very well validated.

The optimisation process has led to plausible boundary conditions, however maybe a better control of the optimisation constraints could have led to a closer response within the corridor boundaries. A way to achieve this, for each collision, would be to compute a CORA rating [23] above 0.86 ("excellent" rating), to assess how well the responses relate to the mean and standard

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deviation corridors. This said, the optimisation outcomes are plausible, as they confirm that the linear acceleration, i.e. the impact force intensity is of the correct magnitude compared to the tests, and that the angular acceleration, representing the lever arm of that impact force, i.e. the distance between the transducer and the application of the force are also in the correct magnitude to the test. Consequently, the accidents reconstructed in this paper have some merit and can be used to study concussion.

It can be noted, that, in this study, concussion predictions based on HIC are not reliable as for case 3, a HIC of 202 does not lead to concussion, while in case 2, a HIC of 185 does. HIC values under 200 leads to AIS values in the region of '1', which is minor injury, however concussion and injury scale are not the same, hence it is felt that acceleration-based criteria cannot predict localized mid-brain distortions. BRiC has not been investigated, as no target test result values were provided. The test data used in this study are based on research in 2005 in the US. Consequently, the 'severity' of concussion is not known because the Glasgow Coma Scale (GCS) was not used there. As the players in the examined cases came back into the game, it can be reasonably assumed that any concussion recorded were mild, hence with a GCS between 15 and 13 (Table 3) [24].

Table 3: Glasgow Coma Scale [24].

GCS score	Ranking
15-13	Mild
12-9	Moderate
8-3	Severe

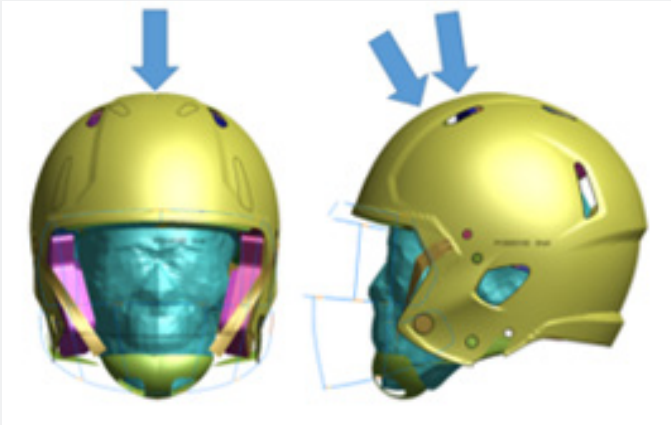

Figure 16: Proposed areas of helmet where concussion is avoided when PVP values <1.184mW.

As the three struck players were impacted on the top of the helmet and were not concussed (Figure 16), it is possible to propose a maximum allowable PVP value of 1.184mW to be measured in the white matter centre of the brain. This value can be used to design future safety protection gear or assess contact sports safety when subjected to vertical impacts. More data would be required to verify whether this value can be increased. For the frontal and side impact directions, the information provided suggest mild concussion. Consequently, in order to avoid moderate and severe concussions, the critical threshold PVP for side impact, measured in the mid-brain, must be less than 0.928mW and for frontal impact 9.405mW. More information is necessary to extract the PVP values to avoid any concussions in these two directions.

The dataset used in this study, retrieved from literature, was limited, hence future work would require international collaborations to include more cases, including a wide range of concussion levels, to define the PVP threshold for concussion in all head impact direction. More information would also be needed to prove where in the brain concussion would occur in each collision, even if most research is suggesting the corpus callosum and midbrain locations to be the areas of interest. In this paper, the authors have proposed a new approach to calculating PVP for soft tissues, by proposing a dimensionally equivalent method taking the full strain tensor instead of the plastic tensor. This was necessary because of the lack of plastic response from the brain, due its viscoelastic material properties used. It is proposed, in further work, to quantify the increase of PVP value due to this assumption.

In the three accident reconstruction cases, the primary impact is suggested to be translational and not rotational. Maybe a more detailed video analysis of future events could be considered to ensure that rotational velocities are included, so that their relevance can be assessed. When these rotations are measured, a new test regime will need to be conducted to provide updated corridors for linear and angular accelerations, but also measuring the other acceleration components of the Brain Rotation Injury Criteria (BRiC), so that the CAE boundary conditions are captured in all directions. It is therefore proposed to use the same methodology for this paper and extract PVP in the mid-brain and link it to the GCS observed in the collision. As discussed in

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the paper, the number of cases is small and needs to be greatly increased in order to capture cut-off PVP values causing severe, mind and minor concussions. However, these few samples have allowed the authors to develop a rigorous method to computer the concussion phenomenon.

Conclusion

The present study provided evidence that computational models utilizing Peak Virtual Power (PVP) are able to assess the concussion level, with the concussion level varying depending on the impact direction. This was achieved by using previous re-search [12] and applying the PVP method which extracted plausible maximum brain distortion segment locations, validated using imaging.

To assess in depth the PVP based computational method the present study recon-structed 3 NFL helmet-to-helmet impacts by means of finite element (FE) and using a validated Biocore helmets fitted with calibrated Hybrid III headforms. An optimisation routine extracted for each scenarios the helmeted head impact angles to correlate against linear and angular acceleration impact corridors defined as realistic impact conditions. Once the exact impact conditions are known, the Hybrid III headform was replaced by a validated THUMS 4.02 human head model in which the PVP was computed for each head, at the centre of the brain. The research has concluded that in the vertical impact direction, no concussion would occur for a PVP value less than 1.184mW and that moderate and severe concussions could be avoided for side impacts and frontal impacts with PVP values lower than 0.928mW and 9.405mW respectively. With this innovative approach, it is possible to design safer head gear, as well as assess the safety of contact sports.

Future Work

In this paper, it has been evidenced that PVP has the potential to investigate concussion. As future work it is proposed to investigate whether a relationship between PVP and the GCS can be established, in the same manner that the relationship between PVP and AIS is a cubic function for brain injuries. It is proposed to investigate whether a regression could be used to estimate values of this relationship. The format is expected to be of the shape of Equation 2.

$$PVP = PVP_T + K(15 - GCS)^n \qquad (4)$$

Where:

a) PVPT is the PVP threshold for a GCS value of 3, i.e. the lowest value for concussion

b) K is a constant of proportionality

c) 15 is the value of the GCS for no concussion.

d) n is an exponent

This relationship can be explained. The lower the PVP value, the lower the level of concussion, and vice versa. As per (Table 3), a dangerous level of concussion is rated as a low number, this number being '3' and will be attributed a high PVP value. A minor level of concussion is rated as a high GCS number, the maximum being 15 and will be attributed a low PVP value. The relationship will therefore be decreasing and the se-verity of the drop of PVP, related to the GCS, can be adjusted with exponent 'n'. This relationship will be impacting direction dependent, as during the impact, the stress wave travelling through the brain will depend on the skull stiffness and shape which differs in frontal, occipital and lateral.

The dataset used in this study, found in literature, was limited, hence future work would require international collaborations to include more cases including a wide range of concussion levels to test the validity of Equation 2. It is proposed to improve the accident reconstruction phase using a CORA rating for striking and struck head-forms, so that the computer responses consider the curvature of the mean response as well as the spread of tests captured by the corridors.

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