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Modelling and Analysis of Cerebrospinal Fluid Flow in The Human Brain – Is Cerebrospinal Fluid an Effective Protective Mechanism During High-Impact Loading?

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Abstract

Aim: This study investigates cerebrospinal fluid (CSF) flow dynamics to enhance the understanding of brain biomechanics and the importance of CSF during high-impact loading.
Methods: Comparative analyses were conducted using the benchmark model with smoothed particle hydrodynamics (SPH), without cerebrospinal fluid, and with an additional element — the arachnoid trabeculae — which functions as rigid connections between the brain and skull. The numerical modelling of cerebrospinal fluid and the derived conclusions were validated and calibrated through experiments performed in the additional research phase.
Results: The research emphasises the challenges of accurately modelling cerebrospinal fluid dynamics and brain biomechanics. The results were unexpected in several ways. Initially, a rigid cortex-skull connection was anticipated to yield results nearly identical to those observed in Hardy's experiments. Even more surprising were the results for the models with cerebrospinal fluid modelled as smoothed particle hydrodynamics and the model without cerebrospinal fluid, which showed almost identical results in comparison to each other. The novel physical experiment with a gelatine insert subjected to controlled loading and numerical model simulations revealed that SPH models exhibited closely resembling fluid displacement, while tetrahedral elements imposed unrealistic rigidity.
Conclusions: The simulations and the novel experiment provide key insights into cerebrospinal fluid dynamics during traumatic brain injury. The findings suggest that the protective function of CSF might be less pronounced under extreme conditions than previously assumed. The smoothed particle hydrodynamics method demonstrates clear advantages over tetrahedral finite element approaches by offering superior brain-in-skull flexibility and avoiding the excessive rigidity inherent to traditional finite element models. We concluded that mechanism of brain protection by CSF is performed rather by hydraulic damping than the brain immersion in vast volume of CSF.
Keywords: cerebrospinal fluid, head model, brain, finite element, numerical model, particle hydrodynamics

1 Introduction

Traumatic brain injury (TBI) is defined as a disruption in the normal function of the brain induced by external mechanical forces. Depending on the severity, such injuries may cause

physiological or structural damage to the brain. This damage can manifest as loss of consciousness, post-traumatic amnesia, and transient or permanent neurological deficits. TBI is a major health concern worldwide, contributing significantly to long-term disability and mortality rates. According to the 2016 Global Burden of Disease Study, approximately 27 million new cases of TBI are reported annually [15]. The consequences of TBI extend beyond physical impairments, often resulting in cognitive deficits, emotional disturbances, behavioural changes, and varying degrees of lifelong impacts [3,21].

Cerebrospinal fluid (CSF) fills the ventricular system and sulci of the brain and surrounds it as a narrow layer between the meninges (blue arrows) – Figure 1. CSF is produced mainly in the choroid plexus of the lateral ventricles (yellow arrow), flows through two foramina of Monro (green arrow) to the third ventricle, through the cerebral aqueduct to the fourth ventricle and leaves the ventricular system through the apertures of Luschka (red arrows) and foramen of Magendie (dash orange arrow). In adults, the total volume of cerebrospinal fluid is about 150 ml [7], yet the amount of fluid that can be accounted for mechanical protection in a healthy adult head is no more than 50 ml.

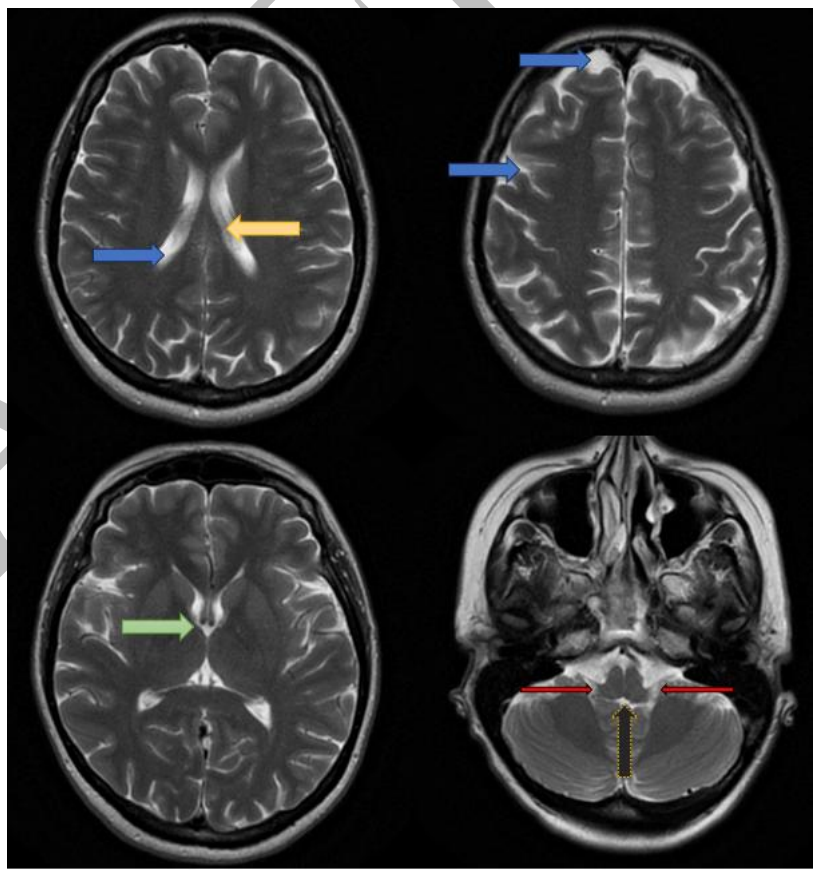


Figure 1. Magnetic Resonance Imaging (MRI) scan (T2) illustrating CSF (white) in the brain. CSF is produced in the choroid plexus of the lateral ventricles (yellow arrow), flows through the foramina of Monro (green

arrow), third ventricle, cerebral aqueduct, and exits via the apertures of Luschka (red arrows) and foramen of Magendie (dash orange arrow), surrounding the brain between the meninges (blue arrows).

Moderate and severe TBI mechanisms are classified into localised injuries, such as contusions and haemorrhages caused by direct impacts, and diffuse injuries, such as diffuse axonal injury (DAI), caused by rapid acceleration or deceleration forces [16,37]. As a protective medium for the brain, cerebrospinal fluid (CSF) plays a role in reducing mechanical forces during traumatic impacts [32]. Unfortunately, the protective effects of CSF are insufficient to prevent or mitigate significant damage under severe impacts [24]. CSF exhibits Newtonian fluid behaviour and has a composition similar to water, with a density and kinematic viscosity comparable to plasma at 37°C [5]. CSF is produced primarily by the choroid plexus of the ventricular system at a rate of approximately 350 μL/min [5]. The production of CSF is a complex process involving passive filtration and active ion transport within the choroid plexus (Figure 2) [5,9,12,36]. From the ventricular system, CSF flows outside the brain through the apertures in the fourth ventricle, surrounds the brain and is absorbed by arachnoid granulations of the meninges of the brain. In addition to its mechanical protective function, this continuous flow of CSF ensures the removal of metabolic waste products and maintains intracranial pressure within a narrow physiological range [8,36].

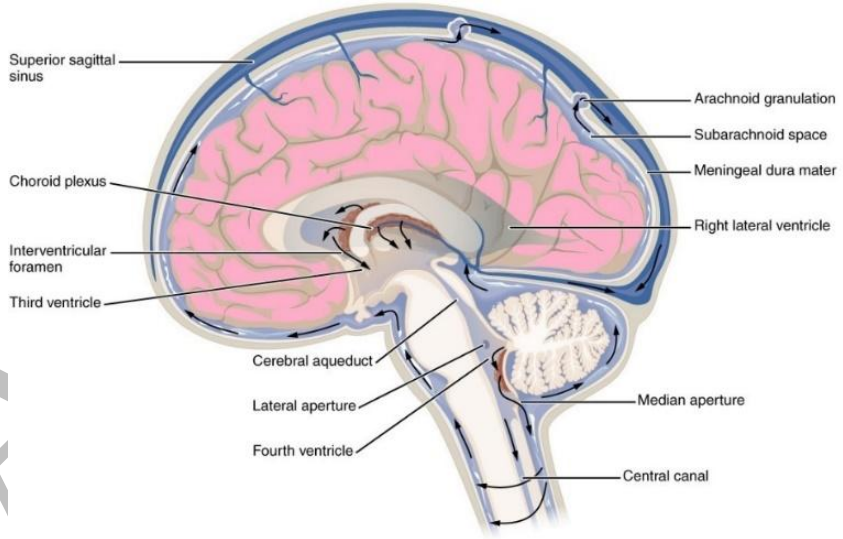


Figure 2. Cerebrospinal fluid circulation, adapted from [12]

Accurately simulating CSF behaviour is essential for understanding the mechanisms of TBI and developing effective protective measures [8,27]. Biomechanical models, particularly finite element head models (FEHMs), are critical tools for simulating the mechanical behaviour of the human head under various impact conditions [2,4,31]. FEHMs are developed using detailed imaging techniques such as computed tomography (CT) and magnetic resonance imaging

(MRI) to create accurate geometric representations of the skull, brain, and cerebrospinal fluid [4,11,23]. The process of segmentation, followed by mesh generation, ensures that each anatomical structure is accurately represented, allowing for precise simulations of head biomechanics during trauma [6,11].

However, for these simulations to be effective, the mechanical behaviour of brain tissues, meninges, and CSF must be accurately modelled. This leads to developing sophisticated brain models that capture complex interactions between different structures during traumatic events [33]. Human brain models, such as the Wayne State University Head Injury Model (WSUHIM) [35] and the Global Human Body Models Consortium (GHBMC) M50 finite element (FE) model [20], incorporate detailed representations of brain tissues, the skull, and the CSF, designed to simulate various injury mechanisms, including skull fractures, brain deformation, and the dynamic response of CSF during impacts. To achieve more accurate simulations, it is crucial to incorporate realistic representations of CSF flow dynamics and its interaction with surrounding brain tissues. Traditional finite element methods (FEM) often simplify CSF as a solid or quasi-solid layer, which can lead to inaccuracies in simulations of fluid-structure interactions [1,17,18,28,39,40]. Smoothed Particle Hydrodynamics (SPH) offers a more realistic approach by treating CSF as fluid to address this limitation. Ptak et al. studies demonstrated that SPH-based models improve the accuracy of simulations by capturing the rapid shifts in CSF during sudden accelerations or decelerations [25]. By incorporating these advanced techniques into brain models, researchers can better understand how CSF dynamics influence brain injuries and develop more effective protective measures. This adaptation represents a significant step forward in creating comprehensive and reliable numerical human body models [13,19,30,34].

Nevertheless, ongoing refinement is necessary to improve the accuracy and predictive performance of the numerical head models. This research aims to develop a more precise simulation of the human head’s response to sudden acceleration by evaluating existing models and introducing specific modifications. Ensuring the reliability of numerical models requires validation against experimental data. To achieve this, an innovative experiment has been designed using a fluid-filled container that replicates the human skull with a gelatine insert which mimics the brain and is subjected to sudden acceleration. High-speed camera recordings captured the fluid's behaviour during dynamic conditions, delivering essential data for verifying and calibrating the numerical model. This validation process is crucial for narrowing the gap between theoretical simulations and practical, real-world scenarios.

2 Materials and methods

The aHEAD (advanced Head models for safety Enhancement And medical Development) benchmark model (Figure 3) represents a significant advancement in the computational modelling of human head biomechanics for safety and medical applications [25]. The development of the aHEAD model followed a structured five-stage process: medical data acquisition and craniometric measurements, segmentation and 3D modelling, finite element (FE) modelling and material testing, modelling of the central nervous system, as well as validating the model using experimental data [25].

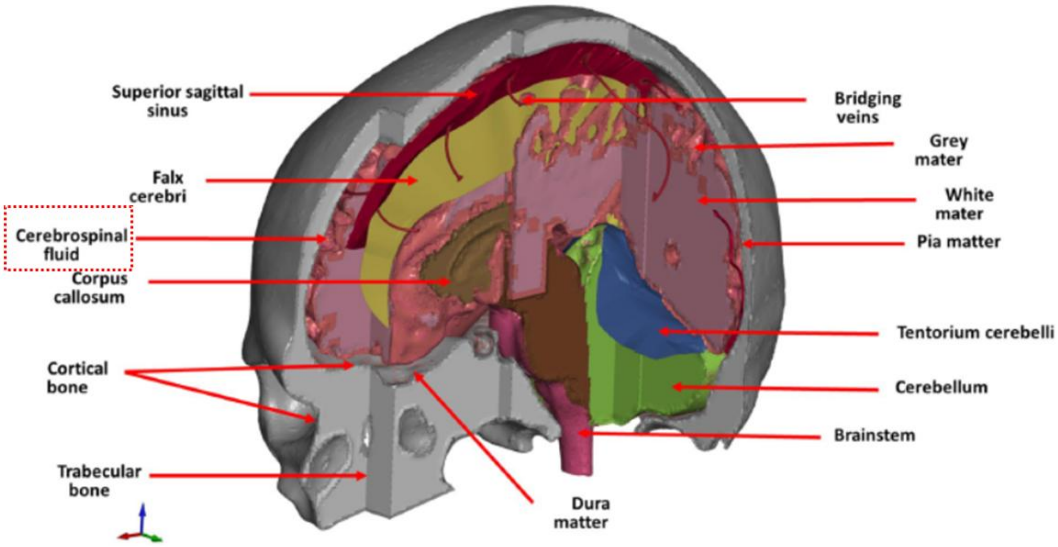


Figure 3. aHEAD benchmark model, developed by authors [25]. The further modified cerebrospinal fluid is indicated in a dashed frame.

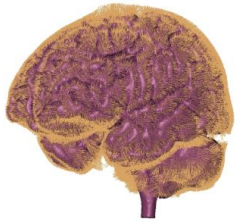
Advanced simulation software, including HyperWorks 2021 and LS-DYNA, facilitated in-depth analysis and iterative model refinement [25]. Unlike conventional models that simplify cerebrospinal fluid (CSF) using hexahedral or tetrahedral elements, the aHEAD model utilised smoothed particle hydrodynamics (SPH) elements for more realistic fluid-structure interactions during impact events. The mechanical properties used in the model were selected to reflect real tissue behaviour and are summarised in [26,38].

2.1 Head model modifications

Two primary approaches were explored to enhance the accuracy of CSF representation – comparing the aHEAD model with and without CSF to evaluate SPH's effectiveness in capturing fluid dynamics, as well as introducing beam elements (Hughes-Liu beam

formulation) to rigidly link the brain to the skull, simulating the mechanical role of arachnoid trabeculae [25] (Table 1).

Table 1. Properties of the analysed models: benchmark aHEAD model (CSF as SPH), no CSF and rigid junction

Model and no. of finite elements	**Approach for the cerebrospinal fluid**	**Type and no. of finite elements/particles for CSF**	**Visualisation**
aHEAD benchmark (CSF as SPH) 3,156,379	CSF as a smoothed particle hydrodynamics	SPH 140,199	
no CSF 3,016,180	Model without CSF	No finite elements are used – there is a gap between the cortex and dura mater	
rigid junction 3,075,239	Model with rigid arachnoid trabeculae connection	Rigid beams 59,059	

To achieve the first approach the new material model of SPH elements was introduced. The Murnaghan equation of state (EOS) (1) was applied for accurate fluid dynamics simulate of the incompressible fluid.

$$p = k_0 \left[\left(\frac{\rho}{\rho_0} \right)^{\gamma} - 1 \right] \tag{1}$$

Where:

p – pressure,

k_0 – modulus of incompressibility (the bulk modulus of the material),

γ – Gruneisen parameter, which characterises the sensitivity of the material volume to changes in pressure,

ρ – density of the fluid.

For precise simulation of fluid dynamics, γ is typically assigned a value of 7, and k_0 is selected accordingly:

$$c_0 = \sqrt{\frac{\gamma k_0}{\rho_0}} \geq 10 v_{max} \tag{2}$$

where:

c_0 – speed of sound in the fluid,

v_{max} – the maximum expected fluid flow velocity.

In the considered model, k_0 was set to 0.15. It should be mentioned that not all material models are required to use EOS. The above considerations apply to this model approach, in which a material charter – null (*MAT_NULL) – was assigned for SPH elements. The null material lacks yield strength and exhibits fluid-like properties. The mass density was set as 1e-9 t/mm^3, the pressure cutoff was set as -1e6 MPa, and the viscosity coefficient was set as 7e-10. This was compared with a version of the model where CSF was removed entirely. The goal was to assess whether the SPH method could accurately simulate the fluid properties of CSF, particularly during traumatic events such as head impacts.

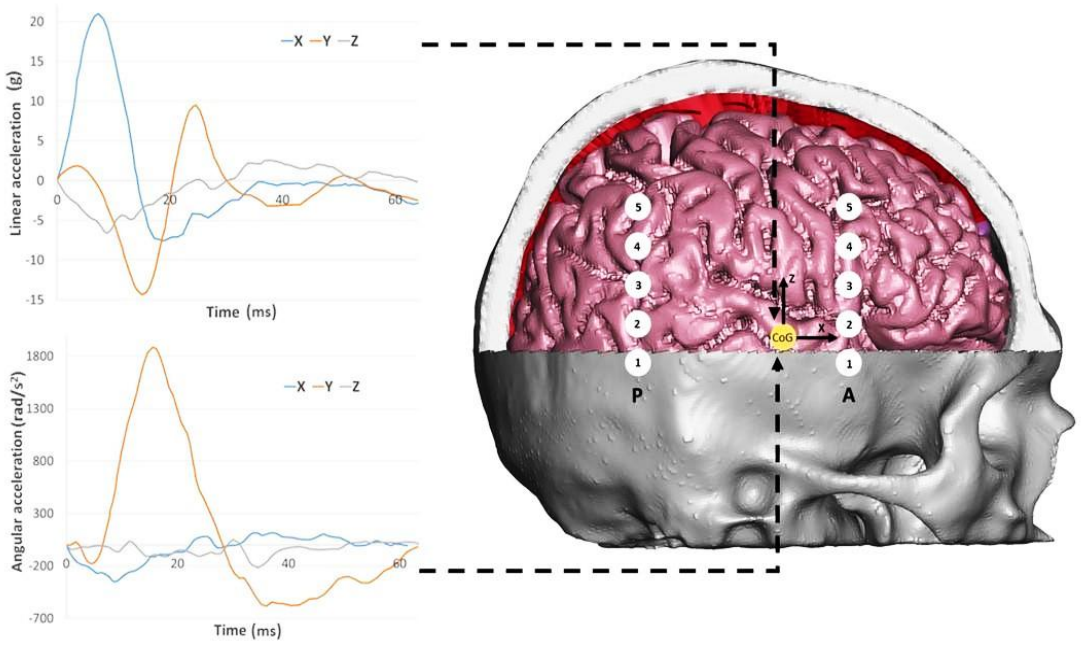
The following major modification involved the incorporation of arachnoid trabeculae, which are structures that constrain the brain's movement relative to the skull, influencing the internal forces that affect brain tissue during a head impact. The arachnoid trabeculae were modelled using beam elements (Hughes-Liu beam formulation) that rigidly connect the brain to the skull. An elastic material model (*MAT_ELASTIC) was chosen for the beam elements, with the mass density set at 1.8e-10 t/mm^3 and Young's modulus of 1.5e5 MPa.

The refined models were subjected to identical boundary conditions derived from Hardy's C755-T2 test [14], ensuring simulation consistency. The acceleration functions served as input parameters for the finite element analysis, targeting the centre of mass in the head model (Figure 4). The coordinates of the surveyed points (Figure 4) in the aHEAD model were accurately mapped onto the brain's white matter. Incorporating these reference points into the computational framework allowed the authors to quantitatively evaluate the precision of model simulations by comparing them with data derived from actual Hardy's experiment. These reference points serve as fixed markers, enabling the evaluation of critical metrics such as displacements.

Boundary conditions for the full-head model simulations

Based on the methodology described by Hardy et al. [14]. the simulation boundary conditions were established. The C755-T2 test, which is the most commonly referenced test in the literature, was employed in this study. Data collection was performed using a 6-axis accelerometer that captured both linear and angular accelerations. The recorded data consisted

of six acceleration components as functions of time (three translational and three rotational), which were subsequently applied as input to the centre of mass in the finite element (FE) head numerical code (Figure 4). It is important to note that in the numerical model, all markers, except for P1 (located in the grey matter), were positioned (i.e. indicated on FE nodes) within the white matter.



Point and position (mm)	CoG	p5	p4	p3	p2	p1	a5	a4	a3	a2	a1
X	0.0	-37.2	-34.3	-33.8	-30.1	-26.3	12.4	12.6	13.0	13.3	13.6
Y	0.0	-30.6	-29.9	-29.5	-28.0	-27.9	-34.4	-33.4	-33.4	-33.7	-31.5
Z	0.0	33.3	26.9	18.0	7.0	-2.8	39.5	28.0	22.5	7.2	-3.7

Figure 4. Boundary conditions for a C755-T2 Hardy test; points a1–a5 refer to the anterior, and the p1–p5 points to posterior [25]

The model compared the relative displacements of ten specific points within the brain, as defined by Hardy's experimental data [25]. These points were used to assess the accuracy of the virtual models in replicating real-world brain motion during traumatic events. By aligning simulation outputs with Hardy's benchmark data, the reliability and predictive accuracy of the aHEAD model were evaluated.

2.2 Experimental and numerical setup for gelatine insert

Experimental setup

The experiment aimed to validate and calibrate the numerical model of cerebrospinal fluid (CSF) using a gelatine-based fluid as a surrogate. The experimental setup (Figure 5) included a container filled with water (1) and the gelatine insert (2). The fluid was enclosed by a membrane end cap (3), preventing its escape. The experiment utilised a sledge mounting structure (4) with aluminium rails (5) to minimise the friction between the container bearings and the rails. The high-speed Phantom v.7312 camera was employed to capture the gelatine insert dynamics and the container kinematics via high-contrast markers (6) at 1,400 frames per second, with a resolution of 1664x1000 pixels and an exposure time of 712 µs. The container's initial velocity was induced by mechanical loading (7).

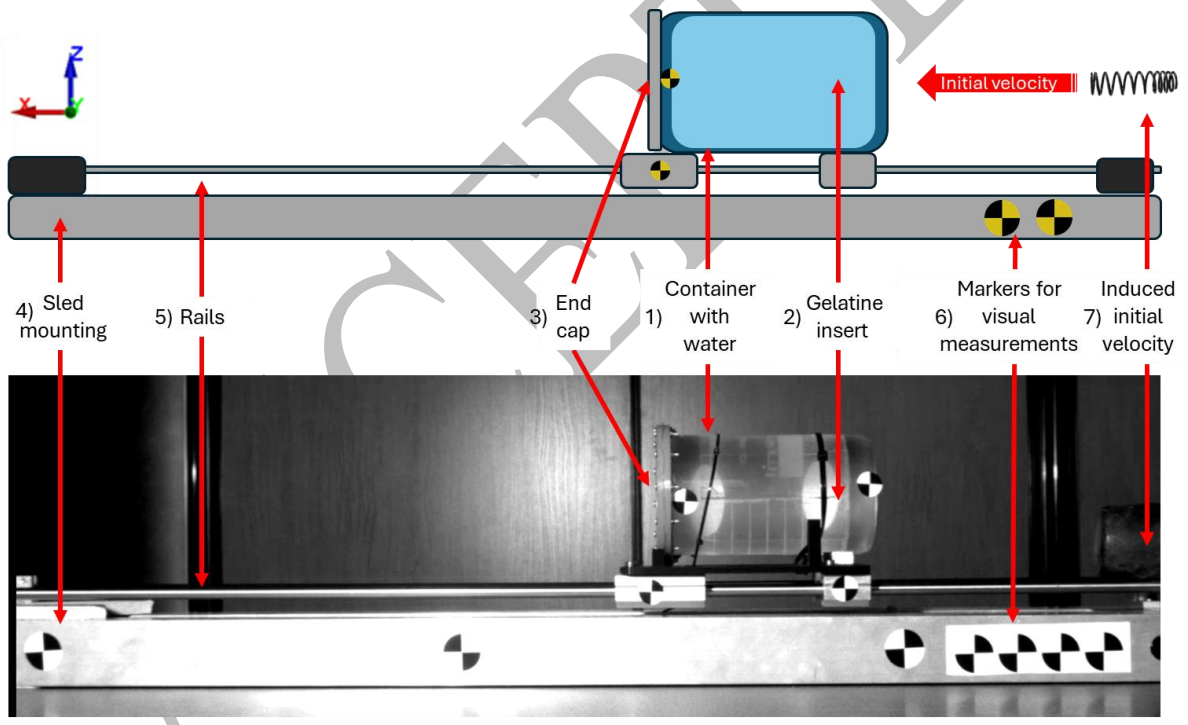


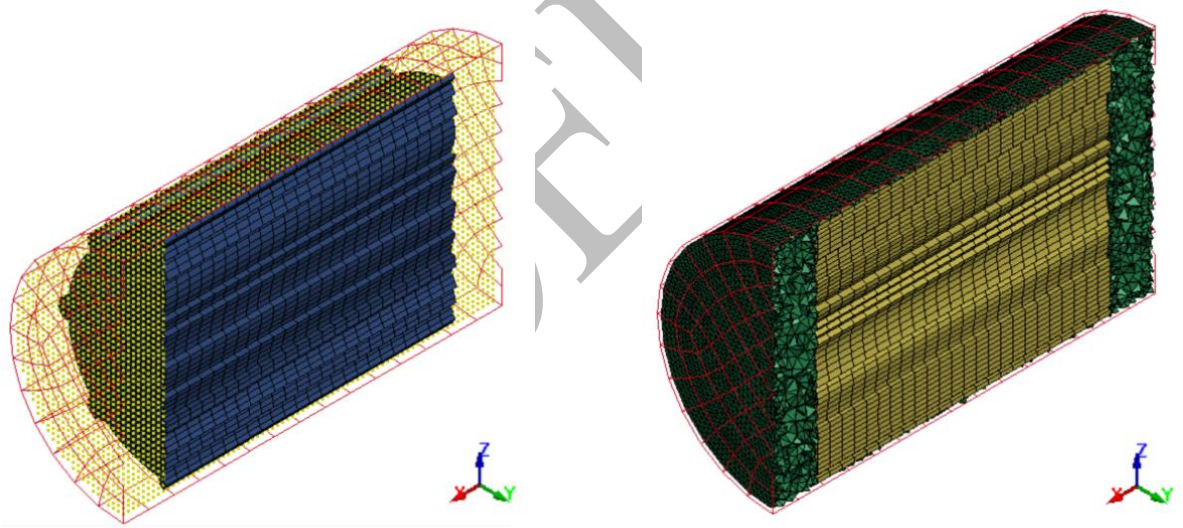
Figure 5. The sledge experimental setup for gelatine insert measurements

The procedure involved filling the container with the gelatine insert, sealing it, and ensuring no leaks. The container was fixed to the setup, and additional lighting was used to enhance image quality. After securing the container, the fluid was subjected to a controlled impact loading (Figure 5), and the camera recorded the fluid and gelatine insert kinematics. The experiment was performed 9 times. The initial velocity of the container was set to approximately 1000

mm/s. The recorded footage was analysed using the Kinovea and TEMA software, which allowed for extracting key quantitative data, such as velocity and displacement.

Numerical setup

In addition, two numerical models (Figure 6) were developed to simulate the experimental setup. The first model utilised smoothed particle hydrodynamics to represent the cerebrospinal fluid, with a 1e-9 t/mm^3 density, pressure cutoff of -1e6 MPa, and viscosity coefficient of 7e-10. The second model simulated CSF using 10-node tetrahedral elements (TET10), characterised by viscoelastic material properties, including a density of 1e-9 t/mm^3, bulk elastic modulus of 2000 MPa, short-time shear modulus of 5e-4 MPa, long-time shear modulus of -1e-4 MPa, and decay constant β of 80.0. The first model incorporated approximately 45,095 smoothed particle hydrodynamics (SPH) particles, while the second used 263,187 TET10 elements.



(a) (b)

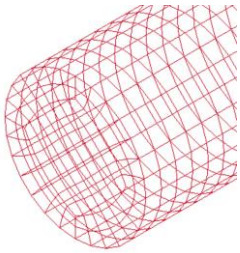
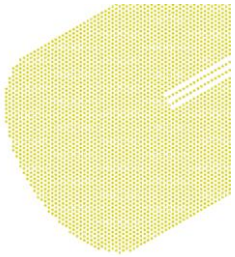
Figure 6. Cross-section of modelled container including liquid fluid and gelatine insert: (a) model with SPH (b) model with TET10 finite elements

Specific boundary conditions were established to ensure the system's behaviour adhered to the desired constraints. The motion was constrained to a forced trajectory, strictly linear, with a constant linear velocity prescribed as an inducted initial condition. Notably, the simulation focused solely on the initial phase, isolating the starting conditions to highlight the immediate system response. The velocity of the finite element mesh nodes was set along the x-axis to replicate the motion of the container in the experiment. The initial velocity of the container was

set at 1000 mm/s (as in the experiment) and 2000 mm/s to imply higher fluid dynamics. Two numerical tests were conducted for each model, varying the initial velocity, with the experimental conditions summarised in Table 2.

After the simulations, the relative displacement of the gelatine insert was calculated. The analysis focused on two extreme points on the finite element mesh of the gelatine insert. The results indicated that the gelatine insert experienced slight compression, reflecting the physical properties of the brain.

Table 2. Mechanical properties of the experiment models components

Component	Material model	Type and no. of finite element/ particles	Density [t/mm^3]	Young's / Bulk modulus [MPa]	Other parameters	Visualisation of FE/SPH model
Container	*MAT_RIGID	Shell 630	1.095e-9	1565	Initial velocity: 1000 mm/s and 2000 mm/s	
Gelatine insert	*MAT_KELVIN_MAXWELL_VISCOELASTIC	Hexa 44,472	1.060e-9	Bulk modulus: 2160	Material parameters: G_0 = 1.0e-2 G_1 = 5.0e-3	
Cerebrospinal fluid	*MAT_NULL	SPH 45,095	1e-9	-	Viscosity coefficient = 7e-10	
	*MAT_NULL	Tetrahedral 263,187	1e-9	-	Viscosity coefficient = 7e-10	

3 Results

3.1 Full head numerical model results

The primary goal of this study was to evaluate the precision and credibility of the finite element model of the human brain by comparing its predictions with experimental data from Hardy et al. [14]. Several simulations were performed, including the one where cerebrospinal fluid (CSF) was modelled as smoothed particle hydrodynamics (Figure 7a), one without CSF (Figure 7b), and one with the additional inclusion of arachnoid trabeculae in the rigid brain-cranial junction

(Figure 7c). The results of these simulations are compared below, highlighting key trends observed in the numerical models.

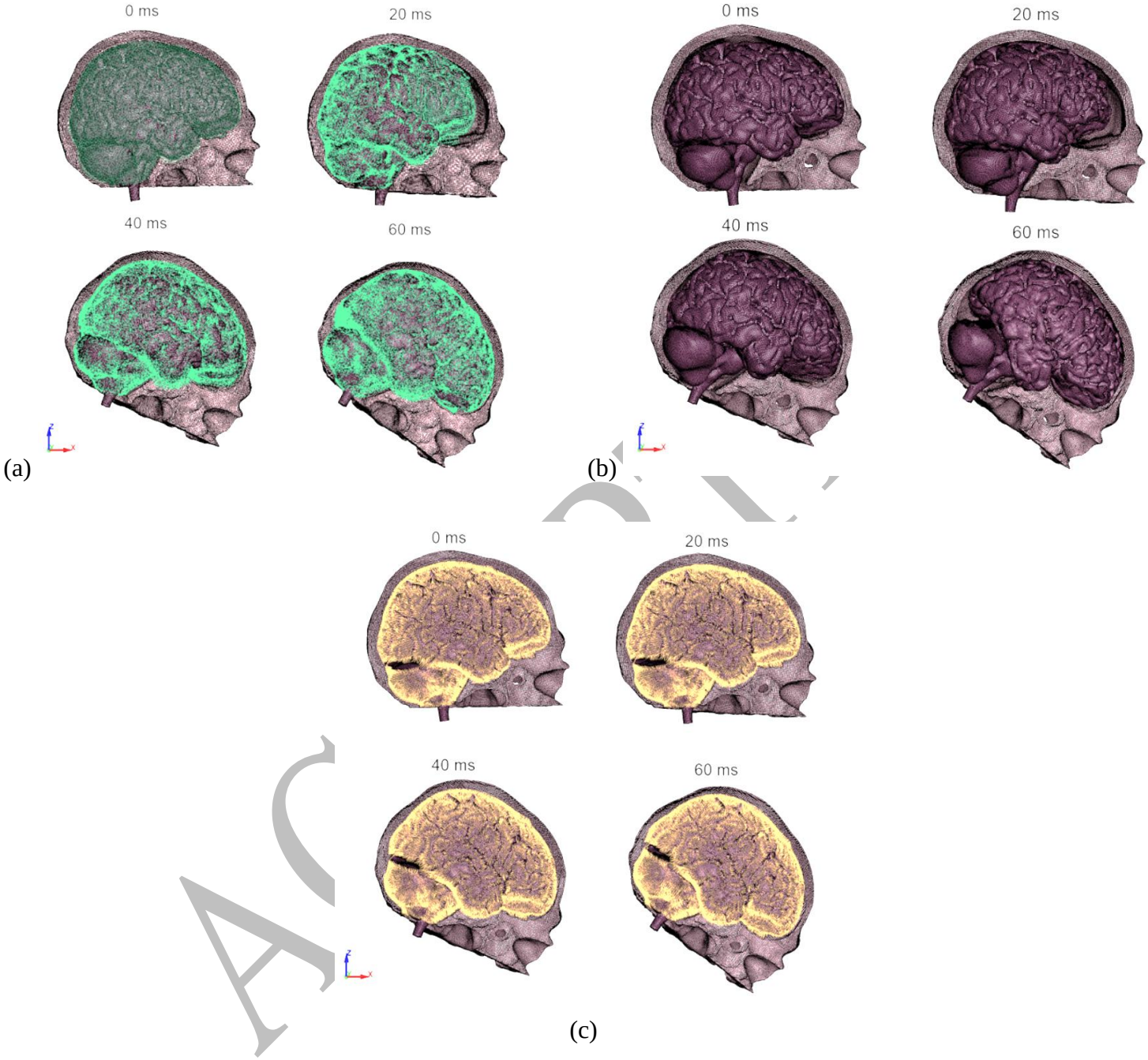


Figure 7. Test results: (a) benchmark aHEAD (CSF as SPH); (b) no CSF – i.e. empty space (c) rigid junction (model with arachnoid trabeculae); the simulations mimic the C755-T2 Hardy experiment

In the simulations, the models with CSF (SPH) and without CSF showed similar deformation of the pia mater, the brain's outermost layer. However, there should be no relative displacement between the brain cortex and skull in a rigid brain-cranial junction. These differences were observed in the simulations as graphs comparing Hardy's experimental data (Figure 8).

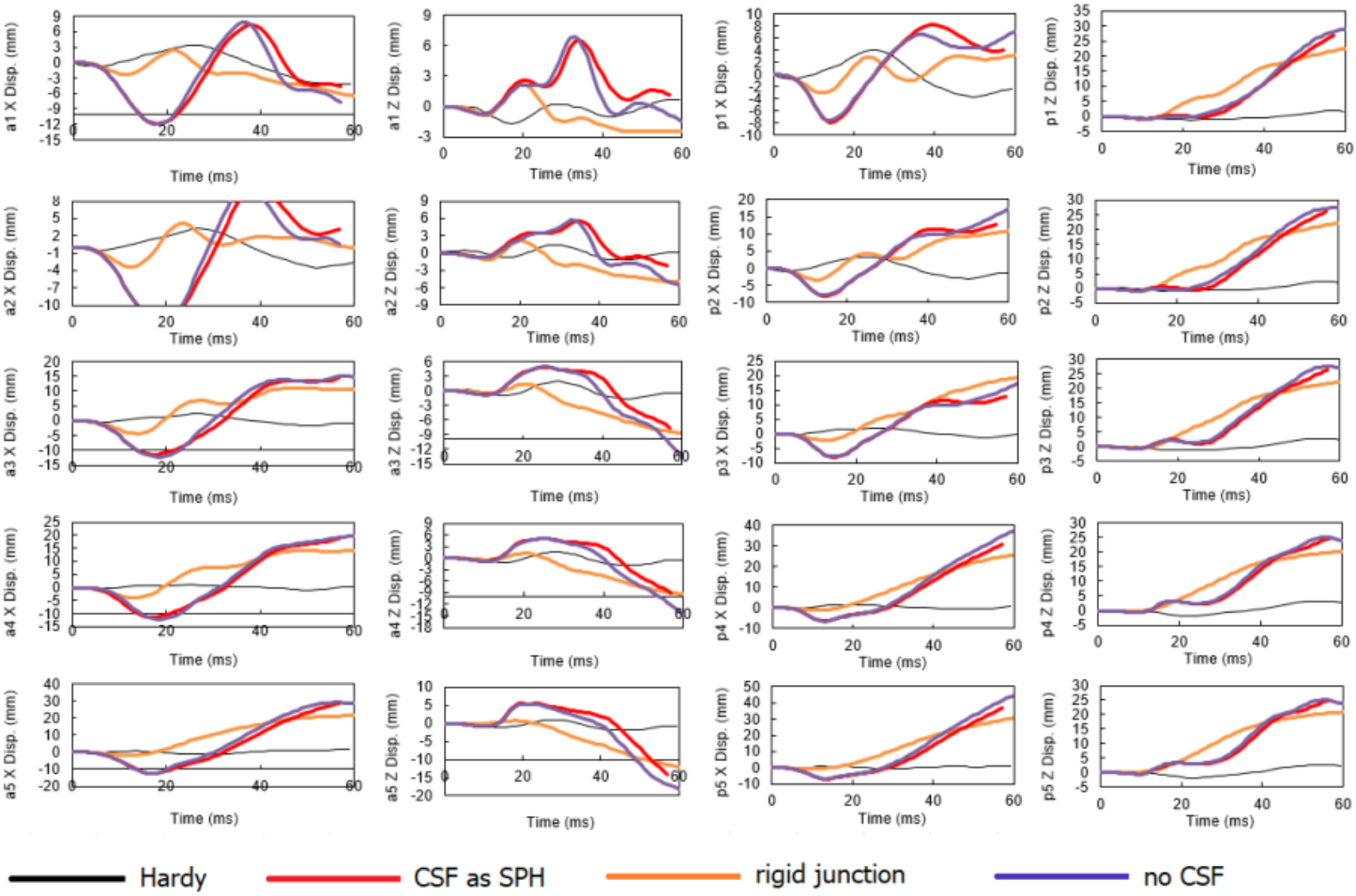


Figure 8. Benchmark aHEAD simulation (CSF as SPH) with two modified head models: "rigid junction" and "no CSF". The displacements are compared with Hardy C755-T2 test results.

3.2 Physical and numerical gelatine insert kinematic results

Using the high-speed camera footage data, the initial displacement (referred to here and after as X-direction displacement) was approximately 2.29 mm (averaged from 9 physical tests). Reference points were established according to the setup to compare the experimental and numerical results. The relative displacement of the gelatine insert concerning the container was numerically simulated for different initial velocities, and the results are presented in Table 3.

Table 3. The relative displacement of the gelatine insert in relation to the container for the simulations and experiments

No. of the test	Numerical approach to CSF	Container initial velocity [mm/s]	Relative displacement of gelatine insert [mm]
LC01	SPH	v_0=1000	2.45
LC02		v_0=2000	6.52
LC04	TET10	v_0=1000	0.07
LC05		v_0=2000	0.22
Physical experiments (averaged)	n.a.	v_0=1000	2.29

4 Discussion

Full-head numerical model

The comparison reveals that for the first 30 milliseconds of impact, the displacement history of the posterior column in the X-direction shows a similar trend between the experimental data and the numerical simulations. However, after the initial 30 ms, significant deviations appear. The numerical models predict much higher displacements than those recorded in Hardy's experiment, indicating a divergence between the computational model and the cadaveric head behaviour. In the Z-direction, the displacement of the posterior column initially aligns with the experimental data for approximately the first 20 ms. After this period, however, the numerical model once again predicts displacements higher than those observed in the experiments, further indicating discrepancies between the simulated and real-world results.

Overall, the numerical simulations exhibit much higher displacement magnitudes in the first 30 ms than the experimental data. This overestimation continues throughout the simulation, with the finite element (FE) models showing excessive increases in displacement beyond the initial period. These results highlight a significant overestimation of brain displacements by the FE models. Unlike the X-direction, the displacement of the anterior column in the Z-direction aligns more closely with Hardy's experimental results. The numerical model's predictions in this direction are more consistent with the experimental data, indicating better accuracy in predicting Z-direction displacements for the anterior column.

The literature on brain mechanics contains limited descriptions of experimental studies conducted on dissected human cadaver specimens, which could serve as a reference for validating numerical models. When considering the comprehensive construction and multivariate testing of materials in these models, the obtained results contrast with those from other FE models in the literature that have been successfully validated against Hardy's

experiments. These alternative models typically use more simplified geometries, employ linear elastic material models, and utilise solid FE elements for fluids such as cerebrospinal fluid (CSF). Often, lower-quality tetrahedral elements are used in these models. The representation of fluids with solid elements might significantly affect the overall behaviour of the model. The stiffness introduced by using solid elements to model fluids could alter brain displacement predictions, potentially explaining the incompatibilities observed in the results. The results were unexpected in several ways. Initially, a rigid brain-skull connection was anticipated to yield results nearly identical to those observed in Hardy's experiments. Even more surprising were the results for the models with CSF modelled as smoothed particle hydrodynamics (SPH) and the model without CSF, which showed almost identical results in comparison to each other.

Gelatine insert kinematics

Focusing on the provided experiment, the numerical simulation relative displacement values exhibit considerable variation compared to the experimental results. In the simulations with SPH (LC01 and LC02), the relative displacement is insignificantly higher than the experimental value of 2.29 mm. In the simulation with an initial velocity of 1000 mm/s (LC01), the displacement is 2.45 mm, only 7% greater than the experimental value. For the simulation with an initial velocity of 2000 mm/s (LC02), the displacement reaches 6.52 mm, 3 times greater than the experimental result. Conversely, the simulations using tetrahedral TET10 elements (LC04, LC05) yield much lower displacement values. The displacement for the simulation with an initial velocity of 1000 mm/s (LC04) is only 0.07 mm, and for the simulation with an initial velocity of 2000 mm/s (LC05), it is 0.22 mm. These values are significantly lower than the experimental result of 2.29 mm, with the TET10 model underestimating the displacement by almost 33 times.

5 Conclusions

The detailed review and application of the aHEAD model underscore the complexity of accurately simulating brain behaviour. The smoothed particle hydrodynamics (SPH) method for cerebrospinal fluid (CSF) modelling allowed for dynamic fluid representation, providing new perspectives on brain deformation. Unlike other finite element (FE) models that use solid elements or brain-cranial interfaces [10,22,29], the current models did not accurately replicate Hardy's findings. Nevertheless, these deviations do not invalidate the tests but underscore the complexity of accurately simulating brain mechanics. The observed discrepancies between simulation results and Hardy's experimental data highlight ongoing challenges. Specifically,

the findings suggest that the protective function of CSF might be less pronounced under extreme conditions than previously assumed. Additionally, the rigid brain-skull junction model yielded results that call into question the applicability of Hardy-based benchmarks to CSF biomechanics.

The physical experiments and numerical simulations revealed limitations in current modelling techniques. SPH models exhibited closely resembling fluid displacement, while tetrahedral elements imposed unrealistic rigidity. Continued refinement of these models, supported by strict experimental validation, is essential for bridging the gap between theoretical simulations and real-world biomechanics.

What is important, the biomechanics of cerebrospinal fluid may lead to misconceptions about its role in brain protection. Instead of providing cushioning through immersion in the surrounding fluid, the results suggest that CSF primarily acts through hydraulic damping and energy dissipation facilitated by fluid exchange in capillary-like channels. This conclusion is supported by experiments with gelatine insert. When the container was positioned upside-down, the gelatine moved downward very slowly, and turbulent flow was observed in the small side channels. Overall, the presented physical experiment and the numerical simulations explain why CSF does not offer soft support for the brain but rather mitigates brain motion through hydraulic damping.

While the current study provides important insights into the numerical modelling of traumatic brain injury (TBI), addressing limitations through ongoing refinement and validation will enhance the fidelity and applicability of the FE models in medical and biomechanical research. Future research should focus on conducting new experiments that explore a broader range of impact conditions and brain states. Incorporating real-time imaging and advanced sensor technologies will help generate more comprehensive datasets for model validation. Furthermore, developing hybrid approaches that combine SPH and traditional FE methods could address the current limitations, offering more accurate and reliable simulations of CSF dynamics and brain biomechanics.

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