

# Cerebrospinal fluid model formulation affects global and local behaviour of the spinal cord submitted to transverse traumatic compression

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**Abstract.** The cerebrospinal fluid (CSF) is a major shock absorber in traumatic spinal cord injuries (SCI). It can be modelled using various fluid formulations, and there is a need to understand their implications on results when performing numerical simulations of SCI. Therefore, four formulations (pressurized volumes – PV, arbitrary lagrangian-eulerian – ALE, smoothed particle hydrodynamics – SPH, and lagrangian – Solid) and two fluid linearity models (Newtonian – N, Mie-Grüneisen – G) were tested in idealised conditions replicating a thoracolumbar burst fracture in a porcine finite element model. The PV formulation proved to be unfit for modelling traumatic SCI, while the solid formulation presented an excessive stiffness of the CSF layer and high computational cost. Both ALE and SPH could be likely candidates, presenting similar results on fragment kinematics, central canal displacement, and anteroposterior deformation of the spinal cord, but dissimilar results in subarachnoid space occlusion, maximal von Mises stress, strain, and volumetric stress. Providing experimental *in vivo* data on the local dynamics of the spinal will further determine the best formulation to use to replicate the mechanism of injury during traumatic SCI.

**Keywords:** Spinal cord injuries, cerebrospinal fluid, finite element modelling.

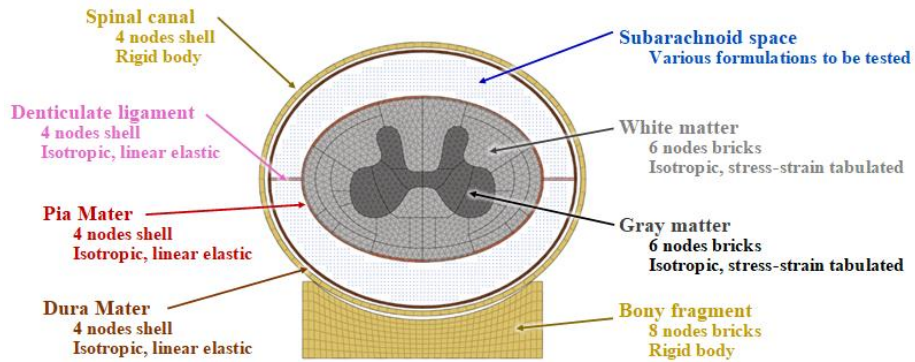
## 1 Introduction

The cerebrospinal fluid (CSF) is a circulating fluid, filling out the subarachnoid space surrounding the central nervous system (CNS). It is assumed to have four main biological roles: homeostasis, waste clearing, nutrition and immune function of the CNS. It also presents two main mechanical roles: buoyancy of the CNS and shock absorber [1, 2]. Its shock absorbing properties play a major role in CNS protection during traumatic events, such as burst fractures inducing spinal cord injuries (SCI), leading to its experimental *ex vivo* study [3, 4] and modelling attempts [5-10]. Out of these

attempts, one study addressed the variety of fluid formulations and their effects when replicating *ex vivo* studies [7]. Although replicating *ex vivo* studies does provide a necessary and robust validation, the understanding of formulation effects is limited to their *ex vivo* environment, the global kinematics of a thrown pellet, and the global compression of the dura mater. However, considering that the spinal cord is enclosed in the spinal canal, the dura mater would not have the same ability to freely dilates under compression, as observed in *ex vivo* experimental studies for which the spinal canal has been altered. Moreover, these studies used a cylindrical pellet as an impactor, which differs geometrically and biomechanically from the bony fragment of a vertebral body impacting the spinal cord during traumatic events. The purpose of the present study is therefore to expand on the investigation of fluid formulations by studying their effects under conditions more closely related to *in vivo* SCI, as well as investigating the internal dynamics of the spinal cord.

## 2 Material and methods

An idealised porcine model of the spinal cord was created and solved with Altair Hyperworks 2019 software suite (Altair Engineering Inc., U.S.A.). Transsectional geometries were derived from histological and magnetic resonance imaging slices of the high-thoracic spine of one of our specimens. The diameters of the elliptical cord were 6.4 by 9.4 mm, and 10 by 12 mm for the elliptical spinal canal, falling within range of values reported in the literature [11]. The dura mater was assumed to be pressed against the spinal canal by the CSF filling out the subarachnoid space, and was linked to the pia mater through denticulate ligaments. These geometries were then extruded over 50 mm, relating to three vertebral levels, in order to limit reflecting effects at the model borders. Based on the proportions of a previous numerical study of burst fractures [9], a full upper-half of vertebral body fragment (mass of 0.27 g) reproducing the typical bony fragment between the pedicles retropulsed in the spinal canal with thoracolumbar burst fractures in humans was modelled and located at the anterior mid-length of the spinal cord (Fig. 1).



**Fig. 1.** Transversal view and brief description of the model

Meshing and materials of solids were based on Diotalevi et al. (2020) [9], and are presented on Fig. 1. Meshing and materials of the CSF were based on Rycman et al. (2022) [7] and the following fluid formulations were studied: pressurised volumes (**PV**), lagrangian (**Solid**), arbitrary lagrangian-eulerian (**ALE**), and smoothed particles hydrodynamics (**SPH**).

Except for PV that does not support it, the CSF was studied both as a linear Newtonian fluid (**N**), and non-linear with the Mie-Grüneisen approach (**G**). Following a convergence study, the global characteristic length of solid meshes was of 0.3 mm on average, and 0.2 mm for the CSF mesh / pitch.

The spinal canal was set immobile, and the fragment was projected onto the dura mater at a velocity of  $4.5 \text{ m.s}^{-1}$  [3, 4]. Rostro-caudal translation of the model extremities were blocked.

Each model was solved over 5 ms on a Dell Precision Tower 7820 (Dell Technologies Inc., U.S.A.). The effects of fluid formulations were then compared based on four groups of metrics:

- **Computation:** computation time normalised to a model without CSF, number of elements normalised to a model without CSF.
- **Fragment kinematics:** displacement diagram (FD), maximal displacement, time at maximal displacement, and final velocity.
- **Spinal cord global dynamics:** central canal displacement diagram (CCD), antero-posterior deformation of the spinal cord diagram (APDSC), subarachnoid space occlusion diagram (SSO), and spinal cord maximal energy.
- **Spinal cord local dynamics:** maximal von Mises stress, maximal von Mises strain, maximal and minimal volumetric stress.

Pairwise comparison of diagrams was achieved by computing two intraclass correlation coefficients (ICC) in a custom MATLAB R2018b script: ICC(A,1) to assess for shape similarities, and ICC(C,1) to assess their proximity.

## 3 Results

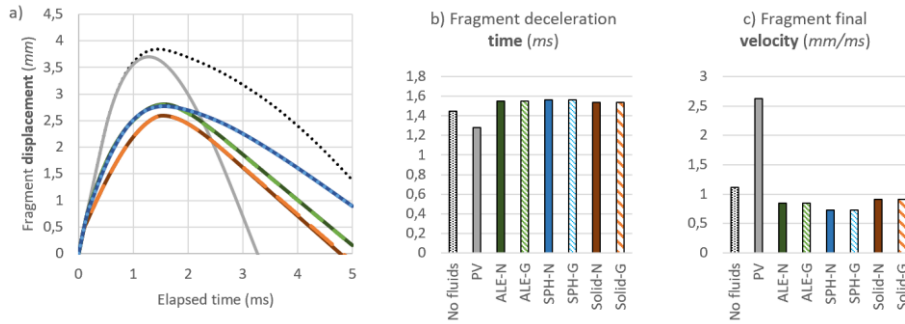
### 3.1 Computation

Implementation of the fluid with the PV formulation didn't require more elements than the model devoid of CSF, and computation was faster (x0.80). Using a solid formulation for fluids required 1.21 times more elements, and computation was 22.3 (Solid-N) and 21.6 (Solid-G) times slower. SPH required 1.55 times more elements and computation was 9.7 (SPH-N) and 10.9 (SPH-G) times slower. ALE formulation had the highest elements number (2.57 times), but computation was only 10.1 (ALE-N) and 10.6 (ALE-G) times slower, which is comparable to SPH formulation.

### 3.2 Fragment kinematics

The fragment deceleration time is comparable for every linearity of ALE, SPH and Solid formulation (range of 1.54 to 1.56 ms), which is slightly higher than a model without CSF (1.45 ms), as shown on Fig. 2.a and 2.b The PV formulation is the only one presenting a shorter deceleration time (1.28 ms).

The fragment trajectory with a PV formulation and the model devoid of CSF is identical when it penetrates the spinal canal. The fragment is then pushed back at a higher velocity ( $2.63$  against  $1.11 \text{ m.s}^{-1}$ ). Every other formulation decelerated the fragment during its penetration, and pushed it back with a slower velocity (range of  $0.73$  to  $0.91 \text{ m.s}^{-1}$ ) (Fig. 2.c).



**Fig. 2.** Fragment kinematics, with: (a) the fragment displacement over time; (b) the duration for the fragment to reach full deceleration and begin being pushed back; and (c) the fragment final velocity when pushed back.

Fluid linearity didn't affect the fragment trajectory in every fluid formulation (ICC A,1 and C,1 > 0.99). Fragment trajectory diagrams for ALE, SPH and Solid formulations appear to be similar in shape (ICC A,1 > 0.75) and proximity (ICC C,1 > 0.92), as shown in Table 1. However, PV is dissimilar to very other formulation, and even to the model without CSF (every ICC ranging from 0.28 to 0.55).

### 3.3 Spinal cord global dynamics

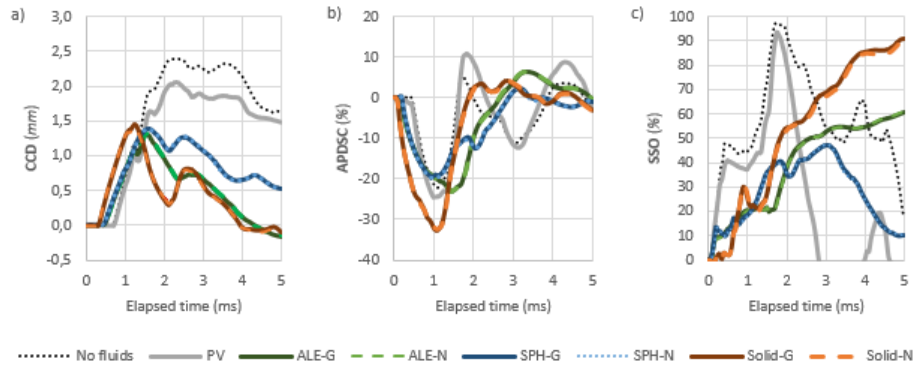
Fluid linearity didn't affect the results for any fluid formulation, in every metrics (ICC A,1 and C,1 > 0.99), as shown on Fig. 2 and Table 1.

The model with the PV formulation behaved more closely to the model without CSF (ICC A,1 and C,1 > 0.91) than to any over fluid formulation for the metrics CCD and APDSC. It also presents the same behaviour over the first 2 ms for the metric SSO, and then over-inflates the dura mater.

Both ALE and SPH formulation are similar up to 1 or 1.5 ms for every metric, but then diverge. For the SSO metric, the SPH formulation allows an inflation of the sub-arachnoid space, but the ALE formulation just limits its occlusion. According to their ICC, they are globally similar in shape (ICC A,1 > 0.63) and proximity (ICC C,1

> 0.82) for the CCD and APDSC metrics, but globally dissimilar in shape (ICC A,1 = 0.26) and proximity (ICC C,1 = 0.33) for the SSO metric.

The SSO increases later with a solid formulation, which is mirrored by an earlier variation of CCD and APDSC metric, when compared to other metrics. With this formulation, the SSO keeps increasing throughout the simulation, and the cord presents a higher peak of global compression (-32.6 %, compared to a range of -19.5 to 24.4 % for other formulations and an absence of CSF)



**Fig. 3.** Spinal cord global dynamics, with: (a) the central canal displacement over time (CCD); (b) the antero-posterior deformation of the spinal cord over time (APDSC); and (c) the sub-arachnoid space occlusion over time (SSO).

As for the global energy sustained by the spinal cord (Fig. 4.d), results follow the same trend. The PV formulation is comparable to a model without CSF (0.16 and 0.17 mJ respectively), ALE and SPH are mutually comparable (0.08 to 0.09 mJ), the solid formulation peaks at a higher value (0.14 mJ) than ALE and SPH, and effects of fluid linearity appear to be negligible.

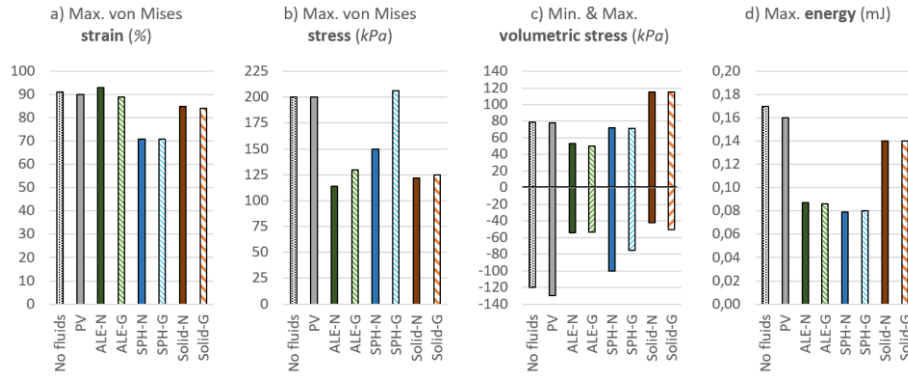
**Table 1.** Summarised results of pairwise comparison of diagrams

| Metric           | --- FD --- |       | -- CCD -- |       | - APDSC - |       | -- SSO -- |       |
|------------------|------------|-------|-----------|-------|-----------|-------|-----------|-------|
| ICC              | (A,1)      | (C,1) | (A,1)     | (C,1) | (A,1)     | (C,1) | (A,1)     | (C,1) |
| G vs. N          | >0.99      | >0.99 | >0.99     | >0.99 | >0.99     | >0.99 | >0.99     | >0.99 |
| ALE vs. SPH      | 0.87       | 0.93  | 0.63      | 0.82  | 0.86      | 0.87  | 0.26      | 0.33  |
| ALE vs. Solid    | 0.96       | 1.00  | 0.82      | 0.82  | 0.77      | 0.77  | 0.76      | 0.88  |
| SPH vs. Solid    | 0.75       | 0.92  | 0.41      | 0.54  | 0.72      | 0.72  | 0.14      | 0.23  |
| PV vs. No fluids | 0.28       | 0.44  | 0.93      | 0.98  | 0.91      | 0.92  | 0.30      | 0.54  |
| PV vs. Solid     | 0.51       | 0.55  | 0.00      | 0.00  | 0.67      | 0.69  | 0.00      | 0.00  |
| PV vs. ALE       | 0.48       | 0.54  | 0.11      | 0.23  | 0.45      | 0.47  | 0.00      | 0.00  |
| PV vs. SPH       | 0.30       | 0.37  | 0.42      | 0.60  | 0.46      | 0.47  | 0.03      | 0.03  |

### 3.4 Spinal cord local dynamics

Effects of fluid linearity are again negligible, except on the maximal von Mises stress for ALE (114 kPa for ALE-N, 130 kPa for ALE-G) and SPH (150 kPa for SPH-N, 206 kPa for SPH-G) formulations, and on the outer bounds of volumetric stress for SPH formulation ([-100; 72] kPa for SPH-N, [-75; 71] kPa for SPH-G).

The PV formulation presents comparable results to the model without CSF, and every other formulations reach different values of maximal von Mises strain (from 71 to 93 %), stress (from 114 to 206 kPa), and outer bounds of volumetric stress (from -100 to 115 kPa), as shown on Fig. 4.a, 4.b and 4.c.



**Fig. 4.** Spinal cord local dynamics and global energy, with: (a) the maximal von Mises strain recorded; (b) the maximal von Mises strain recorded; (c) the outer bounds of volumetric stress recorded; and (d) the maximal energy sustained by the spinal cord.

## 4 Discussion

This study provided insight on fragment kinematics, spinal cord global dynamics, and spinal cord local dynamics, based on numerical simulations related to *in vivo* burst fractures.

The PV formulation, although quick to compute, presents close to no effects on results when compared to the model without CSF, and introduces an erratic behaviour of the dura mater (Fig. 3.b). It would therefore be a poor candidate for implementation in numerical model for spinal cord injury due to a thoracolumbar burst fracture, which corroborates conclusions from Rycman et al. (2022) [7].

Our results give evidence to an excessive stiffness of the CSF layer when using a solid formulation, further corroborating Rycman et al. (2022) [7]. When compared to other formulations, it is shown by a delay in the increase of the SSO metric (Fig. 2.c), a premature increase of the CCD metric (Fig. 2.a), and a higher peak of global compression of the spinal cord (Fig. 2.b). These elements all imply that the CSF can't dissipate energy as much as other formulation, and therefore transfers more of it to the spinal cord, which is validated by direct measurement of the maximal energy sus-

tained by the spinal cord (Fig. 3.d). This inability to dissipate energy is most likely due to the inability of a lagrangian mesh to properly sustain hyperdeformations, and flow. Considering this excessive stiffness and a computation time twice as long as for ALE and SPH formulations, a solid lagrangian formulation is also an unlikely candidate for modelling the CSF in the studies of traumatic spinal cord injuries. It is however often used for modelling brain injuries, but these studies don't face comparable traumatic conditions [12]. On one hand, brain injuries usually result to a non-penetrative impact to the head, where the CSF is a relatively thin layer compared to the size of the brain and the skull. On the other hand, spinal cord injuries usually result to a penetrative impact where a fragment, or an osteophytic feature, will displace a thick layer of CSF to directly impact the spinal cord.

The ALE and SPH formulations present the same results over the first 1 to 1.5 ms of simulation for the fragment kinematics and the global spinal dynamics. This lapse of time relates to the deceleration of the fragment, before it gets pushed back and isn't in contact with the dura mater anymore. Results are then similar, except for the occlusion of the subarachnoid space (Fig. 3.c), due to a different behaviour of the dura mater under these formulations. Indeed, SPH particles didn't stick to the dura mater, which let it freely bounce back in place and artificially decrease the SSO metric. Particles could be tied to meninges, as recently proposed in a brain model, and SPH parameters could further be investigated to prevent inherent instability of particles [13]. Then, ALE and SPH formulations exhibit different results for the local dynamics of the spinal cord.

Finally, this study can only present likely and unlikely candidates for fluid formulations. A lack of experimental data on the local dynamics of the spinal cord and the CSF flow under SCI can't provide a proper verification and validation of formulations.

## 5 Conclusion

Out of the existing fluid formulations, ALE and SPH are the most likely candidate for modelling the CSF in traumatic SCI. However, these formulations still exhibit different results in the local dynamics of the spinal cord and the dura mater behaviour. Therefore, proper selection of a formulation would require experimental data on the global and local dynamics of the spinal cord under conditions more closely related to *in vivo* SCI.

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