Comparison of Spinal Cord Impact Response with a Cerebrospinal Fluid Layer Using Three Numerical Formulations

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I. INTRODUCTION

Acute spinal cord injuries (SCIs) have a global annual occurrence rate of 14–40 per million population [1]. The primary mechanism of injury involves physical damage of the nervous tissues [2]; however, experimental findings have indicated that neurological sequela can occur without radiographic abnormalities of the neural tissues [3-4]. Compression of the spinal cord may result from fracture or dislocation of the vertebra, and studies have suggested that the cerebrospinal fluid (CSF) layer plays a protective role for the spinal cord during impact [5]. Computational Human Body Models (HBMs) are an important tool to investigate the potential for spinal cord injury [6]. However, HBMs require numerical representation of the CSF layer for accurate prediction of spinal cord compression upon impact. Most existing computational spinal cord models either do not incorporate the CSF layer, or instead include a simplified representation based on an Arbitrary Eulerian-Lagrangian mesh (ALE) [7] or Lagrangian mesh [8]. Both of these methods have limitations in modelling CSF. The ALE method introduces inaccuracies when filling volumes occur between small gaps [9], whereas a Lagrangian mesh may not be suitable for modelling very large deformations associated with flow of the CSF. The aim of the current study was to compare three numerical formulations for modelling the CSF layer: smooth particle hydrodynamics (SPH), ALE, and the traditional Lagrangian method, assessed using available experimental spinal cord impact data.

II. METHODS

The finite element (FE) model (Fig. 1) included the tissue complex (spinal cord, pia mater, dura mater, CSF) and rigid parts representing the impactor (pellet) and support (posterior wall), as reported in experimental impacts on bovine tissue complex specimens [10]. Prior to the impact, an axial tensile strain of 8% was applied to the spinal cord, pia mater and the dura mater, as was done in the experiments. An initial velocity of 4.5 m/s was applied to the pellet, matching the experimental value. The trajectory of the pellet (Fig. 2) and spinal cord compression were extracted from all models and compared to the experimental results. Four versions of the model were created with three different CSF numerical representations: SPH, ALE and Lagrangian mesh, and one model without CSF. The CSF was modelled as a fluid with a bulk modulus of 2.19 GPa and a density of 1040 kg/m^3 [11]. Previously published material properties were used for the spinal cord, pia mater and dura mater in this study [12] and were not calibrated to match the impact experimental results. The FE models were solved using a commercial explicit FE solver (LS-DYNA v9.2, LST, Livermore, California, US).

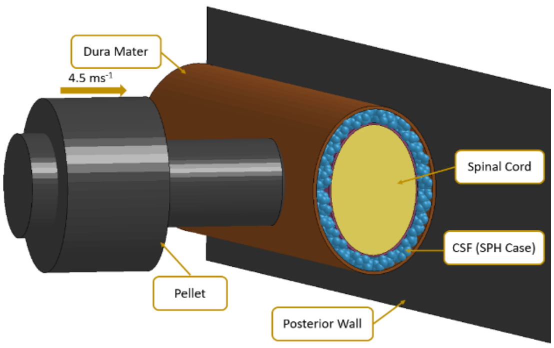

Fig. 1. Tissue complex and pellet FE model.

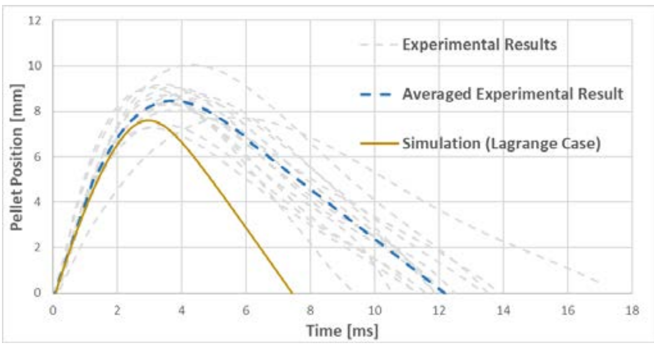

Fig. 2. Trajectory of the pellet impacting the tissue complex, CSF modelled with Lagrangian mesh.

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III. INITIAL FINDINGS

The three numerical formulations investigated for the CSF layer gave different results in terms of the maximum deformation of the tissue complex (Fig. 3). All CSF implementations were within the range of the reported experimental data, except for the model without the CSF layer. The CSF implementation with the SPH method was closest to the average experimental results, while the stiffest compression response was observed when the CSF was modelled with the Lagrangian mesh.

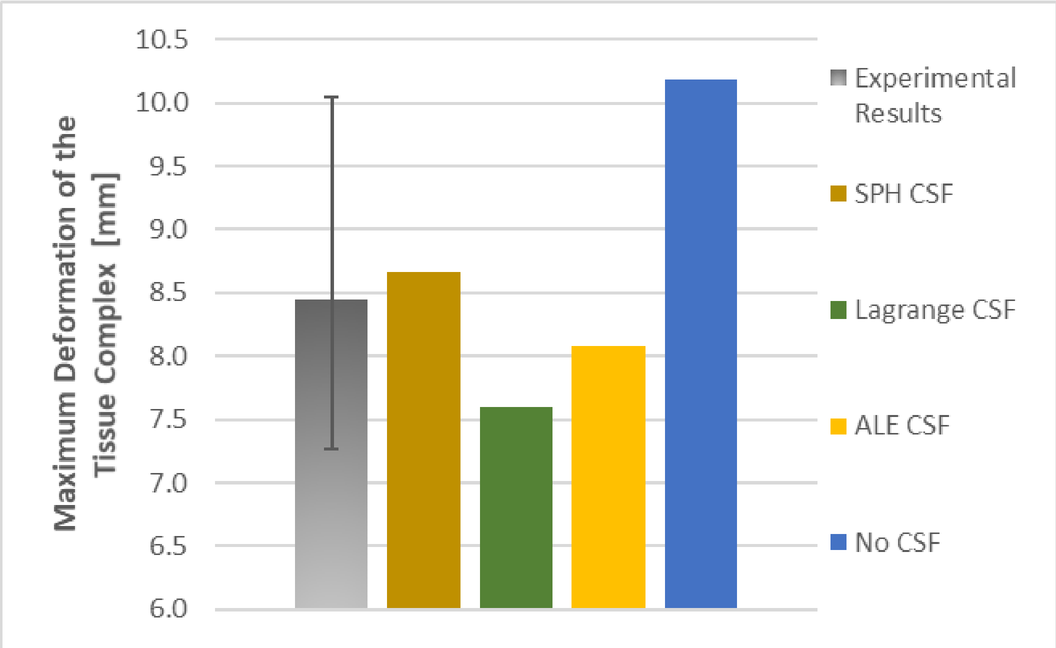

Fig. 3. Maximum deformation of the tissue complex; error bars indicate the range (maximum and minimum) of the experimental data.

IV. DISCUSSION

The CSF layer reduced the total deformation of the entire tissue complex by 20% on average, compared to the model without the CSF layer. More specifically, spinal cord compression during impact was reduced by 12.5% using the SPH method, by 20% for the ALE model and by 6.5% using the Lagrangian mesh, compared to the model without the CSF layer. The ALE formulation of the CSF layer allowed material to move through the fixed grid as the CSF was squeezed between the dura mater and the pia mater during impact. This effect resulted in the largest reduction in the spinal cord compression for the ALE formulation. The Lagrangian mesh restricted CSF flow, leading to an overall stiffer response and lower compression of the tissue complex. Using SPH to model the CSF layer enabled fluid to flow out of the impact zone, as reported in the experimental data, resulting in nearly full collapse of the CSF layer. Overall, increased compression of the CSF layer during impact due to fluid flow resulted in reduced deformation of the spinal cord, potentially indicating the protective role of the CSF layer.

V. ACKNOWLEDGEMENTS

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VI. REFERENCES

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