Comparative Studies of Finite Element Modeling of Human Head Impact

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Abstract – In computational impact biomechanics, several human head injury models have been proposed using explicit finite element techniques. The analyses of these models are often carried out by utilizing several commercially available finite element codes. While the classical theories behind these commercial codes are essentially the same, yet they are different in terms of technical treatments of material modeling, element formulation, contact search algorithms, etc. It is not clear whether there is a difference in model response, how much is the difference if there is any, when the same model was analyzed with different explicit codes. In addition, it could be misleading by the simulations done with different explicit finite element codes without pre-knowledge of the performance of these codes. The new challenge in this area is then on how to best use the technology to meet the research needs. In this study, a 50th percentile finite element human head model in its Pamcrah version was transposed to Ls-Dyna version to perform a comparative study. The responses of these models were compared against each other and the results showed that they are comparable.

Keywords: Human head model, finite element, head Injury, impact.

1 Introduction

Computational method has always been an invaluable tool in supplemental to experimental method in scientific research. This is especially true in the field of impact biomechanics because any experiment involving living thing is undesirable by many senses. The current state of computer-aided technology has made computational method even more viable in impact simulation of head injury. Computational methods/software; graphical methods/software; and simulation technologies have provided a significant opportunity for biomechanical researchers to develop sophisticated mathematical models, in particular Finite Element (FE) models, of the human head. The computer systems and tools have allowed investigators to perform parametric studies cost-effectively that otherwise would be very difficult to obtain through experimental studies involving cadavers, or other human surrogates. Computational models also furnish head trauma analysts with a powerful tool to extrapolate major experimental findings in human cadavers and other human surrogates. In addition, these models provide head injury researchers with a method to establish the relationship between neurological deficit and mechanical dosage. Moreover, these models overcome some of the difficulties associated with experimental models such as, test repeatability and societal issues when animal models are used.

It is currently feasible and desirable to develop sophisticated head injury models to incorporate more anatomical details of the human head, utilizing non-linear, large deformation, finite element techniques and a constitutive representation of head tissues. The current trend in head trauma analysis is to identify human head impact responses, injury mechanisms, injury tolerances and injury criteria through detailed finite element models of the human head. Future studies of head injury response and injury mechanisms will rely less and less on human cadavers and other human surrogate models. Therefore, expectations of finite element human head models are high.

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As technologies advance, more comprehensive and sophisticated finite element models of the human head are expected to be introduced. Today, with the availability of MRI technology and advanced graphical technology, it is possible to create a human head model, which closely simulates the geometry of a real head. Thanks to the advancement of computational methods and software, several commercial finite element codes are available to ease the computation burdens and meet the analysis needs. Numerical problems such as stability and shear-locking associated with the earlier finite element head models can be resolved by utilization of powerful commercial, explicit FE programs such as Abaqus™, MARC™, Pamcrash™, LS-Dyna™, and Radioss™. Furthermore, many sophisticated contact algorithms have allowed modelers to simulate various realistic impact conditions and boundary conditions.

While the basic principles and numerical algorithms are essentially the same for the above mentioned commercial codes, yet they are different in terms of technical treatments of material modeling, element formulation, contact search algorithms, etc. The new challenges then, are: how to best use the technology to meet the research needs, and how to best feed the model into the technology to search the real physics, not the best approximation. It is not clear whether there is a difference in model response, how much is the difference if there is any, when the analyses were performed with different explicit codes. Accordingly, the purpose of this study is to explore if these differences exist. In this study, a 50th percentile finite element human head model in its Pamcrah version was transposed to Ls-Dyna version to perform a comparative study. The responses of these models were compared against each other to see if they are comparable.

2 Methods

The finite element human head model used in this study was published in 1993 by Ruan et al [1]. This is the first finite element human head model based on explicit finite element technology. The skull is a three-layer structure with the diploë in between the inner and outer tables. The rough floor of the cranial base is designated by the frontal fossa, middle fossa, and posterior fossa. The cerebral-spinal-fluid (CSF) is surrounding the brain surface and runs through the foramen magnum surrounding the brain stem. The brain includes the left and right hemispheres, cerebellum, midbrain, and brain stem. The meninges modeled were the dura mater, and falx cerebri. Fig. 1 shows a midsagittal view of the finite element human head model.

Model validation was accomplished by comparing the model predicted head impact force, head acceleration, coup and contrecoup pressures, and pressures in three other locations of the brain with those from experimental measurements by Nahum et al [2]. Details validation of the model can be found in Ruan et al [1]. Recent applications of this model in biomechanical studies of head injury can be found in Ruan and Prasad [3, 4], where the influence of variations in the thickness of the CSF layer and the thickness of the skull on head impact response was investigated. In the present study, the analysis of this finite element human head model was carried out by using Ls-Dyna explicit code. The resulting responses were compared with previous analysis using Pamcrash.

The loading condition as shown in Fig. 1 used in both Pamcrash and Ls-Dyna runs was a free head impacted by a rigid cylinder with a mass of 5.23 kg and an initial velocity of 6.33 m/s, the same condition used in the model validation by Ruan et al [1].

Table 1 Comparison of material models of the head tissues and contact algorithm

Tissues And Contact	Material models		Remark
	Pamcrash	Ls-Dyna	
Outer Table	Type 1	MAT_PLASTIC_KINEMATIC	Bilinear stress-strain
Inner Table	Type 1	MAT_PLASTIC_KINEMATIC	Bilinear stress-strain
Diploë	Type 1	MAT_PLASTIC_KINEMATIC	Bilinear stress-strain
CSF	Type 1	MAT_PLASTIC_KINEMATIC	Bilinear stress-strain
Brain	Type 5	MAT_VISCOELSTIC	$G(t)=G_\infty+(G_0-G_\infty)e^{-\beta t}$
Dura	Type 102	MAT_PLASTIC_KINEMATIC	Bilinear stress-strain
Scalp	Type 103	MAT_PIECEWISE_LINEAR_PLASTICITY	Piecewise stress-strain
Impactor	Type 100	MAT_RIGID	No element calculation
Contact	Type 4	CONTACT_SURFACE_TO_SURFACE	Node-to segment

Material models for the head tissues are shown in Table 1 for both Pamcrash and Ls-Dyna. The values assigned to those material models are the same for both codes and they can be found in Ruan et al [1]. More detailed descriptions of these material models can be found from the Pamcrash and Ls-Dyna user's manual.[5, 6]

Stiffness method for hourglass control was available for both Pamcrash and Ls-Dyna and it was used in the analyses. Energy balance was checked to ensure that there is no energy lost due to the ill-defined contact or other errors. Time-history of impact forces, head accelerations measured at the center-of-gravity, intracranial pressures at the coup and contrecoup sites were outputted from the models and compared against each other. Pressure distribution were also outputted as contour plots and compared.

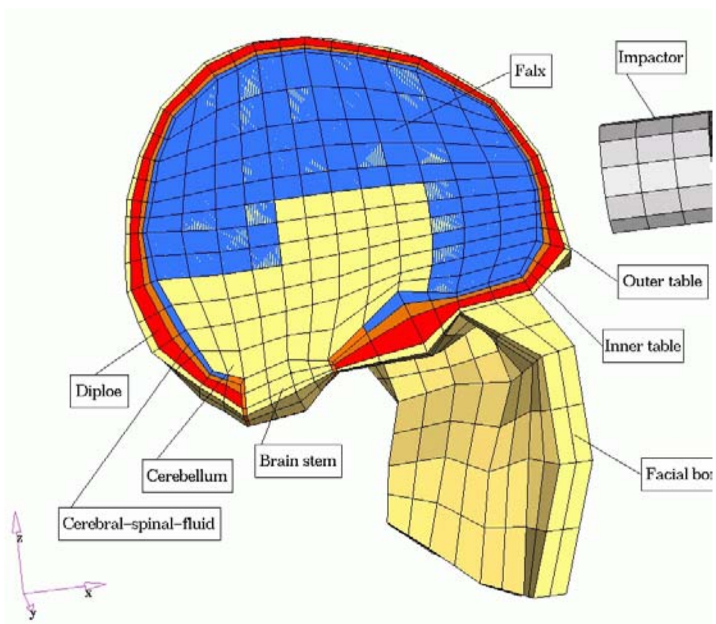

Fig. 1 Finite element model of a 50th percentile human head

3 Results

Since model validation of this human head model has been reported previously by Ruan et al [1], model comparisons with experimental cadaver data are not shown in this section. The followings present model responses from the analyses by both Pamcrash and Ls-Dyna.

Fig. 2 shows impact forces from both Pamcrash and Ls-Dyna runs. As indicated in Fig. 2, impact forces predicted by both Pamcrash and Ls-Dyna are very close. This is also true for the resultant head accelerations, which are shown in Fig. 3. However, there is a discrepancy in coup pressures, as shown in Fig. 4. The difference in coup pressures predicted by Pamcrash and Ls-Dyna is about 10%, with the prediction from Pamcrash is higher than that from Ls-Dyna, as seen in Fig. 4.

There is also a discrepancy in contrecoup pressures predicted by Pamcrash and Ls-Dyna. The difference is also in the 10% range, as seen in Fig. 5. The predicted negative pressure (contrecoup pressures) by Ls-Dyna is higher than that predicted by Pamcrash, while the predicted positive pressure (coup pressure) by Pamcrash is higher than that predicted by Ls-Dyna. This implies that the zero pressures point predicted by Pamcrash is shifted to the contrecoup side compared with the one predicted by Ls-Dyna. Previous studies by Kopecky and Ripperger[7] and Ruan et al[8] have shown that the shifting of zero pressure point could be contributed by stiffness of the skull, the incompressibility of the cerebral-spinal-fluid (CSF) and the brain tissues.

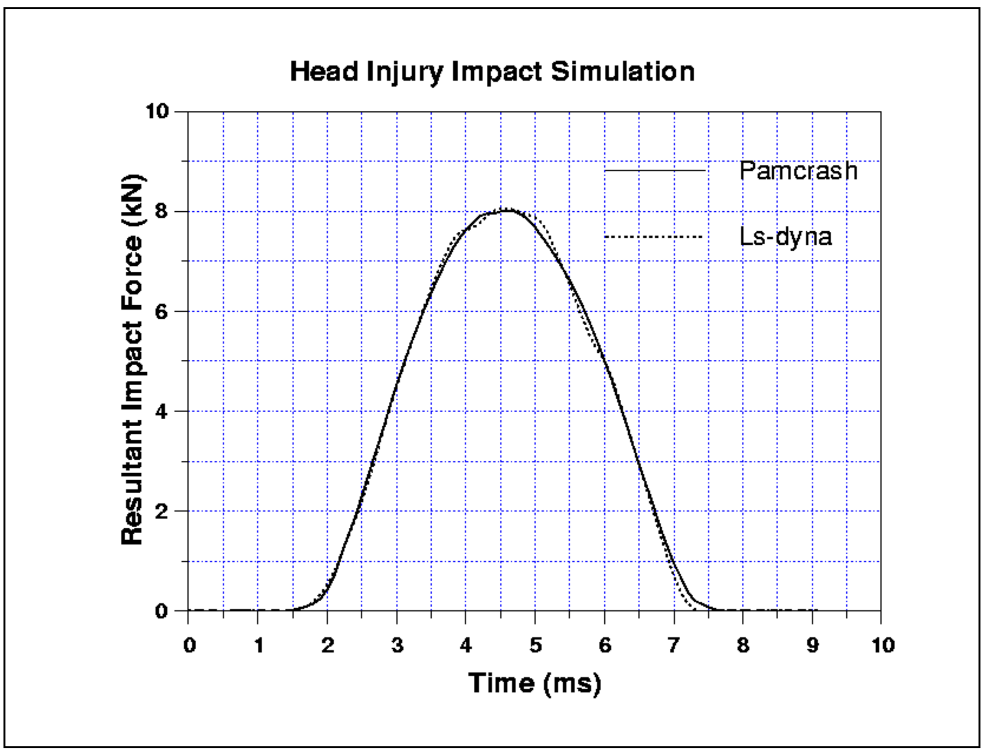

Fig. 2 Impact force comparison

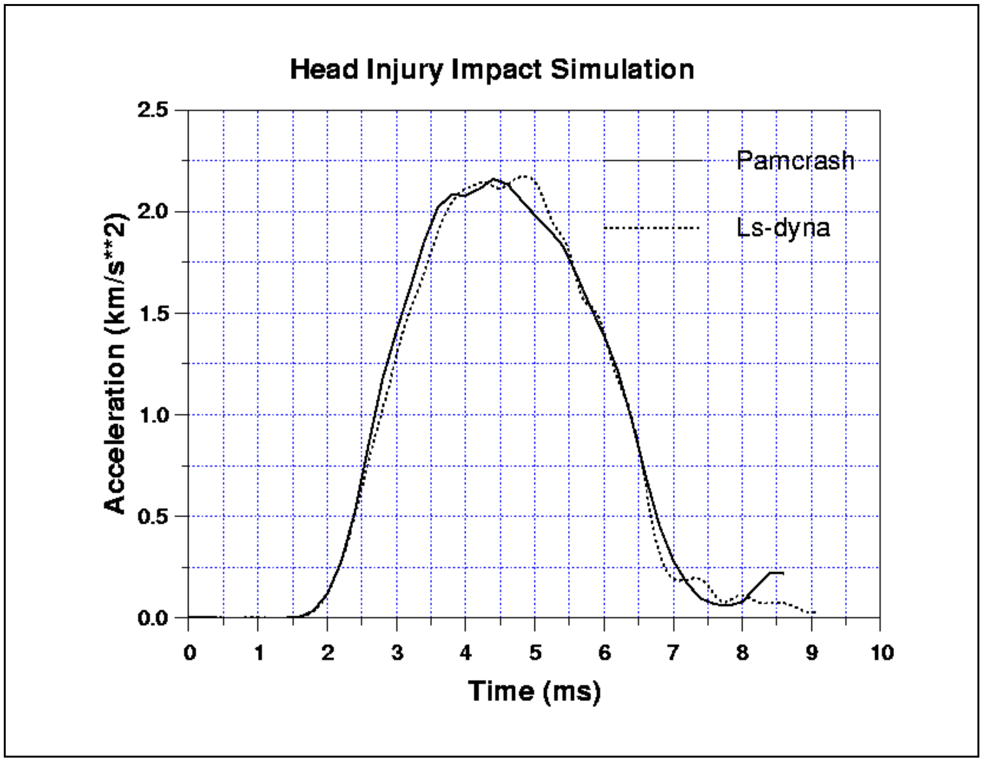

Fig. 3 Resultant head acceleration comparison

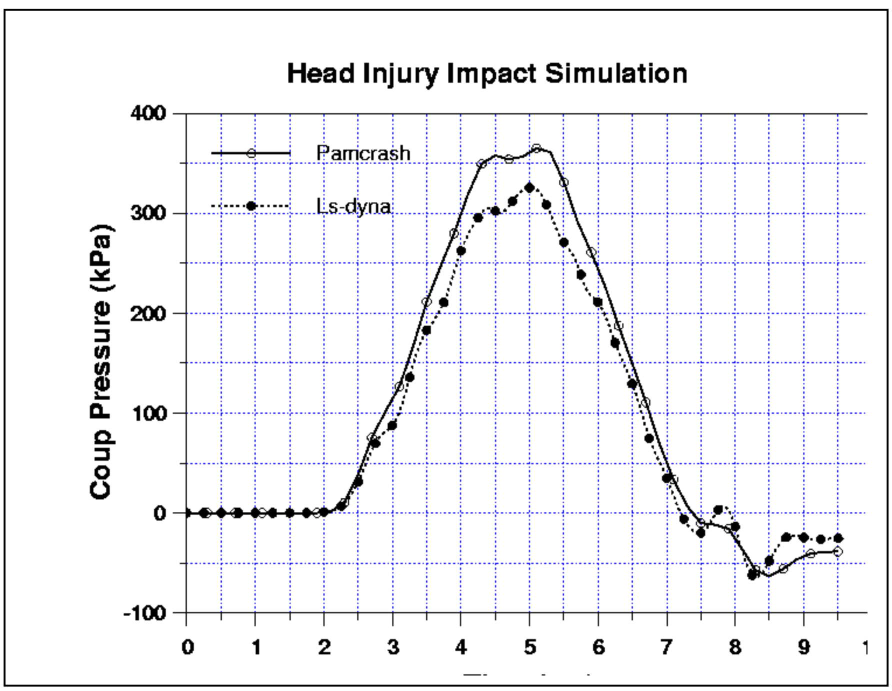

Fig. 4 Coup pressure comparison

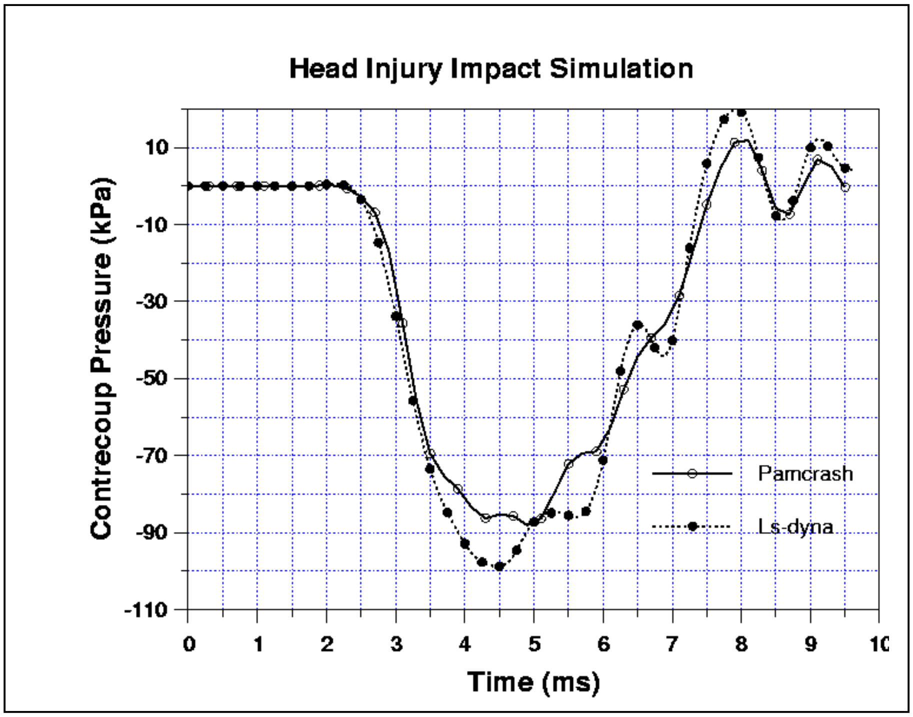

Fig. 5 Contrecoup pressure comparison

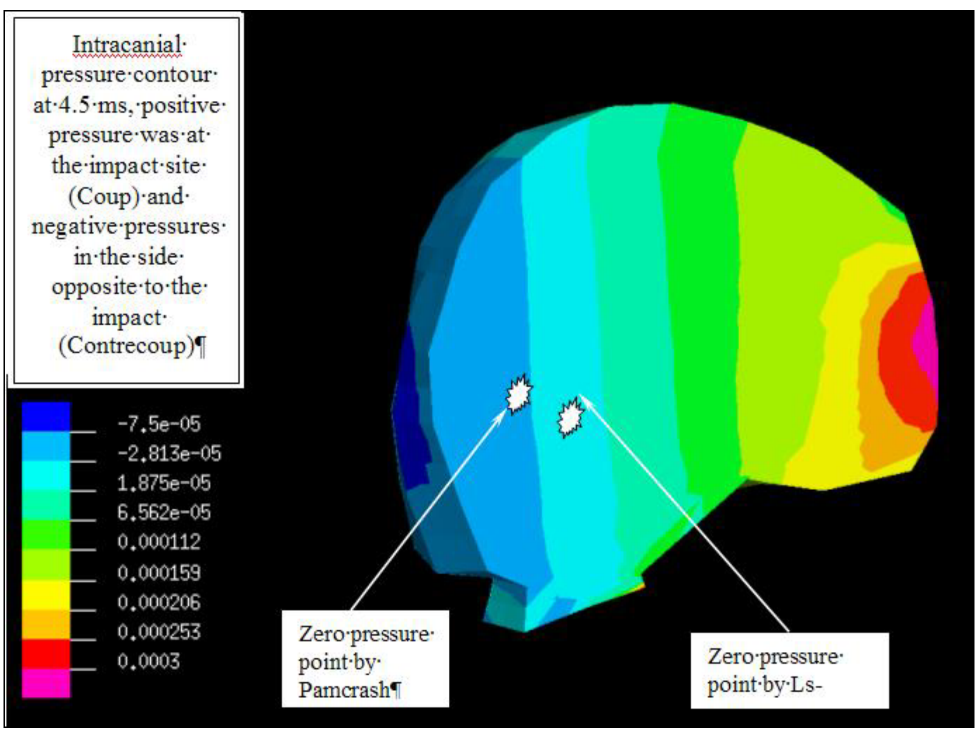

Fig. 6 Intracranial pressures contour at 4.5 ms

Intracranial pressures contour plot is shown in Fig. 6. During a direct head impact, intracranial pressure was initiated at the impact point (coup) and propagated from the impact site to the side opposite to the impact (contrecoup) through the brain and skull tissues. As a result, positive pressure (coup pressure) was generated at the impact site and negative pressure (contrecoup pressure) was at the opposite side to the impact. Along the axis of the impact, we can always find a point where pressure is zero (pressure change sign here). As mentioned above the zero pressure point predicted by Pamcrash tended to close to the contrecoup in compared with the one predicted by Ls-Dyna, as shown in Fig. 6.

4 Discussion

While Pamcrash and Ls-Dyna did give very similar results for the same model, yet a little difference does exist in terms stress computation. The causes for the difference could be attributed to several factors yet the reasons for these causes are beyond the scope of this report. Experienced bioengineers should have no problem to choose which codes can meet their research and application needs. The question that which codes the modelers should trust the most should not be answered until the uncertainty of input data of the model and the issues of trustworthiness of the computation results are resolved. Other important aspect in this regard is that impact bioengineers need to know whether the software tool can solve the problem in question and how to validate the results vigorously.

There is no short-cut in the process of model validation. The strategies and practical ways to validate the modeling results could be to validate the computation results against classical theory, experimental data, published data, clinical data, real world accident data, and the similar computation done by others. As a matter of fact, the current study is in a way to validate the finite element human head model developed by Ruan et al [1] more than a decade ago.

Finite element technology has been regarded as one of the industrial success stories in the last two decades. This is also true in the area of impact biomechanics. With the availability of many powerful commercial finite element codes and the advancement of computer technology, many more sophisticate finite element human head models have been developed in recent years. As technology advanced, computational power will increase substantially yet the requirements for increased accuracy and reliability in head injury simulation are also increasing. The new challenge is then on how to best use the technology to meet these research needs and to deliver the promise of simulation-based injury analysis.

5 Conclusions

A previously developed finite element human head model has been used to compare the analysis responses performed by Pamcrash and Ls-Dyna explicit finite element codes. Model responses showed similar computation results in terms of impact forces and resultant head accelerations, but with slight difference in terms of intracranial stresses.

Although many powerful commercially available finite element codes are handy to meet the impact injury analysis needs, yet they are black-box liked, the sagacious choice of them needs a complete understanding of the problems in question and how to validate the results.

6 References

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