Adv. Theor. Appl. Mech., Vol. 3, 2010, no. 7, 339 - 347

Parametric Study of Closed Head Injuries

Hongxi Chen and Yunhua Luo[1]

[1,2]Department of Mechanical and Manufacturing Engineering
Faculty of Engineering, University of Manitoba
Winnipeg, MB R3T 5V6, Canada

Abstract

In this paper, a two-dimensional finite element model was constructed and used to study several factors affecting closed head injuries (CHI). The finite element model was constructed using magnetic resonance images to faithfully represent the geometry and material properties of the head. The investigated factors include: cranial elasticity modulus, duration of impact and contact area of impact. The maximum peak values of mechanical responses were used to measure the effects of the investigated factors. From the obtained results, we observed that these factors have considerable influence on strain fields in the brain and stress fields in the skull. By increasing either cranial elasticity modulus, or impact duration, or contact area of impact, the maximum peak strains in the brain can be significantly reduced. But this trend does not continue when these factors rise to a certain level. Further increase will not lead to any additional reduction. Conclusions obtained from the investigations may be helpful for improving protective devices such as helmets.

Keywords: Closed Head Injury (CHI), Impact, Finite Element Model

1 Introduction

Head injury (HI) is one of the most critical public health problems. It is a significant contributor to mortality and disability worldwide. It causes

---

[1] Corresponding author; E-mail: luoy@cc.umanitoba.ca

catastrophic and grievous alteration to the patients' life, and leads to numerous economic loss as well. Based on estimation from the surveys, in US, about 75,000 to 100,000 people died each year due to head injuries. Medical costs spent on hospitalization, acute care and various rehabilitation services are estimated at $48.3 billion per year [1]. Traffic accidents are the biggest contributor of head injuries. Summarized from the surveys done by Injury Surveillance, Health Canada and Road Safety in 2001, nearly every one minute one person is killed in a traffic accident, and approximately 70% of the death is due to head injury [2].

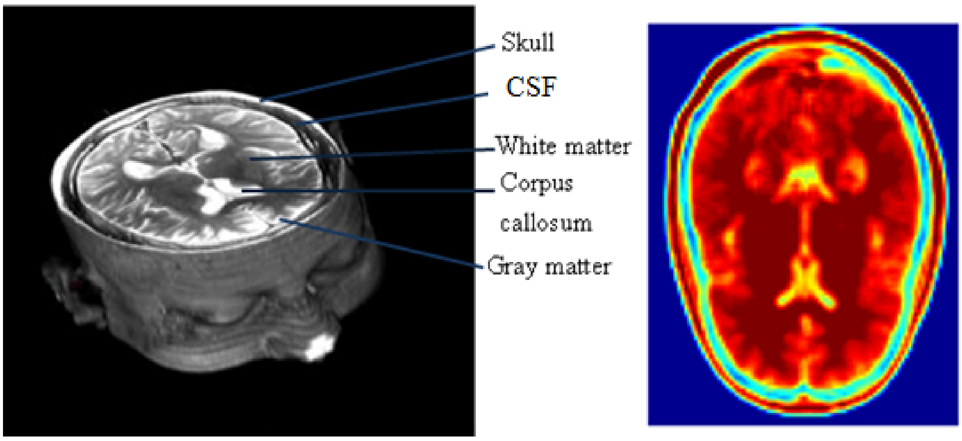
Considerable research effort has been made to study mechanical mechanisms involved in head injuries and more effective way for preventing head injuries. Protective devices such as helmets, seat belts and air bags have been widely adopted for preventing head injury. However, the incidence of head injury is still high. Take helmet as an example, it is considered as one of the most effective ways to protect the head. In war fields, helmets are indispensable protective gears. Helmets are required in recreational activities and sports, e.g. horse racing, American football, ice hockey, and rock climbing. Protective helmet is also required in dangerous work activities such as construction and mining. The effectiveness of motorcycle helmet in reducing motorcycle-related accident death was 29% during 1972-1987, while during 1993-2000 the effectiveness increased to 37%, possibly due to improvement in helmet design and materials [3]. A lot of researches are still going on to further improve the effectiveness of head protective devices.

Head injuries are usually grouped into closed head injury (CHI) and open head injury for study purpose. Open head injury refers to head injuries characterized by skull damage. While closed head injury (CHI) refers to traumatic brain injuries without any skull fracture or loss of continuity of mucous membranes. Open head injuries can be effectively prevented by helmets. The outer hard shell of a helmet can prevent a sharp object from penetrating the skull by mitigating the impact force. However, how to protect people from CHI is still a topic of lots of on-going research, since mechanism of CHI has not been fully revealed due to its greater complexity.

The main objective of this paper is to investigate three factors that have effects on closed head injuries. The three investigated factors are: cranial elasticity modulus, contact area of impact, and impact duration. Finite element method was employed in the investigations. The maximum peak values of mechanical responses are used to measure the effects of the investigated factors. The objective of the study is to provide some insight into improvement of helmet design.

2 Governing Equations and Finite Element Model

The investigations are based on a two-dimensional finite element model. The model represents a horizontal cross section of a human head as shown in Fig.1 (a). The two-dimensional finite element model is constructed using the magnetic resonance image (MRI) as shown in Fig.1 (b).



(a) (b)

Figure 1 (a) Anatomical structure of the head; (b) Magnetic resonance image

Human brain has extraordinarily complex geometry and material properties [4]. To simplify the problem, the following simplifications and assumptions are introduced:

- Rigid motions of the head are constrained in the simulations. Displacements and strains are assumed small.
- No skull fracture occurs during impact.
- The two-dimensional head model is roughly segmented into three parts as shown in Fig. 2(b): the skull, cerebrospinal fluid (CSF), and the brain. Skull and brain are assumed elastic or viscoelastic, isotropic and homogeneous. CSF is considered as an inviscid fluid.
- Convective effect of CSF is omitted; deviatoric stresses induced by viscous effects are neglected.

Under the above simplifications and assumptions, the following equations governing the three parts and their interactions are established.

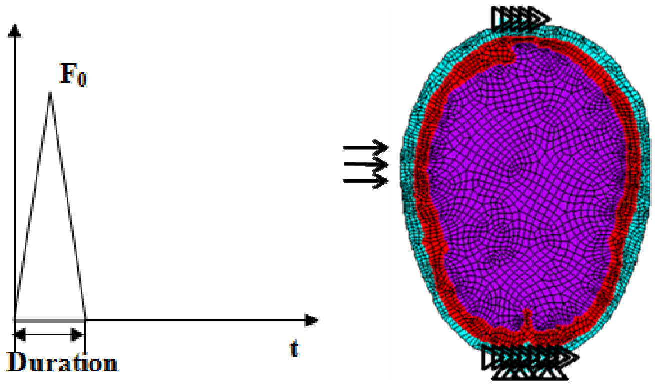
2.1 Governing Equations

The skull is considered as an elastic solid. Body force is not considered. The equation of motion is given by [5]:

$$\nabla \cdot \sigma = \rho_s \ddot{u} \tag{1}$$

where σ is the stress tensor; u is the displacement vector; ρ_s is the mass density of the skull.

The brain tissue is considered as a viscoelastic material. Its motion is governed by the following equation [5],



(a) (b)

Figure 2 (a) Triangular pulse; (b) Model segmentation and finite element mesh

$$\nabla \cdot \sigma + \mu \dot{u} = \rho_s \ddot{u} \tag{2}$$

where μ is a set of viscosity parameters describing the damping property of brain tissue.

CSF is governed by the following wave equation [5]

$$\frac{1}{c^2}\frac{\partial^2 p}{\partial t^2} - \nabla^2 p = 0 \tag{3}$$

where $c=\sqrt{k/\rho_0}$ is sound speed in the fluid; k is the bulk modulus of CSF; ρ_0 is the fluid density; p is the acoustic pressure; and t is the time variable.

Interaction between CSF and the solid parts is described by [6]

$$n^T \cdot \nabla p = -n^T \cdot (\rho_0 \ddot{u}) \tag{4}$$

where n is the normal of solid-fluid interface pointing to the fluid region.

2.2 Finite Element Formulations

By following the standard Galerkin procedure [6], the governing equations (1) - (4) can be transformed into the following finite element equations.

$$\begin{bmatrix} M^S & 0 \\ M^{fS} & M^f \end{bmatrix}\begin{Bmatrix} \ddot{\tilde{u}} \\ \ddot{\tilde{p}} \end{Bmatrix} + \begin{bmatrix} C^S & 0 \\ 0 & C^f \end{bmatrix}\begin{Bmatrix} \dot{\tilde{u}} \\ \dot{\tilde{p}} \end{Bmatrix} + \begin{bmatrix} K^S & K^{Sf} \\ 0 & K^f \end{bmatrix}\begin{Bmatrix} \tilde{u} \\ \tilde{p} \end{Bmatrix} = \begin{Bmatrix} 0 \\ 0 \end{Bmatrix} \tag{5}$$

where

$M^s = \int_{\Omega_s} \rho_s N_u^T N_u d\Omega$, $M^f = \frac{1}{c^2}\int_{\Omega_s} \rho_s N_u^T N_u d\Omega$,

$M^{fs} = \rho_0 \int_s \rho_s N_p^T n^T N_u dS$, $C^s = \int_{\Omega_S} N_u^T u \; N_u d\Omega$,

$C^f = \frac{1}{c}\int_s N_p^T u \; N_p dS$, $K^s = \int_{\Omega_S} B_u^T D B_s d\Omega$

$$K^f = \int_{\Omega_f} B_p^T B_p d\Omega, \qquad K^{sf} = -\int_S N_u^T n N_p dS.$$

In the above expressions, N_u and N_p represent element shape functions for the displacements and for the pressure field respectively; B_u and B_p are the B-matrices obtained by operating the corresponding shape functions with the appropriate differential operator. D is the material property matrix for the solid parts. Ω_s and Ω_f are the sub-domains occupied by the solids and the fluid, respectively. S is the interface surface between the solids and the fluid.

Material properties of the components are obtained from references [7, 8, 9] and listed in Table 1. In all numerical simulations, the same boundary conditions shown in Fig.2 (b) are applied to constrain rigid body modes. The triangular pulse in Fig.2 (a) is applied for simulating impact causing brain injury.

3 Simulations and Results

Table 1: Material properties of each part in human head

	Young's Modulus (MPa)	Bulk Modulus (MPa)	Poisson's Ratio	Mass Density (kg/m^3)
Skull	6650	-	0.220	2080
Brain	0.5581	2190	0.485	1040
CSF	0.1485	2190	0.499	1040

The main purpose of the following simulations is to find how the maximum peak stresses in the skull and the maximum peak strains in the brain are affected by cranium elastic modulus, contact area of impact and impact duration.

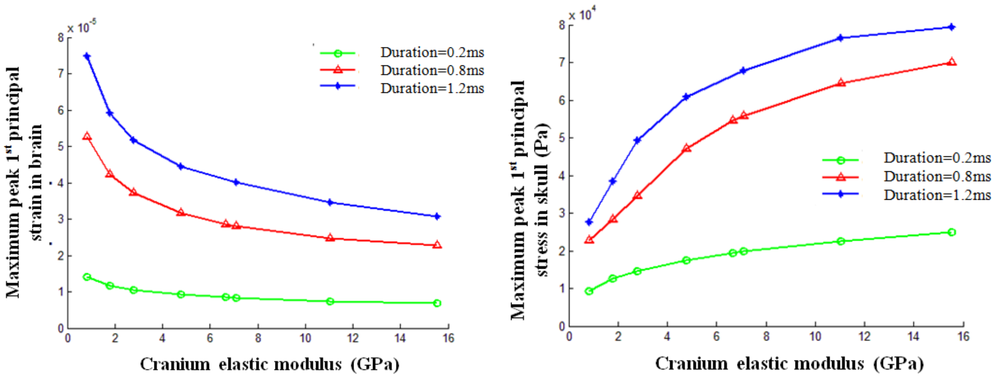
3.1 Effects of Cranium Elastic Modulus

Cranium elastic modulus changes significantly with respect to the age. For a one week old infant, the elastic modulus of cranial bone is about 820.9MPa [10]. For an adult it can be as large as 15.54GPa [11]. Therefore, we selected a number of cranium elastic moduli from the range of 820.9MPa to 5.54GPa. Impacts with different durations were applied. The obtained simulation results are plotted in Fig.3. In Fig.3 (a) the maximum peak first-principal strains in the brain are displayed. In Fig.3 (b) the maximum peak first-principal stresses in skull are shown. The following observations can be made from the obtained results:

- The maximum peak first-principal strains in the brain decrease significantly with increases in cranium elastic modulus. For all of the impact durations, the maximum peak strain is decreased by approximately 50% with cranium elastic modules increased from 820.9Mpa to 15.54Gpa. At the same time, the maximum peak first-principal stresses is increased.
- When the stiffness of skull reaches a certain level, increases in cranium elastic modulus do not show any further effect on the maximum peak strains in the brain. The increases of the maximum peak first-principal stresses in the skull also become less obvious.

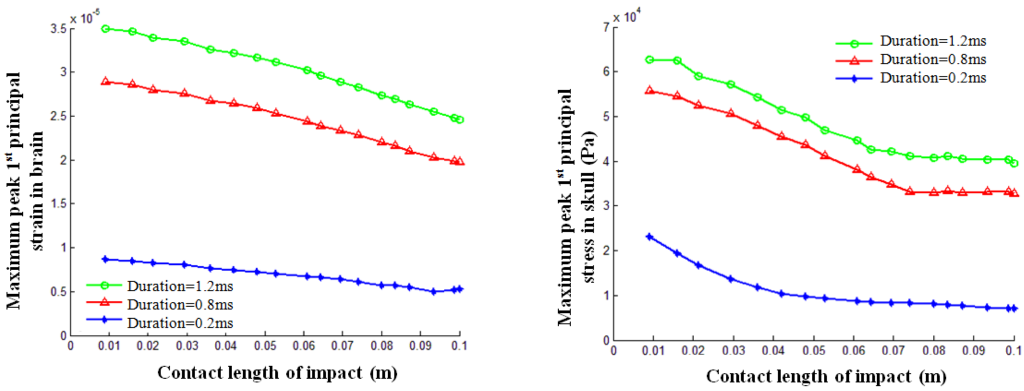
3.2 Effects of Contact Area of Impact

The second set of simulations was performed to study the effects of contact area of impact. The resultant of the distributed force was kept constant. The contact length (contact area) of the distributed force on the skull was changed from 0.9cm to 10cm. Similar as in the previous set of simulations, three impacts with different durations of 1ms, 0.8ms, and 0.2ms were applied, respectively. Simulation results are plotted in Fig.4. The three curves have similar trends. The maximum peak strain decreases by about 25% when the contact length of impact is increased from a small value (0.9cm) to almost half of the head height (10cm). But after that, the maximum peak strain does not have any obvious decrease even the contact length is increased. Increasing contact length of impact also helps reduce stresses in the skull. The reduction in the maximum peak stress is more considerable for shorter impact durations. The maximum peak stress in skull is reduced by 35%-50% with contact length increased from 0.9cm to 7cm. When the contact length is larger than 7cm (less than half of the head height), the maximum peak stress stops dropping even the contact length is increased.



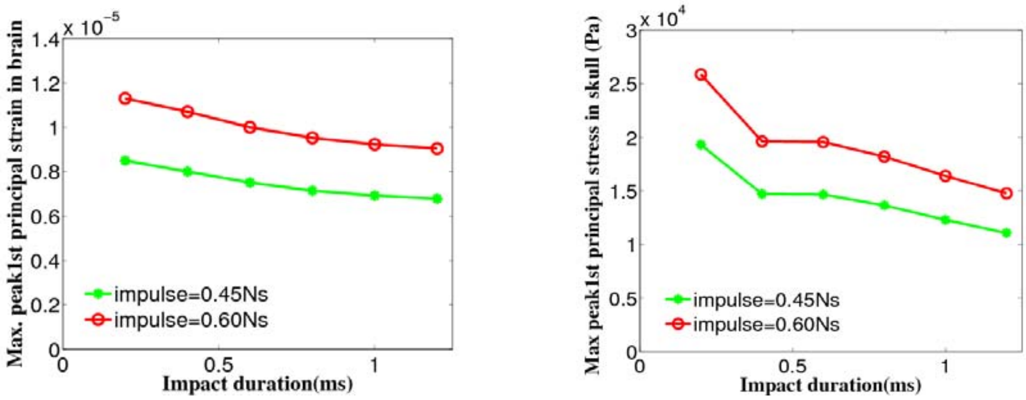
(a) Strains in the brain (b) Stresses in the skull

Figure 3 Effects of cranium elastic modulus



(a) Strains in the brain (b) Stresses in the skull

Figure 4 Effects of contact area of impact



(a) Strains in the brain (b) Stresses in the skull

Figure 5 Effects of impact duration

3.3 Effects of Impact Duration

Two simulations were performed to investigate the effects of impact duration on closed head injuries. In each simulation, the total input impulse was kept unchanged, i.e., the area of the triangle pulse shown in Fig.3 (b) was kept as a constant. The obtained investigation results are plotted in Fig.5. The results show that:

- The maximum peak first-principal strain in the brain is reduced by approximately 20% with impact duration increased from 0.2ms to 1ms. After that, the change becomes flat.
- Impact duration also has a significant effect on the maximum peak first-principal stress in the skull. The stress is reduced by around 50% with impact duration increased from 0.2ms to 1.0ms. The trend indicates that further decrease in the stress is still possible.

4 Conclusions and Discussions

The maximum peak first-principal stress in the skull is an important indicator for skull fracture. The maximum peak first-principal strain in the brain is directly related to closed head injuries.

Increase of cranium modulus, contact area of impact and impact duration will lead to reduction of the strains in the brain. Hence, protective devices that can increase cranium elastic modulus, distribute the impact loads to a larger area, or increase impact durations can buffer the impact, and help to protect the brain during impact.

Simulation results also indicate that, increasing cranium elastic modulus is the most effective way to reduce the maximum peak strains in the brain among the three measures. The reduction of maximum peak strain in the brain is more considerable for the range of 1.0GPa to 6.0GPa, meaning that wearing a protective helmet is more crucial for a child. The average elastic modulus of a child's cranium is below 6GPa. Wearing a protective helmet has the equivalent effect of increasing the elastic modulus of the protective layers of the brain.

Increasing impact duration and contact area of impact is also helpful to reduce the strain values in the brain. At the same time, they are effective to reduce the stresses in the skull. The above observations may help design more effective protective helmets by selecting a more effective padding material. The contact area and the impact duration can be changed by a helmet using different shell stiffness and different padding material.

Acknowledgement. The authors are grateful to Manitoba Medical Service Foundation (MMSF) for the awarded grant.

References

[1] Brain Damage. Coup and counter-coup injury. Available from: http://www.braindamageattorneys.com/topics.cfm/coup-contrecoup-injury.

[2] Y. Luo, Q. Zhang and M. Del Bigio. Recent progress in application of FEM in study of non-penetrating brain injuries. *Adv. Theor. Appl. Mech.*, 1(5):225 – 240, 2008.

[3] Mau-Roung Lin and Jess F. Kraus. A review of risk factors and patterns of motorcycle injuries. *Accident analysis and prevention*, 41:710 – 722, 2009.

[4] Mulabagula R. Mehta, B. and J. Patel. Finite element analysis of the human skull considering the brain and bone material properties. In Proceedings of *Computer Methods in Biomechanics and Biomedical Engineering*, pages 217 – 227, 1996.

[5] J. Davis. *Wave propagation in solids and fluids*. Springer-Verlag, 1988.

[6] O. C. Zienkiewicz and J. Z. Zhu. *The Finite Element Method: Its Basis and Fundamentals*. Elsevier Butterworth-Heinemann, 2005.

[7] G. Dassios, M.K. Kiriakopoulos, and V. Kostopoulos. On the sensitivity of the vibrational response of the human head. *Computational Mechanics*, 21:382 – 388, 1998.

[8] M. Gilchrist and D O'Donoghue. Simulation of the development of frontal head impact injury. *Computational Mechanics*, 26:229 – 235, 2000.

[9] Lee H. Gong, S. and C. Lu. Computational simulation of the human head response to non-contact impact. *Computers and Structures*, 86:758 – 770, 2008.

[10] Kirk L.Thibault Susan S.Margulies. Infant skull and suture properties: Measurements and implications for mechanisms of paediatric brain injury. *Journal of Biomechanics Engineering*, 122:364 – 371,2000.

[11] G. V. Perre, J. Vander Sloten Julie A.Motherway, P. Verschueren and M. D. Gilchrist. The mechanical properties of cranial bone: The effect of loading rate and cranial sampling position. *Journal of Biomechanics*, 42:2129–2135, 2009.

Received: March, 2010