Analysis of Traumatic Brain Injury Using a Finite Element Model

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In this study, head injury by impact force was evaluated by numerical analysis with 3-dimensional finite element (FE) model. Brain deformation by frontal head impact was analyzed to evaluate traumatic brain injury (TBI). The variations of head acceleration and intra-cranial pressure (ICP) during the impact were analyzed. Relative displacement between the skull and the brain due to head impact was investigated from this simulation. In addition, pathological severity was evaluated according to head injury criterion (HIC) from simulation with FE model. The analytic result of brain damage was accorded with that of the cadaver test performed by Nahum et al. (1977) and many medical reports. The main emphasis of this study is that our FE model was valid to simulate the traumatic brain injury by head impact and the variation of the HIC value was evaluated according to various impact conditions using the FE model.

Key Words : Traumatic Brain Injury (TBI), Finite Element Model, Head Injury Criterion (HIC), Brain Damage, Skull & Brain

Nomenclature

[M] : Mass matrix
[C] : Damping matrix
[K] : Stiffness matrix
G : Shear stress relaxation modulus
G_0 : Short term shear modulus
G_∞ : Long term shear modulus
β : Decay factor

1. Introduction

Head injury is very common damage in modern life due to car crashes, intense sports and so on. It accounts for 80% of the causes of death in car crashes (Luchter and Walz, 1996). Especially, the head injury causes tensile rupture of the blood vessels around the brain by brain deformation and it results in serious aftereffect despite light injury of the brain (Lee and Haut, 1989). Therefore, it is very important to evaluate head injury by impact force and many studies have been carried out to evaluate head injury due to head impact using physical models, cadavers and animals (Gennarelli et al., 1971 ; Margulies et al., 1990). And currently, numerical methods were introduced to evaluate head injury because of the limitation of physical tests.

Holbourn (1943) showed that rotational acceleration causes more serious injury rather than linear acceleration and the relative motion of the

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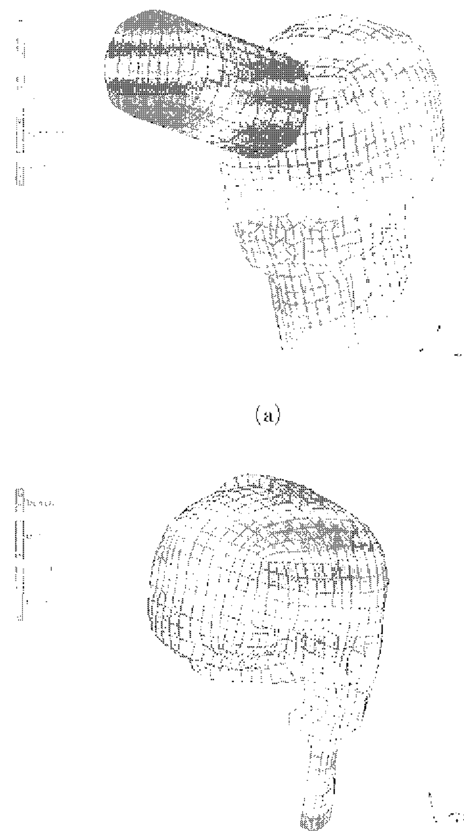
brain results in serious damage of brain tissue and vessels. Miller et al.(1998) performed FE analysis for Diffuse Axonal Injury (DAI) of pig's brain. And they showed that severe DAI is caused by rotational acceleration, and mechanism of DAI and hemorrhage contusion were analyzed by von Mises stress and maximum principal strain distribution. Also, Kleiven et al.(2002) carried out FE analysis, including the head and the neck to investigate the differences of injury mechanism of various sizes of head. And Head Injury Criterion (HIC) showed different values according to variation of head size for the same impact condition.

Therefore, in this study, finite element method was introduced to evaluate head injury by impact force. Especially, in order to evaluate pathological severity of TBI and brain hemorrhage due to rupture of blood vessels on a brain surface, brain deformation was investigated and HIC was applied to the FE model.

The FE model for simulation was constructed from MRI image of a subject, and the same boundary conditions with Nahum's cadaver test were adopted to verify this model. As well, visco-elastic property was applied to brain tissue for increasing analytical accuracy. In addition, the HIC was applied to the FE model so as to evaluate damaged level of the brain in the simulation. And, the HIC value was calculated according to various impact conditions and the variation of the HIC value was also investigated.

2. Modeling Procedure

A finite element model was constructed from MRI image of a male subject. The MRI image was transformed to CAD model through image processing (Scion Image Ver. 4.02, Scion Co.). And FE model was constructed from CAD model using MSC/MENTAT (MSC Software Co.) and FE analysis was performed by MSC/MARC (MSC Software Co.). Figure 1(a) shows a view of the head model with the steel impactor and Fig. 1(b) shows a cross-sectional view of the skull with the brain.



(a)

(b)

Fig. 1 View of a three-dimensional head model for the FE analysis (a) View of the head model with the steel impactor (b) Cross-sectional view of the skull with the brain

A dynamic equivalent equation is required in FEM modeling of the impact analysis for this study. Generally, the dynamic equivalent equation at a time, t, for the FE analysis is as follows.

$$[M]\{\ddot{u}(t)\}+[C]\{\dot{u}(t)\}[K]\{u(t)\}=\{R(t)\} \quad (1)$$

where $[M]$ is a mass matrix, $[C]$ is a damping matrix, and $[K]$ is a stiffness matrix, respectively. And $\{\ddot{u}(t)\}$, $\{\dot{u}(t)\}$, and $\{u(t)\}$ represent an acceleration vector, a velocity vector, and a displacement vector at time t, respectively. The Newmark β-method as direct integration method was applied to the numerical solution to the dynamic equivalent Eq. (1) (Suh et al., 2001).

Table 1 shows material properties of steel im-

Table 1 Material properties of the biological tissues and the steel impactor (Shuck et al., 1972)

Tissues	E (kPa)	ν	ρ (kg/m^3)
Skull	6×10^6	0.21	2100
CSF	1.2×10^1	0.485	1040
Brain	6.75×10^2	0.499	1000
Impactor	2×10^8	0.27	7890

pactor and biological tissues modeled in this simulation (Shuck et al., 1972). Cerebro-spinal fluid (CSF) was modeled as incompressible material instead of liquid, and brain was modeled as linear visco-elastic material.

The following equation represents the shear stress relaxation modulus, $G(t)$ for linear visco-elastic behavior in shear.

$$G(t)=G_\infty+(G_0-G_\infty)e^{-\beta t} \qquad (2)$$

where G_0 is a short term shear modulus, G_∞ is a long term shear moduls, and β is a decay factor. This study used experimental properties as $G_0=528$ kPa, $G_\infty=168$ kPa, and $\beta=0.035$ s^{-1}, respectively, determined by Shuck et al. (1972) in their experiment.

As for boundary conditions, the free-boundary condition was applied to this simulation because the neck constraint doesn't effect head responses to the short duration of such force (Willinger et al., 1999). And as for impact conditions, initial velocity of 9.94 m/s was applied to the steel impactor of 5.6 kg mass, which is the same condition as in Nahum et al. (1977)'s cadaver test. The impact position was set on the frontal bone, the same as in the cadaver test.

3. Results and Discussion

3.1 Verification of the FE model

After the simulation with the FE model, some parameters were compared with the experimental results recorded from Nahum et al.'s cadaver test in order to validate this simulation : impact force, head acceleration, ICP, and so on.

Figure 2 shows a variation of the impact force which is transmitted from the impactor to the head, and the impact force of the simulation

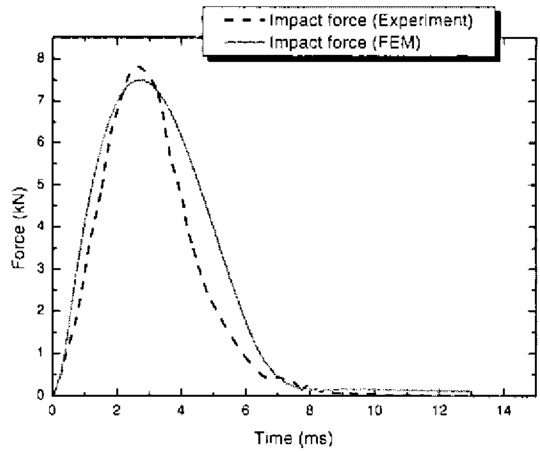


Fig. 2 Variation of the impact force during the impact

shows good agreement with the experimental data in the cadaver test as shown in Fig. 2. But the maximum value of the impact force of the FE analysis was slightly lower than that of the experiment. It is supposed that it was produced by the numerical errors in the simulation.

Especially, the experimental impact force converged to zero force but the FE analysis results converged to 0.1 kN. It is due to the difference of the boundary condition for constraints by the neck since free boundary condition was applied to the simulation. In other words, the neck constraints had an effect on the force variation after 7 ms. This shows a similar result to Willinger et al. (1999)'s simulation that the neck constraints influence the kinematic response of the head after 6 ms.

Figure 3 shows a variation of the head acceleration during the impact, which was measured on the centroid of the temporal bone of the skull. Even though qualitative trend corresponded with each other, the maximum value of the acceleration was lower than that of the experiment. And there were some fluctuations in the experimental result after the peak of the acceleration but the FE analysis result showed single fluctuation at 7 ms after the peak.

It is supposed that it was originated from the simplification of the material property and the geometry in formulating the dynamic characteristic for the simulation. However, since the quantitative difference in the acceleration variation was trivial, it is appropriate for the analysis of the brain damage with the HIC. Besides, although the converged values of the acceleration after 7 ms are different from each other in the experiment and the simulation, which is caused by the effect of the difference of the neck constraints after 6 ms, it doesn't influence the calculation of the HIC value because the acceleration value after 7 ms is much lower than the peak of the acceleration variation.

Figure 4 represents a variation of the frontal

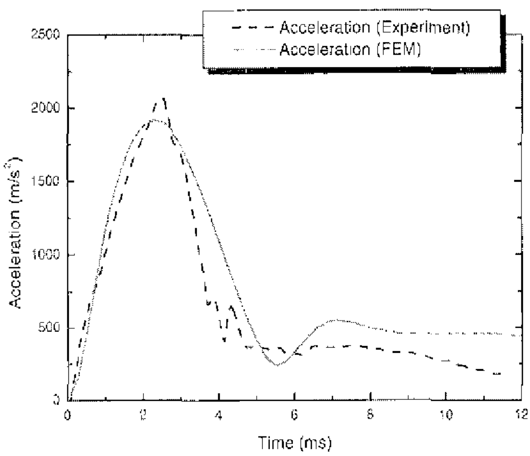


Fig. 3 Variation of the head acceleration during the impact

pressure during the impact. As shown in Fig. 4, positive pressure was produced on the frontal part by the direct contact force on the frontal bone from the impactor and it diminished gradually in both of the experiment and the simulation. However, although negative pressure occurred at 9 ms in the experiment for a while and it converged to zero pressure, the data from the FE analysis converged to positive pressure of 15 MPa after an approach to zero pressure.

Figure 5 shows a variation of the upper occipital pressure during the impact. It shows a good agreement in the qualitative variation in the occipital pressure causing contre-coup injury by the frontal impact. Namely, negative pressure was produced initially and then it was changed into positive pressure in both of the experiment and the simulation. It means that the cavitation occurs in the occipital lobe and causes severe damage to brain tissue on the surface of occipital lobe after a dull peak.

Accordingly, as comparing the result of the FEM simulation, it accorded with the experimental data obtained from Nahum et al.'s cadaver test even though there were differences in the maximum values. Consequently, this model is appropriate to analyze the mechanism of brain vessel rupture and brain injury although there is a little difference in the quantitative value, considering the differences of the geometry, the mec-

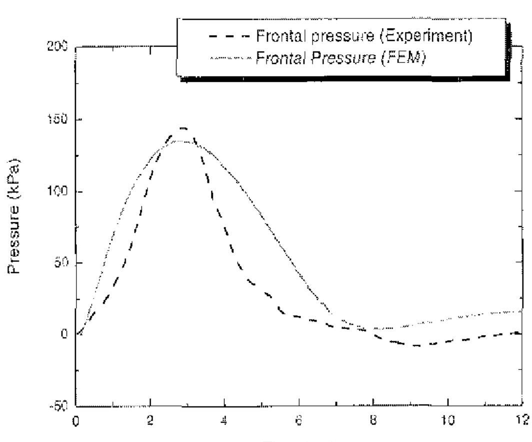


Fig. 4 Variation of the frontal pressure during the impact

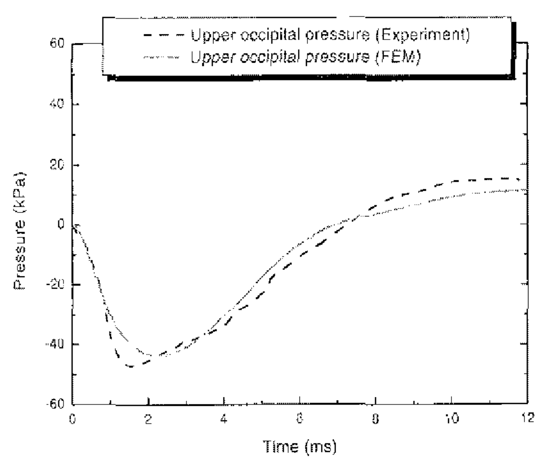


Fig. 5 Variation of the upper occipital pressure during the impact

hanical property, the numerical error and so on.

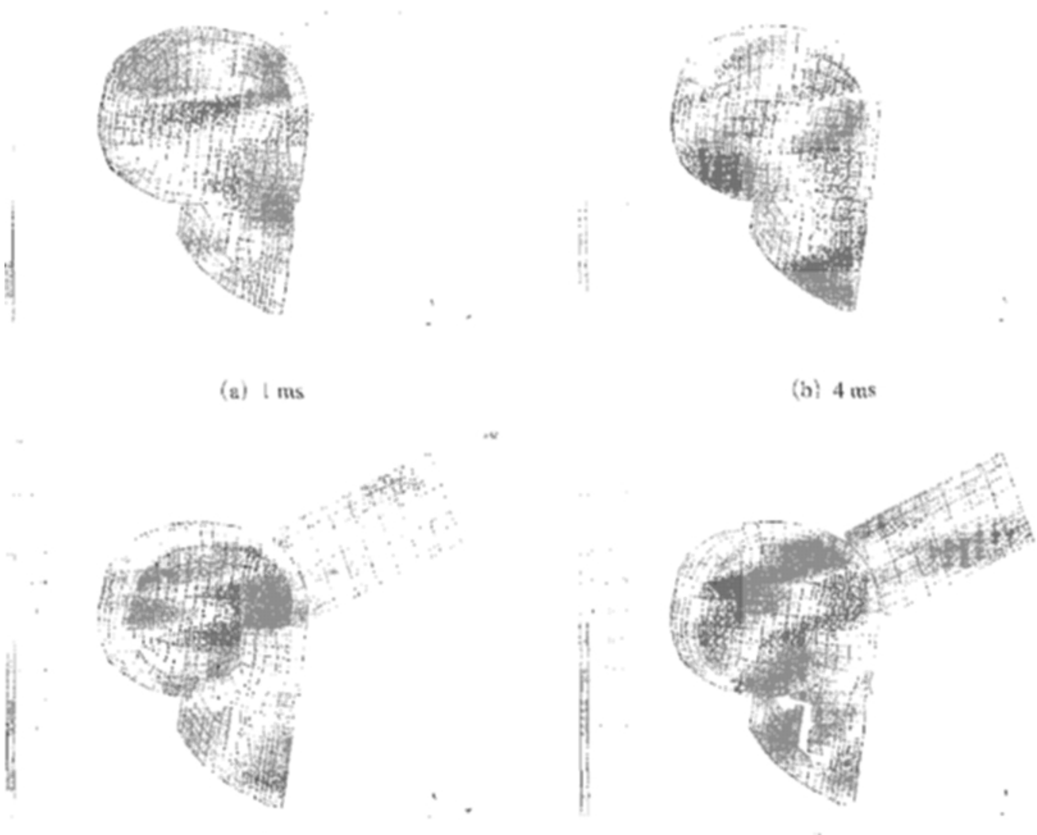
Figure 6 shows a variation of the displacement distribution during the impact as a result of the simulation. The skull deformed initially as soon as the impactor contacts with the frontal bone, and then it moved to the impacted direction during the impact. Even though the whole head moved to the impacted direction because of the free boundary condition, there was a difference in the displacement between the brain and the skull by the inertia and the mechanical property of the brain and CSF. The maximum relative displacement between the skull and the brain was 3 mm at 6 ms.

Therefore, this result corresponded with the medical reports and the experimental researches (Holbourn, 1943; Al-Bsharat et al., 1999). This is the most dangerous factor to produce the serious brain injury at the head impact because it causes the rupture of the bridging veins distributed in sagittal sinus owing to the excessive tension.

3.2 Application of HIC

The HIC value of the FEM simulation was calculated by FORTRAN program (Visual FORTRAN Ver. 6.0, DIGITAL Equipment Co.) which was coded according to Eq. (3).

In order to verify the HIC value calculated from the FEM simulation, the experimental data obtained from Nahum et al.'s cadaver test was



(a) 1 ms (b) 4 ms (c) 8 ms (d) 10 ms

Fig. 6 Displacement distribution during the impact compared with the HIC value in this simulation.

$$HIC = Max(t_1, t_2)\left\{(t_2-t_1)\left[\frac{1}{(t_2-t_1)}\int_{t_1}^{t_2} a(t)\,dt\right]^{2.5}\right\} \quad (3)$$

where t_1 and t_2 are two ends of time interval of the duration that makes the maximum value of HIC.

Meanwhile, although there is a difference in anatomical geometry between the cadaver in the experiment and the FE model in the simulation, qualitative behavior of the mechanical response of the head and the brain were similar with each other ; head acceleration, ICP, impact force, and so on. Moreover, the results of the simulation accorded well with medical reports about brain injury and rupture of brain blood vessels. Consequently, the FE model in the simulation is valid to simulate the traumatic brain injury by head impact. Thus, it is reasonable enough to determine the HIC value and to analyze the variation of the HIC value according to the impact condition.

Impact condition and the result of the cadaver test performed by Nahum et al.(1977) are shown in Table 2. And those of the simulation performed in this study are shown in Table 3. As represented in Tables 2 and 3, although impact condition was the same in the experiment and the simulation, the results show a little difference in the acceleration and the impact force owing to the difference in the geometry for simplification of the FE model.

Table 2 Impact condition and result of the cadaver test (Nahum et al., 1977)

Impactor mass (kg)	Impact velocity (m/s)	Peak of impact force (kN)	Peak of acceleration (m/s^2)	HIC
5.59	9.94	7.90	2000	744

Table 3 Impact condition and result of the simulation

Impactor mass (kg)	Impact velocity (m/s)	Peak of impact force (kN)	Peak of acceleration (m/s^2)	HIC
5.59	9.94	7.51	1912	715

Therefore, the HIC value calculated from the simulation, 715, was lower than that of the experiment, 744. However, the difference of the HIC value between the experiment and the simulation was just 4%, so it seems to be a reasonable value as considering other mechanical responses of head acceleration and impact force. Because the HIC value didn't exceed the value of 1,000 in Eq. (3), this impact condition is not so dangerous to cause death or severe head injury by frontal head impact in general case. In the simulation, the impact velocity and the mass of the impactor were varied in order to investigate the variation of the HIC value for the same head model.

Figure 7 represents a variation of the HIC value according to the impactor mass. In order to maintain the geometry of the impactor in the impact condition, the density of the impactor was changed and adjusted for changing the impactor mass, instead of the volume of the impactor. As shown in Fig. 7, the HIC value increased as the impactor mass increased. It increased gradually in lower mass less than 4 kg, and then increased rapidly in higher mass. The HIC value exceeded 1,000 at the impact with 7 kg mass and reached about 2,130 with 10 kg. Also, this result represents a similar trend to the FEM simulation performed by Kleiven et al.(2002), who reported that the HIC value decreased with increasing the head size by investigating the variation of the HIC value according to the head size for the same impact condition.

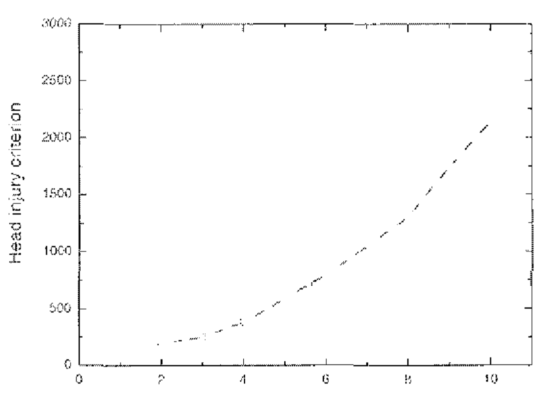


Fig. 7 Variation of the HIC value according to the impactor mass

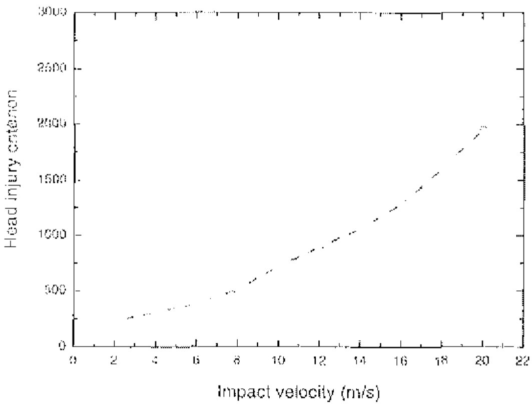


Fig. 8 Variation of the HIC value according to the impact velocity

Figure 8 shows a variation of the HIC value according to the impact velocity. As represented in Fig. 8, the HIC value also increased as the impact velocity increased, similar to the relationship with the impactor mass. Although the increasing rate of the HIC value was slightly rapid in the higher velocity and mass as a concave curve in Figs. 7 and 8, it showed almost linear relationship according to the impact velocity and the impactor mass. Thus, the simulation and the FE model of this study would offer important data to estimate the traumatic brain injury by the frontal head impact.

4. Conclusion

In this study, the FE analysis with a three-dimensional FE model was performed in order to analyze mechanism of brain injury. As the results of the FE analysis, the mechanical behavior from the simulation agreed with the experimental study performed in Nahum et al.'s cadaver test and the medical reports.

This simulation was validated from the experimental data by analyzing physical phenomenon about brain injury during the impact such as the coup, the contre-coup, and the relative displacement between the skull and the brain. Therefore, the FEM simulation with this model would be an effective method to evaluate the brain injury according to various impact conditions, using the HIC value.

Acknowledgments

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