Development and Application of Stress-Based Skull Fracture Criteria Using a Head Finite Element Model*

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Abstract
Skull fractures occurring under various impacts are required to be evaluated during forensic investigations or during design of a head protection device. Finite element analysis using a human head model is an effective evaluation tool for it, because investigating of the behavior of skull fracture such as the start and the progress can be easily performed. JARI Head Tolerance Curve (JHTC), which is an existing criterion for skull fracture based on the cadaver head drop tests, uses head acceleration as the parameter, though the fractures are mainly caused by stress concentration on local area. In this study, we tried to develop stress-based criteria for skull fracture, and to simulate skull fracture by using a head finite element model, which we previously developed. Forehead dropping simulations with five different conditions were performed, and the results were found to be nearly equivalent to the JHTC tests in terms of effective acceleration of head and its duration time. We then determined critical stresses for skull fracture as a function of duration time, based on the time histories of stresses on the skull surface in the five simulation results. Furthermore, we applied our criteria to two forehead impact simulations. We could observe fracture on the skull followed by sharply decreasing in stress on fractured elements as well as increasing in stress on its neighboring elements, and also variation in shapes and ranges of fractured element distribution. These results indicate that the criteria could evaluate skull fracture in detail under different patterns of impact.

Key words: Skull Fracture, Impact, Finite Element Analysis, Human Head Model, Injury Criteria, Stress, Duration Time

1. Introduction

Head injuries can occur under various impact patterns. The patterns can generally be categorized into two groups. One is indirect and the other is direct. The former imposes accelerations on the head and causes brain injuries such as concussion or contusion, due to relative motion between the skull and the brain. This type of impact notably happens in the automotive accidents and leads to fatal injuries (1). On the other hand, under a direct impact, the head is subjected to a sudden load acting on a local area, which causes skull fracture (2). Though the injury may not be so fatal compared to brain injury, it commonly occurs in
various accidents or crimes, and can become permanent. Therefore, in forensic investigations or in designing head protection devices, skull fractures are necessary to be evaluated in a mechanical way. In order to analyze head injury due to direct impact, experiments and computer simulations are being actively conducted in recent years. However, experiments with cadavers or animals are difficult to perform because of the high costs and ethical problems. Meanwhile, finite element analysis that enables to simulate local deformation is now attracting attention as a tool to analyze the internal dynamic response of human head during impact and the mechanisms of head injuries. Whatever the method, to evaluate brain injury and skull fracture, appropriate injury criteria are required.

Head Injury Criteria (HIC) \(^{(3)}\), an index based on Wayne State Tolerance Curve (WSTC) \(^{(4)}\) is an acceleration-based criterion and widely used to predict brain injury during impact. However, skull fracture that occurs due to concentration of load on local area is difficult to predict by an acceleration-based criterion. Therefore, skull fracture should be evaluated by a stress or strain-based criterion. From the fracture mechanics point of view, several fracture models can be adopted, and the parameter value of the fracture model can be determined by tests using human or animal specimens. However, human head has complicated structures, geometries and characteristics, resulting in the difficulty of determining generalized characteristics for all structures. Therefore, we considered an alternative way to easily develop stress-based skull fracture criteria by considering external force or energy acting on whole head during impact. Ono et al. \(^{(5)}\) focused on head acceleration as an injury parameter as WSTC suggested, and performed dozens of impact tests with human or ape cadaver heads. They measured head acceleration and investigated injury occurrences. From the results, JARI Head Tolerance Curve (JHTC) was suggested as an acceleration-based criterion, which distinguishes skull fracture from brain injury. Therefore, by replicating the tests in JHTC with a numerical head model, the stress distribution on the skull when fracture occurred, which depends on the model, could be determined.

In this study, we intended to show a method to develop new skull fracture criteria expressed by the relations between stress and its duration time based on JHTC, which determines the relations between effective stress of head’s center of gravity and its duration time as skull fracture parameters.

The flow of this study is illustrated in Fig. 1. Firstly, we validated elastic response of the finite element head model which we previously developed \(^{(6,7)}\) by replicating forehead and vertex direct impact tests. Secondly, we developed stress-based skull fracture criteria by undertaking forehead dropping simulations to replicate the head drop tests used in developing JHTC. In the first stage of this simulation, the rigid head model which has the same geometry and mass as the original deformable one was used to determine impact conditions where head dynamic response corresponded to JHTC in regard to acceleration of its center of gravity. In the next stage, from the simulations with the deformable head model under the conditions, the time histories of stresses on the skull at the impacted area were obtained. By analyzing the results, we proposed new stress-based skull fracture criteria. Furthermore, we introduced the criteria into the finite element analysis and tried to predict skull fractures in two direct impact situations. Thus, this study showed that the use of head finite element model can provide us stress-based skull fracture criteria which can improve the possibility of predicting and reproducing skull fractures due to direct impact.

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**Fig. 1 Flow chart of this study**

- **Validation of head FE model**
  - Numerical replications of forehead impact tests
  - Numerical replications of vertex impact tests
- **Development of stress-based skull fracture criteria**
  - Numerical replications of head dynamic response in JHTC
  - Numerical replications of stress distribution on skull in JHTC
  - Correlations study between stress and its duration time
- **Implementation of criteria**
  - Numerical simulations of forehead golf ball impact
  - Numerical simulations of forehead plate impact

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2. Head Finite Element Model and Its Validation against Impact on Skull

A finite element model of the human head was used to simulate the head drop tests used in developing JHTC \(^5\) and to simulate two direct impacts including a drop to an elastic plate and a hit by a golf ball. We omitted the mandible part which has no significant effect on the fracture response of skull under forehead or vertex impacts. The head finite element model had been validated in our previous studies regarding brain dynamic responses \(^6\)(\(^7\)), however validation regarding skull dynamic response has not been conducted yet. In this study, our objective is to develop skull fracture criteria based on stress as mentioned in Chapter 1. However, it is difficult to measure stress in the experiment with human skull. Therefore, validation of the head model during collision in terms of contact force response instead of stress response was considered to be reasonable. Therefore, we validated the modified model against the two cadaver head impact tests \(^8\)(\(^9\)).

The head model shown in Fig. 2 had a faithful geometry of an average Japanese male based on CT images of a Japanese adult male. The model consists of 14 parts including scalp, skull (outer-table, diploe, and inner-table), brain (left/right cerebrum, cerebellum, and brainstem), meninges (dura mater, falx, tentorium, and pia mater), cerebrospinal fluid, and mandible. The model without the mandible part had 15047 nodes, 22345 elements, and weight of 3.36 kg. The outer table and the inner table, which are the outer and inner layers of the skull, is cortical bones that are hard and dense, whereas the middle layer of the skull is spongy called as cancellous bone. Each table consists of a single layer of quadrangular shell elements with uniformly 1.7 mm of thickness which is taken from a literature \(^10\). The thickness could not be replicated based on the CT images because its sampling interval was 2 mm, which is not sufficiently-narrow. The diploe consists of two layers of hexahedral solid elements, whose geometry is based on the CT images. The thickness of this part ranges between about 3 to 8 mm, which represents that the thickness of the skull varies by site. These three layers were elastic, and their material properties were taken from a literature \(^11\). Density \((\rho)\), Young’s modulus \((E)\), and Poisson ratio \((\nu)\) were 1900 kg/m\(^3\), 12.2 GPa, and 0.21 for cortical bone, 1500 kg/m\(^3\), 1.0 GPa, and 0.05 for cancellous bone respectively. The scalp model was made to cover only the impacted area of the skull surface in order to reduce simulation time.

Yoganandan et al. \(^8\) performed an experiment where a rigid sphere impacted the vertex of human cadaver head. Meanwhile, Allsop et al. \(^9\) conducted experiments where a rigid cylinder impacted the forehead of human cadaver head. In the model validation, we designed two simulations reproducing the two tests as shown in Fig. 3. In this study, LS-DYNA ver. 971 was used as a FEA solver. In the simulation of Yoganandan’s test as shown in Fig. 3(a), the vertex of the head model whose bottom region was fully-restrained, was impacted with a velocity of 7.1 m/s by a rigid hemisphere model with a radius of 92 mm and a weight of 0.875 kg. In the simulation of Allsop’s test as shown in Fig. 3(b), the forehead of the head model whose occipital region was fully-restrained, was impacted with a velocity of 3.0 m/s by a rigid cylinder model with a radius of 20 mm, a length of 140 mm and a weight of 14.5 kg.
In the two simulations, the contact force between the impactor and the head model increased along with the impactor displacement. The results of the simulations generally matched to the results of the two tests in terms of force-displacement curve as presented in Fig. 4. In the vertex impact simulation (Fig. 4(a)), the contact force shown by a solid line increases non-linearly as the displacement becomes larger. The maximum difference from the experimental result shown by a broken line is 20% in force and 10% in displacement. In the forehead impact simulation (Fig. 4(b)), the simulation result is among the variation of the test results. Although skull fracture occurred at the edge of each broken line in the both experiments, replication of it was not required in the validation process since reproducibility of contact force response was thought to be the most important. Thus, the validity of the skull dynamic response of the head model against direct impacts was confirmed according to these comparison results.

3. Development of Stress-Based Skull Fracture Criteria

Critical acceleration for head injuries including brain injury and skull fracture were determined as a function of duration time as shown in Fig. 5, based on impact experiments using a numbers of heads of human cadaver and ape(4)(5). Those curves indicate that a skull fracture happens when an “effective” amount of acceleration continuously acts on the head, and that neither an instantaneous acceleration nor a sum of whole acceleration could evaluate head injuries. JHTC for skull fracture utilized in this study represents a total of 78 head drop tests with 15 human cadaver heads. Though existence of skull fracture in each test was reported, the details including the type of fracture was not described(5). Thus, it is impossible to validate the type of fracture obtained from the simulation. However, at present, it is extremely difficult to conduct head impact tests with such numbers of cadavers and get the details of skull fracture. Therefore, in this study, we mainly focused on the development of stress-based criteria which predicts the occurrence of skull fracture and neglected the type of fracture.
When the head is impacted, compressive and tensile stress could occur on the outer and the inner surface of the skull respectively \(^{(2)}\). Concentrations of these stresses considered to be responsible for causing fracture. It was also presumed that skull fracture occurs when all bone layers are broken according to the skull fracture classifications \(^{(2)}\). In strength of material, principal stress is generally used as failure (yield) criterion. Therefore, we considered that the minimum principal stress and the maximum principal stress should be able to simultaneously evaluate the concentration of compressive and tensile stresses occurred on the outer and inner table respectively.

The process for developing the criteria consisted of three phases as shown in Fig. 1. At the first phase, the head model that we validated in Chapter 2 was converted as a rigid head model. Here, deformable head model cannot be used, because its center of gravity always changes during the analysis, and its acceleration cannot be obtained properly. By using this model, head dynamics in the drop tests during development of JHTC were reproduced in forehead dropping simulations. In these simulations, simulation conditions were adjusted so the relations between the effective acceleration and its duration time of the head model were equivalent to certain conditions of the JHTC. That is, the relations could be plotted on the curve. Secondly, under the conditions, the simulations with the deformable head model were conducted, and then the time histories of principal stress on the skull surface of the head model were obtained. At the final phase, the relations between effective critical stresses and its duration time for fracture occurrence were determined and proposed as the skull fracture criteria. We showed that the existing acceleration-based skull fracture criteria can be converted to stress-based criteria. Moreover, not only the magnitude of stress but also its duration time was an important factor for predicting skull fractures.

![Fig. 5 Wayne State Tolerance Curve (WSTC) \(^{(4)}\) and JARI Head Tolerance Curve (JHTC) \(^{(5)}\)](image)

### 3.1 Analytical Methods

The forehead dropping simulation was designed as shown in Fig. 6, to reproduce the head drop tests used in developing JHTC where the forehead dropped to the floor. The rigid square plate model was 220 mm on a side, weighed 0.05 kg, and consisted of 4-mm-square shell elements with a thickness of 0.01 mm. The four springs with a length of 100 mm defined the elasticity of the rigid plate-spring structure. By varying the pairs of dropping velocity \(V\) and spring constant \(k\), various head dynamic responses in the drop test during development of JHTC could be reproduced.

Five simulation conditions, shown in Table 1, were determined by the forehead dropping simulation with the rigid head model, which consisted of a rigid outer table, an elastic scalp, and a lumped mass of the other head parts placed on the center of gravity of the head model in Chapter 2. Each simulation condition led to different dynamic response of the head model. Under these simulation conditions the relation between the effective head acceleration and its duration time were equivalent to certain conditions on the JHTC as shown in Fig. 7. JHTC and WSTC can be considered as similar criteria since they utilize effective acceleration and its duration time as a predictor for head injury. Therefore, HIC \(^{(3)}\), which was originally developed to approximate the curve of WSTC as indicated in Eq. (1),
was used to calculate the effective head acceleration and its duration time in the simulations. In the five impact conditions, the forehead dropping simulations with the head model were performed, and each stress on the skull were investigated.

Fig. 6 In the forehead dropping simulation, the non-restrained head model moved down vertically with a speed of $V_k$ and hits the rigid square plate model whose four corners were supported by elastic springs with a combined spring constant ($k$).

Table 1 Pairs of $k$ and $V$ in the forehead dropping simulation

<table>
<thead>
<tr>
<th>Condition</th>
<th>$k$ [kN/m]</th>
<th>$V$ [m/s]</th>
</tr>
</thead>
<tbody>
<tr>
<td>#1</td>
<td>40000</td>
<td>3.0</td>
</tr>
<tr>
<td>#2</td>
<td>4000</td>
<td>3.7</td>
</tr>
<tr>
<td>#3</td>
<td>2000</td>
<td>3.8</td>
</tr>
<tr>
<td>#4</td>
<td>1000</td>
<td>4.2</td>
</tr>
<tr>
<td>#5</td>
<td>500</td>
<td>4.5</td>
</tr>
</tbody>
</table>

Fig. 7 Head effective acceleration versus its duration time: comparison of the forehead dropping simulations under the five impact conditions and the JHTC.

\[
HIC = \left[ (t_2 - t_1) \frac{1}{t_2 - t_1} \int_{t_1}^{t_2} a(t)dt \right]^{2.5}_{\max}
\]  

(1)

\[
\Delta_t \sigma_{eff}^{2.5} = \left[ (t_2 - t_1) \frac{1}{t_2 - t_1} \int_{t_1}^{t_2} \sigma(t)dt \right]^{2.5}_{\max}
\]

(2)

In Eq. (1), $t_1$ and $t_2$ are the initial and final times of the time interval which HIC attains a maximum value as shown in Fig. 8(a). The interval of $t_1$ and $t_2$ is duration time of effective acceleration ($\Delta_{t\text{acc}}$), and the average of acceleration during the interval is effective acceleration ($a_{\text{eff}}$). From Eq. (1) and Fig. 8(a), there is an interval where accumulation of the acceleration during this interval significantly affects head injuries (shaded area). In other words, head injuries may not occur when there is a lack of magnitude or duration time of acceleration acting on the head.

Considering the concept of HIC, the stress-based criteria were developed with the same concept. Eq. (2), which is a conversion of HIC equation (Eq.(1)), but with principal stress and its duration time as the parameters, was used to calculate the effective stress ($\sigma_{\text{eff}}$) and
its duration time ($\Delta t_{\sigma}$) from the time histories of principal stresses in the forehead dropping simulations using the head finite element model as shown in Fig. 8(b). As described in the HIC functional by Hutchinson et al. (12), Eq. (2) can also be interpreted as an amount of kinetic energy which is required to cause skull fracture. Time integration of stress (force per unit area) in the equation can be considered as impulse (change in momentum). Thus, if the power in the equation is 2, it will represent the maximum average kinetic energy that can be delivered to the head. However, if the power is changed to 2.5, which means larger than the maximum average kinetic energy that head can afford to receive, the equation will indicate the amount of energy required for skull fracture. Furthermore, as we discussed previously, stress-based skull fracture criteria should be expressed by principal stress. Therefore, effective compressive stress and effective tensile stress for fracture occurrence should be obtained from the minimum principal stress on the outer table and the maximum principal stress on the inner table respectively.

Curves of skull fracture threshold for compressive and tensile stresses ($\sigma_{\text{cr}}^c$ and $\sigma_{\text{cr}}^t$) were obtained by exponential approximation of the five pairs of the maximum $\sigma_{\text{eff}}$ out of five largest values and its $\Delta t_{\sigma}$ in each simulation. Any condition that is located on the JHTC which is the criteria of skull fracture can be interpreted as a lowest effective acceleration for triggering skull fracture. Thus, the maximum value among observed $\sigma_{\text{eff}}$ at impacted area was defined as a critical stress for skull fracture.

3.2 Results

Representative stress distribution on the impacted area at each skull layer from the simulation was shown in Fig. 9. At outer table and diploe, minimum principal stress, i.e., compressive stress, was significantly generated. In contrast, at inner table, maximum principal stress, i.e., tensile stress, was significantly generated.

Time histories of stress at each skull layer were also obtained from the simulations as shown in Fig. 10. The magnitude of the minimum principal stress at outer table was the largest followed by the maximum principal stress at inner table and the minimum principal stress at diploe, whereas the period of time of stress generation was the same for all layers.
Fig. 10 Time histories of maximum or minimum principal stresses at each skull layer on the impacted area in the forehead dropping simulation under condition #1

Values of $\sigma_{\text{eff}}$ and $\Delta t_\sigma$ were respectively calculated by equation explained in § 3.1 from the time histories of principal stresses at inner table and outer table on the impacted area. As shown in Fig. 11, tensile and compressive $\sigma_{\text{eff}}$ decreased along with the increase of $\Delta t_\sigma$. The two relations between $\sigma_{\text{eff}}$ and $\Delta t_\sigma$ were obtained as exponential equations, and proposed as stress-based skull fracture criteria ($\sigma_{\text{t crt}}$ and $\sigma_{\text{c crt}}$). The curves drawn by solid and broken lines in Fig. 11 can be expressed by Eqs. (3) and (4) where the units of $\sigma_{\text{crena}}$ and $\Delta t_\sigma$ are MPa and ms respectively. Additionally, the values of $\sigma_{\text{t cnv}}$ and $\sigma_{\text{c cnv}}$ at $\Delta t_\sigma$ of 10 ms where the JHTC converges to a value of approximately 1000 m/s$^2$, were 13.0 MPa and -47.0 MPa respectively. These convergent stresses ($\sigma_{\text{t cnv}}$ and $\sigma_{\text{c cnv}}$) were drawn as horizontal lines in Fig. 10.

$$\sigma_{\text{t crt}}(\Delta t_\sigma) = 44.7 \cdot \Delta t_{\sigma}^{-0.5225}$$ (3)

$$\sigma_{\text{c crt}}(\Delta t_\sigma) = -163.8 \cdot \Delta t_{\sigma}^{-0.5429}$$ (4)

Fig. 11 Effective stress versus duration time: plots of the results of the five forehead dropping simulations and the stress-based skull fracture criteria.

### 3.3 Discussions

To determine critical stresses for skull fracture (skull fracture criteria), we performed finite element simulations to obtain stresses at the skull surface during the head drop tests used for developing JHTC. We obtained the relations of tensile or compressive stress and its duration time for fracture occurrence as shown in the results presented in § 3.2. The method for developing stress-based criteria is considered to be an alternative way for determining the tolerance of head that has complicated structure and also applicable to any validated head finite element model. Furthermore, the criteria may help us elucidate the mechanism of skull fracture and analyze the further phenomenon, because the internal mechanical responses can be clearly observed in the finite element analysis.

The critical stress decreased along with the increase of its duration time, which indicates that not only the magnitude of stress but also its duration time is an important factor for skull fracture occurrence. This mechanism is supported by another rupture model, EWK model that was applied to analyze fracture at human ribs. We considered that...
our criteria are inherent to our head model, which means that the criteria depend on the
mesh size, shape and material properties of the head model used during the development.
The uniformity of the material properties of the skull\(^{(15)}\) makes our criteria equally evaluate
skull fracture at any site of the head model. Additionally, the convergent tensile and
compressive stresses in our criteria mean that those stresses will not cause skull fracture
even if they were generated more than 10 ms. The two \(\sigma_{cnv}\) are smaller than cortical bone
strengths reported in literatures. Bruce et al. \(^{(16)}\) reported that tensile and compressive
strength of bone varied between 53±11 to 135±16 MPa and 105±17 to 131±21 MPa
respectively. But, it must be noted that \(\sigma_{eff}\) is different from bone strength and is smaller
than the maximum stress as shown in Fig. 8. The gap between \(\sigma_{eff}\) and the maximum stress
increases along with the increase of \(\Delta t\), and the value of \(\sigma_{eff}\) at 6 ms of \(\Delta t\) is about
two-thirds of the value of maximum stress. An issue regarding our criteria is that the criteria
only consist of tensile and compressive stresses, i.e. normal stresses, not shear stress.
Fracture by shear stress, which is generated under impact to a sharp thing or under a
tangential impact, is unable to be evaluated by the criteria.

4. Implementation of the Stress-Based Criteria on Finite Element Analysis

Skull fractures occurring under various impact patterns are required to be reconstructed,
for example, in forensic investigations or in designing head protection devices. Types of
skull fracture depend on impact patterns as reported in a literature\(^{(2)}\). Depression fracture
occurs in high impact velocity. In the other hand, linear fracture occurs in low impact
velocity with large contact area. Experiments with cadaver heads or skull specimens are
difficult to be performed because the availability of those samples is limited, and the details
of skull fracture such as the start and the progress of fracture could not be observed.
Contrarily, finite element analysis using head model enables to investigate those details\(^{(17)}\).
In the simulation, stress or strain criteria are necessary to evaluate skull fracture occurrence.
Therefore, we undertook simulations using the head finite element model validated in
Chapter 2 and the stress-based skull fracture criteria proposed in Chapter 3 in order to
reconstruct skull fractures against two impact patterns.

Fracture was defined as a condition when the effective stress on an element and its
duration time exceeded the criteria. The rupture element was then eliminated by sharply
decreasing its Young’s modulus. With this fracture algorithm, skull fractures were
reconstructed in a dropping to a wooden floor simulation and in a golf ball impact
simulation where normal stresses were mainly generated. We could observe the occurrence
of skull fracture and its progress at outer and inner table in the two simulations. Time of the
first crack, magnitude of stress, and area of fracture were different between the two
simulations. We successfully showed that the head finite element model with the
stress-based skull fracture criteria can simulate skull fracture and enables to investigate its
details.

4.1 Analytical Methods

We implemented a fracture algorithm including fracture assessment and post-fracture
treatment based on the stress-based skull fracture criteria that we proposed in Chapter 3 into
LS-DYNA ver. 971 user’s subroutine written by FORTRAN 90 language as User Material
No. 41. An element of the skull model was defined to be a rupture element when time
integration of calculated principal stress reaches a product of critical effective stress and its
duration time as expressed in Eq. (5) and illustrated in Fig. 12. In this algorithm,
accumulation of stress is started to count when the stress exceeds \(\sigma_{cnv}\), which was defined as
insignificant stress for skull fracture as already described in \(\S\) 3.2. In case of reloading
after unloading, the time integration of stress is continued again if the stress generated
during reloading reached a stress value that is equal to the maximum stress value prior to unloading as indicated in Fig. 12(a). The outer and inner table may fracture due to compressive and tensile stress respectively as indicated by the results in § 3.2. Therefore, fractures at outer and inner table should be evaluated in terms of minimum principal stresses ($\sigma_{crt}$) and maximum principal stresses ($\sigma_{t crt}$) respectively. The diploe, a layer between the outer and inner tables, has high possibility of fracturing when the other layers fractured, because this bone layer has spongy structure and is more brittle than the cortical bone layer.

Post-fracture treatment was performed based on the Element Eliminate Technique (EET) commonly used in other research (18). EET makes the transmission of stress from an element to the surrounding elements suppressed after the element ruptured instead of totally eliminate the rupture element. This technique was actualized by decreasing the Young’s modulus of the element close to zero after it was judged as a rupture element. In our algorithm, stress of element should rest to zero at the next time step after rupture, however in order to avoid numerical calculation errors, Young’s modulus of the element was decreased gradually to zero by multiplying it with 0.001 at every time steps after rupture.

$$\int_{t_0}^{t} \sigma(t)dt > \sigma_{cr}(\Delta t) \cdot \Delta t$$

(5)

(a) Time history of principal stress of an element (b) Stress-based skull fracture criterion

Fig. 12 Cumulative effective stress and critical stress for skull fracture: (a) the hatching area means time integration of stress (cumulative stress) which is described by the left-hand side of Eq. (5), and (b) the hatching area means the product of a $\sigma_{cr}$ located on the threshold curve and its $\Delta t$ which is described by the right-hand side of Eq. (5)

Two forehead impact simulations were conducted as shown in Fig. 13 to simulate skull fracture by using the fracture algorithm. In the dropping to a wooden floor simulation shown in Fig. 13(a), the head model moved vertically down to a plate model at a speed of 6.0 m/s which is the fall speed of human to the floor reported in a literature (19). The square plate model was 220 mm on a side, weighed 0.05 kg, and consisted of 4-mm-square shell elements with a thickness of 10.0 mm. The material properties of the wooden floor were also taken from the literature (19). The density $\rho$, Young’s modulus $E$ and Poisson ratio $\nu$ were 900 kg/m$^3$, 11.0 GPa and 0.49 respectively. All nodes at the edges of the wooden plate were fully-restrained, the other nodes were allowed to move on the vertical axis only, and all nodes were tied to the top surface of a steel plate model with 5-mm-cube solid elements, whose material properties were $\rho$=7850000 kg/m$^3$, $E$=11.0 GPa, and $\nu$=0.49. In the golf ball impact simulation shown in Fig. 13(b), the head model without scalp was hit by a ball model at a speed of 60 m/s, which is the intermediate value of typical launching speed in golf that ranges from 44 to 75 m/s (20). The scalp part was removed because its deformation becomes largest immediate after the impact and its impact absorbing ability is negligible in a case where the contact area is small enough and the impact speed is adequately large. The golf ball model consisted of three layers with a diameter of 42.7 mm and a weight of 0.0449 kg according to a study about finite element modeling of a golf ball (21). The three layers
(core, mid, and cover) were modeled as elastic material. Their material properties were $\rho=1150 \text{ kg/m}^3$, $E=50 \text{ MPa}$, $\nu=0.49$ for the core part, $\rho=1150 \text{ kg/m}^3$, $E=25 \text{ MPa}$, $\nu=0.49$ for the mid part, and $\rho=950 \text{ kg/m}^3$, $E=400 \text{ MPa}$, $\nu=0.45$ for the cover part.

Frankfurt line

(a) Forehead dropping to a wooden floor (b) Forehead impact by a golf ball

Fig. 13 Forehead impact simulations to simulate skull fractures

4.2 Results

The two forehead impact simulations utilizing the skull fracture algorithm were found to have fracture on the forehead. Time histories of the principal stresses at each skull layer were obtained from the simulations as shown in Fig. 14. In both simulations, stresses at inner and outer table were continuously generated exceeding the level of $\sigma_{\text{cnv}}$, and then dropped rapidly after the time integration of the stresses exceeded the critical cumulative stresses defined in the skull fracture criteria. It indicates that the element ruptured and the skull fracture occurred. The diploe did not show a rapid drop of stress because fracture algorithm was not used for this part. However, as mentioned in the previous section, the diploe was also considered to get fracture. The times when fracture occurred were the same between inner and outer tables in the dropping to a wooden floor simulation as shown in Fig. 14(a), whereas in the golf ball impact simulation, the inner table fractured approximately 0.8 ms earlier than the outer table as shown in Fig. 14(b). The magnitudes of stresses and the time when fracture occurred in the dropping to a wooden floor simulation were respectively about half and 10 times of that in the golf ball impact simulation.

Fractures occurred at the location of impact varied in shape and range depending on the skull layers and the impact patterns as shown in Fig. 15. In the dropping to a wooden floor simulation, the fracture distributions at outer and inner tables had square and rectangle shape respectively. Moreover, the difference in numbers of rupture elements showed that the range of fracture at the outer table was larger than that at the inner table. In contrast, in the golf ball impact simulation, the range of fracture at the outer table was smaller than that at the inner table. Comparing between the two impact patterns, the ranges of fracture at the outer and inner tables in the dropping to a wooden floor simulation were respectively larger and smaller than those in the golf ball impact simulation.
Fig. 15 Distributions of fractures at the outer and inner tables in the forehead impact simulations: The red-colored elements were rupture elements.

The pattern of stress distribution at the impacted area in the forehead impact simulation will be different without the skull fracture algorithm, even its collision partner and the simulation conditions were the same, as shown in Fig. 16. The magnitudes of stresses on the elements surrounding the rupture elements in the simulation with fracture algorithm (Fig. 16(a)) were larger than that in the simulation without fracture algorithm (Fig. 16(b)), though their responses were very similar until the ruptures occurred. In other words, stress on an element tended to increase after the adjacent element ruptured.

4.3 Discussions

We simulated skull fractures in the two forehead impact cases by using a finite element head model with stress-based skull fracture criteria to demonstrate that the criteria is applicable for analyzing skull fracture in detail. Fracture was determined if cumulative effective stresses in an element exceeded the criteria, and the post-fracture treatment was performed by decreasing the Young’s modulus of the rupture element. In both simulations, dropping to a wooden floor and golf ball impact, skull fractures occurred at both outer and inner tables, resulting to a rapid drop of stresses in the rupture elements. Variety in the shapes and ranges of fracture distributions were also observed. Furthermore, increasing in stress of an element was observed after its adjacent element fractured.

In these results, the magnitude of stress and the time when the fracture occur depended on the impact patterns. A clinical report (22) said that the inner table broke prior to the outer table when linear fractures occurred at the loading area. Meanwhile, the outer table broke prior to the inner table when depressed fractures formed at the loading area. Therefore, our results indicated a possibility that our stress-based skull fracture criteria enable to reconstruct different skull fractures under different impact patterns, although the results could not be discussed quantitatively. Secondly, fractures at the inner table in the golf ball impact simulation expanded to a range that was wider than in the dropping to a wooden floor simulation that was less severe, which indicated that difference in severity of the skull...
fracture is depending on the impact patterns are reproducible.

The gap on fracture times and the difference in fracture areas between the outer and the inner table were caused by differences in the critical stresses between tension and compression. Especially, in the golf ball impact simulation, those differences were larger than in the dropping to a wooden floor simulation, because in short duration time of stress, a difference between the two critical stresses becomes comparatively large, as shown in Fig.10. It indicates that the details of the fracture development can be investigated by combining finite element simulation with the proposed skull fracture criteria.

Stress in an element changes depends on the rupture existence at its adjacent elements. Stress increases when the adjacent element ruptured because the rupture area significantly deformed due to the decrease of stiffness, resulting to a tensile load on the surrounding elements which still have initial stiffness. By adding an anisotropic material model and skull suture model, fracture with propagation behavior, i.e., linear fracture (2), might be also predictable in future.

Several issues have still remained in these applications. The two impact simulations were not validated against real-world accidents that cause fracture, because we could not get a detail skull fracture accident data including information regarding the accident condition and the fracture. Additionally, as we discussed in §3.3, our criteria did not consider shear stress, thus the simulations unable to simulate fracture which is caused by shear stress.

5. Conclusions

We proposed a novel method to develop stress-based skull fracture criteria for finite element analysis by reconstructing several drop tests of cadaver heads conducted in the past for developing an acceleration-based skull fracture criterion (JHTC). Our criteria indicate that the critical stresses for skull fracture decrease along with the increase of its duration time. It can be said that not only the magnitude of effective stress but also its duration time is an important factor for skull fracture to occur.

We successfully implemented the criteria and an element treatment method for post-fracture into a finite element solver. Skull fractures in two representative forehead impact patterns, plate and ball impacts, were then simulated in order to evaluate the potential application of the criteria on skull fracture analysis. In the two simulations, skull fractures occurred under different stresses, at different times and with different fracture distributions, indicates that the proposed analysis method may be possible to predict different skull fracture due to different impact conditions.

References


