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FINITE ELEMENT SIMULATIONS INVESTIGATING THE ROLE OF THE HELMET IN REDUCING HEAD INJURIES

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Abstract
Head injuries in motor vehicle crashes have high significance due to their fatal effects on the nervous system of passengers and pedestrians. In this study, using models of motorcyclists' helmet and the human head, two cases of head impact with a rigid surface are simulated by Finite Element Analysis; first, the impact of the head protected by the helmet and then, that of the unprotected head. In each simulation, impact parameters such as HIC, mass center velocity of the head, and pressures produced in the brain are calculated and corresponding parameters are compared with each other, which quantitatively represents the influence of the helmet on reduction of injuries to the head. Comparison of the results obtained in this study with those of previous researches indicates that if an appropriate finite element model is developed and used, FEM will lead to acceptable results in head impact simulations.
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Key Words: FEM, Simulation, Head Impact, Helmet

1. INTRODUCTION

Head injuries are an important field of study because of their fatal and irreversible effects on the nervous system of passengers and pedestrians. The head injury which leads to damage to the nervous system is neurotrauma. In fact, neurotrauma is a physical damage that happens when the human head is suddenly subjected to high levels of mechanical energy [1]. Traffic accidents, assaults, falls, and injuries occurring during recreational activities are some of the major causes of neurotrauma.

Most of the research in this field was initiated by military aircraft industry in the sixties, but today such researches are carried out and sponsored by car and motorcycle manufacturers. Worldwide head injury mortality rate is about 15 to 30 per 100,000 population annually. If a mean rate of 22 per 100,000 population is considered for the world population of 6 billion people, annually about 1.2 million deaths are due to the neurotrauma [1].

Among the causes of head injuries, those resulting from motor vehicle crashes (MVCs) have more importance due to considerable number of their injuries leading to death. In 1995, MVCs resulted in about 44,000 fatalities in the United States, 84 % of which were occupants of vehicles. Three factors are within the most important ones in determination of the injuries to the occupants [2]:

- The mass of colliding vehicles.
- The change in the velocity of vehicles during impact.
- The use of restraining devices such as safety belts for automobile passengers.

Motorcycles have the least safety among the vehicles because they are instable against impacts and a small disturbance in their motion exposes the motorcyclists to severe impacts. Such impacts lead to injuries to arms, legs, etc but the severest injury is to the head, which is more susceptible to impact. Using safety devices such as helmets can appreciably reduce this

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damage. Helmets are designed and fabricated in such a way that they surround most of the head and protect motorcyclists' against head injuries. It should be noted, although reducing the severity of injuries, a helmet cannot completely protect the motorcyclist against impact.

A standard helmet must pass several tests [3] that are designed to measure its protecting ability. The use of computer simulations among the researchers has recently increased to diminish the expenses of conducting experimental tests. There is a similar trend in the researches carried out on the head impact. It is because of limitations in conducting actual tests on living creatures (e.g. monkeys).

Many studies have been performed in this field. Research of Anzelius [4] in forties is among the first researches in this area. In fifties, Gurdjian et al. [5] conducted impact tests on animals. Nahum et al. [6] carried out similar tests and measured dynamic pressure in the brain of animals. After the advent of Finite Elements, three-dimensional models that included brain tissue were proposed [7]. In the nineties, due to the developments in computers and enhancements in their computational capacity, there was a significant advance in researches on head injuries [8]. Chang et al. [9] developed a finite element model of a helmet and studied impacts on its chin bar.

Aida [10] studied the influence of mechanical characteristics of the brain tissue on head injury criteria using FEM. He studied five different viscoelastic brain models and their linear alternatives using both a simple spherical model and a geometrically realistic head model. He concluded that head impact criterion (*HIC*) is insensitive to type of material used to model the brain tissue. Brands [11] also used FEM to study head injuries. He also used the viscoelastic model for the brain, and analyzed the effect of angular accelerations on head injuries. Kleiven [1] investigated the influence of the head and brain size, and the direction of the impact on injuries to the head using FEM. Kostopulos et al. [12] investigated the protection of helmets with different materials by FEM, and used a very simple model for the human head. A major difference between the present paper and reference [12] is that a geometrically realistic head model is used here instead of the above-mentioned simple model. In addition, unlike the research in question, in current analysis the brain response is also analyzed. Recently, Deck and Willinger [13] also used a simple model for head to optimize the helmet against multidirectional impacts using FEM.

In the present study, the models of a helmet, the human skull and brain are created, and the influence of the helmet on reducing the injuries is quantitatively investigated by FEM using ANSYS Ls-Dyna[1]. A realistic geometry for the skull and an approximate model for the brain are used here. Several parameters such as *HIC*, the velocity of mass center of the head, the pressures and stresses produced in the brain due to the impact are calculated.

2. BIOMECHANICS OF HEAD INJURY

2.1 Impact to the head

In most cases, head injuries are due to an impact to the head but injuries can also occur in the cases that the head does not experience any impact. In these cases, an impulse is transmitted to the head through the neck. In other words, both impact and impulse can accelerate a stationary head or decelerate a moving head, but in case of an impact, there are injuries directly produced due to the contact effects on the head. Severity of the impact depends on the impact velocity and physical properties of the colliding object [14].

From physical point of view, the case in which a moving head strikes a stationary object does not differ from the case where a moving object collides with a head. The only difference

[1] ANSYS Inc. (www.ansys.com), Southpointe 275 Technology Drive Canonsburg, PA 15317, USA

in these cases is in the reference frame that the event is observed. However, in practice, it has been observed that a moving head usually strikes an object which is more massive than the head but a stationary head is usually hit by an object with a similar mass.

Physical properties of striking object that determine the type of impact are shape and stiffness. The shape of striking object determines whether the object penetrates the skull or not (e.g. a bullet) but in cases where penetration does not take place, stiffness is the most important characteristic.

Location of the impact to the head is another parameter characterizing the injuries. It determines the location of local deformations in skull, linear and angular accelerations of the heads, etc [14]. Local deformation of the skull at the point of impact can result in injuries to underlying brain tissue. It occurs when the impact causes fracture in the skull. The possibility of fracture due to a given impact depends on the location of the impact. During an impact, there are also injuries in locations far from the impact point. These injuries can occur in both the skull and the brain.

Another important factor that determines the severity of injuries to the head is whether the head is free to move after the impact or not. If it is not, the injuries are directly related to the location and extent of the deformation in the skull. However, most cases of head injuries result from the impact of a moving head to a stationary object and injuries to brain. In these cases, injuries are due to the acceleration of the head.

For a given impact, if the line of action of impact force passes through the center of mass of the head, it will be subjected to linear accelerations. On the other hand, if it does not, the angular accelerations will be also present. This is a significant factor, which characterizes the severity of injury, and types of stresses produced in the brain [14].

2.2 Head Injury Criteria

Numerous researches [15, 16] were carried out to propose a formula providing a measure of severity of the impact to the head. Many experimental tests [17] were conducted and remarkable results were obtained. Some criteria were proposed by researchers but in seventies the criterion stated in (1) was proposed, which is the most famous criterion in head injuries. This criterion is based on acceleration of mass center of the head and has been frequently used in related studies;

$$HIC = (t_2 - t_1)\{\frac{\int_{t_1}^{t_2} adt}{t_2 - t_1}\}^{2.5} \tag{1}$$

Where, a is acceleration in terms of g (the gravity acceleration) and maximum value of (t_2-t_1) is 15 ms. t_1 and t_2 should be selected in such a way that the maximum value can be obtained for *HIC*. It is clear higher values of *HIC* correspond to higher chance of head injury (e.g. *HIC* value of 1000 corresponds to 16 % chance of head injury [10]).

Intracranial pressure is another parameter determining the severity of an impact. In fact, intracranial pressure is equal to the hydrostatic pressure at a point (i.e. the average of normal stresses with a negative sign);

$$p = -\frac{\sigma_{xx} + \sigma_{yy} + \sigma_{zz}}{3} \tag{2}$$

The intracranial pressure in the brain tissues underlying impact point (coup) is positive while in its opposite point (contracoup) is negative.

3. MATERIALS AND METHODS

In this section, the analysis performed in this study is explained in detail. ANSYS Ls-Dyna is used to conduct two simulations in which the head collides with a rigid obstacle. In the first simulation, the head is protected by a helmet but in the second one, the head directly strikes the surface without any protection. The simulation conditions including initial velocities and boundary conditions, etc are considered the same for both the simulations.

3.1 Modeling the Helmet and the Head

Due to the complicated geometry of a helmet, its modeling is a very sophisticated task. The method used in this study to create a three-dimensional model of a helmet is the direct measurement of dimensions of a typical helmet. In spite of being a time-consuming task, finally the model with a high resemblance to a real helmet is created in Mechanical Desktop (MDT) Software[2] (see Fig. 1). The thicknesses of the shell and the foam are assumed to be 7 mm and 22 mm, respectively.

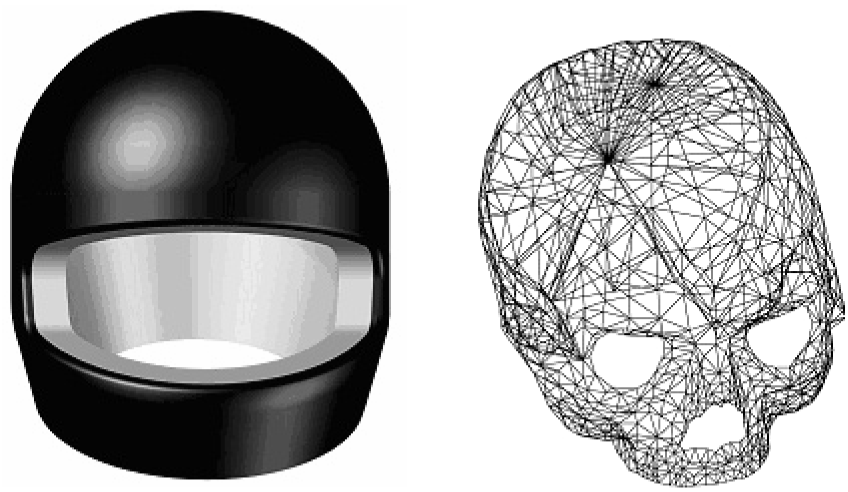

Figure 1: The models of the helmet (left) and the skull (right) used in this study.

In order to reach a properly meshed model, all the curved edges of the helmet shown in Fig. 1 are replaced by sharp edges in the model imported to ANSYS (in the case of not removing the curved corners, there would be numerous unnecessary elements after meshing).

In order to create the head model, a skull model generated by C.M.M is used (see Fig. 1). This model is imported to MDT, and by using the features of the software a three-dimensional model is created. In addition, due to the complicated geometries resulting in long computer runs, a simple model is used for the brain in these simulations. The upper part of the brain has the same curvature as that of the skull but as shown in Fig. 2, a flat plane is used for the lower part of the brain. Although the skull thickness varies for different locations, it is assumed to be constant values of 6 mm around the brain.

Since according to our assumptions, all of the conditions and models in the simulations are symmetric, the right half sections of the models are imported to ANSYS. Models of the skull and brain in the helmet and the rigid surface imported to ANSYS are shown in Fig. 2. The rigid surface is a cylinder with diameter of 200 mm and height of 70 mm.

[2] Autodesk, Inc. (www.autodesk.com), 111 McInnis Parkway, San Rafael, CA 94903, USA

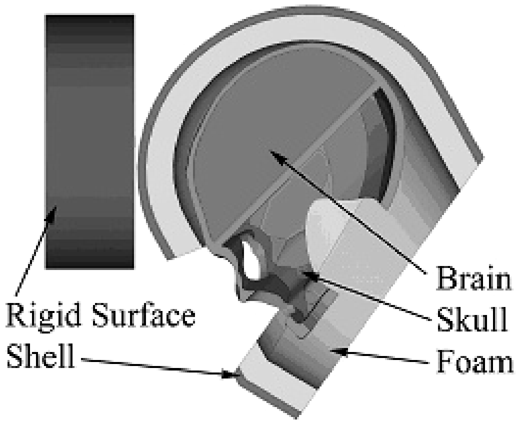

Figure 2: The analysis components imported to ANSYS.

3.2 Materials of the components

A bilinear kinematic model for the shell of the helmet is used. For the foam section of the helmet, the volumetric crushing behavior is assumed (see Fig. 3). A damping coefficient of 0.2 and tension cut-off of 0.5 MPa is also assumed for the foam [12]. Other assumptions and material properties are presented in Table I. The mass of the head model shown in Fig. 2 is assumed to be 1.501 kg (0.924 kg for the skull and 0.577 kg for the brain) while that of the helmet is assumed to be 0.686 kg. It is clear these masses are half of the mass of corresponding complete models.

Table I: Mechanical properties of the materials.

Mechanical Properties	**Rigid Surface**	**Shell**	**Foam**	**Skull**
Density (kg/m^3)	7800	1500	50	1610
E (GPa)	200	8	0.023	6.5
ν	0.3	0.28	0	0.22
Plastic Tangent Modulus (GPa)	-	4	-	-
Yield Stress (MPa)	-	50	-	-
Remarks	*Rigid*	*Bilinear kinematic*	*Volumetric crushing*	*Linear elastic*

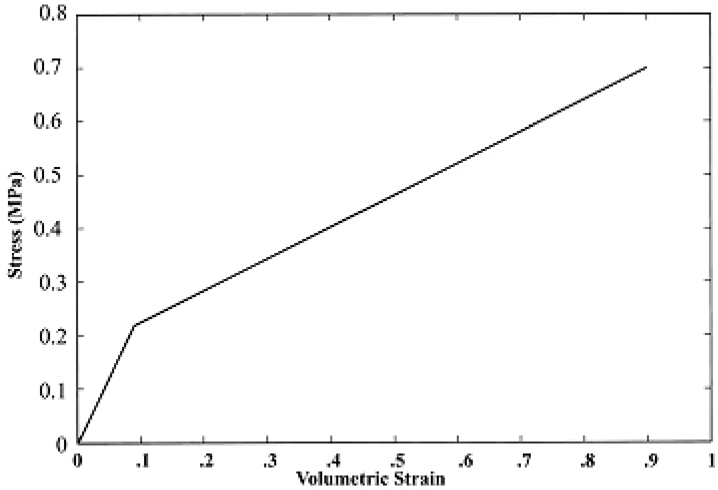

Figure 3: Foam Stress-Strain curve [12].

The viscoelastic model is used for brain tissue as indicated in Table II.

Table II: Mechanical parameters of the brain [11].

Bulk Modulus (Pa)	Relaxation Modulus (Pa)
1.28×10^{8}	$1.68\times10^{5}+\left(5.28\times10^{5}-1.68\times10^{5}\right)e^{-35t}$

3.3 Simulations by FEM

In the first analysis, the impact simulation for the protected head is carried out. Here, 56 elements for the rigid surface, 1215 elements for the brain, 5976 elements for the skull, 1545 elements for the shell, and 1125 elements for the foam of the helmet are used. It should be noted that linear elements (SOLID 164) are used for all the materials.

All the degrees of freedom of the rigid surface are constrained. Since the duration of the impact is very small, the gravity is ignored. The translational degree of freedom in the direction perpendicular to the symmetry plane for all of the colliding components (skull, brain, and helmet) is constrained. The friction coefficients between the helmet and the rigid surface as well as between the head and the foam are assumed to be 0.3 [12]. No specific assumption for strain effects other than ANSYS default assumptions for such dynamic simulations is considered.

In this study, simulations are carried out with the conditions of standard tests [3]. Although the goal of such tests is the evaluation of helmet protection quality, here, we consider these conditions in analyses that include the skull and brain. For the first analysis, the initial velocity of the helmet and the head (i.e. 8.28 m/s) is calculated so that the kinetic energy of the system is 150 J [3] or in other words, 75 J for the model imported to ANSYS. In the second analysis, the helmet is removed from the impact and the other parameters and initial conditions are assumed the same as those of the first.

4. RESULTS AND DISCUSSION

4.1 Results of the analysis

As expected, the comparison of the analysis results shows significant differences between the impact parameters. Fig. 4 represents the changes in the velocity of the mass center of the head during due to the impact.

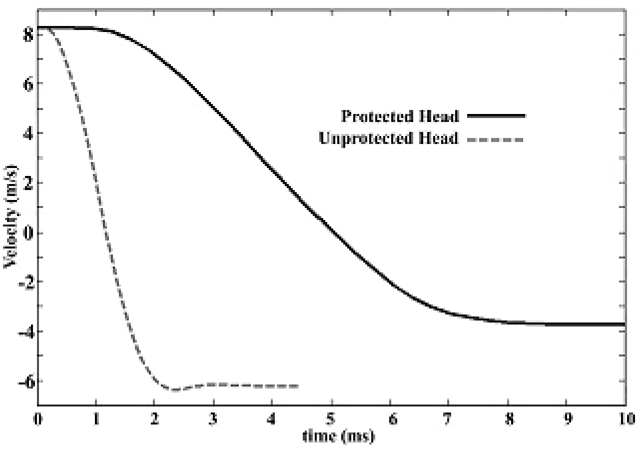

Figure 4: Velocity changes of mass center of the head during the impact in the simulations.

As shown in Fig. 4, the total change in the velocity in the first analysis (protected head) is 83 % of that of the second analysis (unprotected head). Although the difference is not significant, this change in the first analysis, takes place in a longer interval of time, which leads to appreciably lower impact acceleration in comparison with the second one. The mass center acceleration of the head calculated in the simulations is shown in Fig. 5. The maximum value of mass center acceleration for the protected head is 280 g while it is 1180 g for the unprotected head.

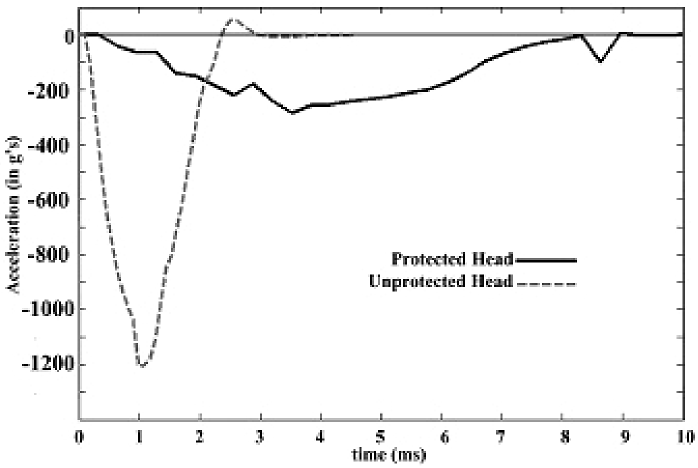

Figure 5: Mass center acceleration of the head during the impact in the simulations

In both simulations, as expected, the maximum pressure occurs in outermost point in the brain under the impact site (coup) while the minimum pressure is produced in its opposite point (contracoup). In Fig. 6, the time-history of coup pressure during the impact at coup site is shown.

The maximum coup pressure produced in the brain in the protected head simulation is 11.5 % of that of the unprotected head simulation. The contracoup pressures are similarly plotted and compared indicating the ratio of 31 % for minima of contracoup pressure, as shown in Fig. 7.

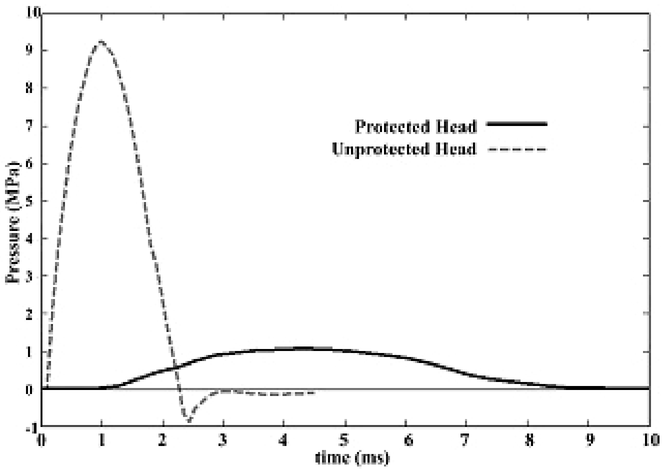

Figure 6: Coup pressure produced during the impact in the simulations.

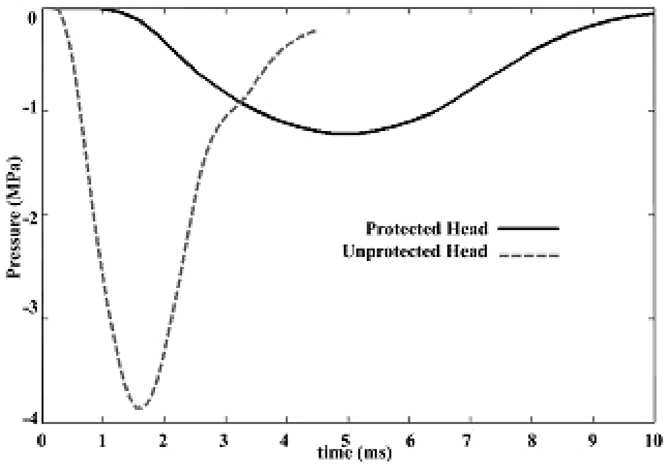

Figure 7: Contracoup pressure produced during the impact in the simulations.

Other parameters extracted from the simulations are presented in Table III. As it is observed, there is a considerable difference in *HIC*'s calculated from the results of the simulations. Actually, as expected, the helmet has absorbed a substantial portion of the impact energy in the protected head simulation.

Table III: Comparison of various impact parameters of the head in the simulations.

Impact Parameters	**Protected Head Simulation**	**Unprotected Head Simulation**
HIC	2070	25600
Maximum Von Mises stress in the brain (Pa)	47609	463852
Maximum Von Mises strain in the brain	0.047	0.413
Maximum Von Mises stress in the skull (MPa)	17.6	378
Maximum Von Mises strain in the skull	0.0033	0.071

4.2 Validation of the results

In this section, in order to investigate the quality of the finite element model and assumptions of previous simulations, a third analysis is set up. In this new analysis, the same geometry and properties as those used in the main simulations of this paper are assumed for the head model while the impact conditions of this analysis are adopted from reference [18]. As shown in Fig. 8, the coup pressure obtained by the present model approximately follows its corresponding graph in the research in question. Noting the complexity of the impact phenomenon and nonlinearities in material properties of the brain, this amount of difference between two graphs can be considered acceptable and therefore, the results obtained by this head model will be reliable.

Although no direct validation was found for the protected head simulation, comparison of the value of *HIC* in the present study (i.e. 2070) with that of previous studies that used the same test conditions demonstrates the validity of the results of the first simulation. As expected, the calculated value for *HIC* in the present research, like that of reference [12] is in the order of 2000. In addition, according to the standard used here, it is required for maximum head acceleration to be less than 300 g, which is met in our protected head simulation.

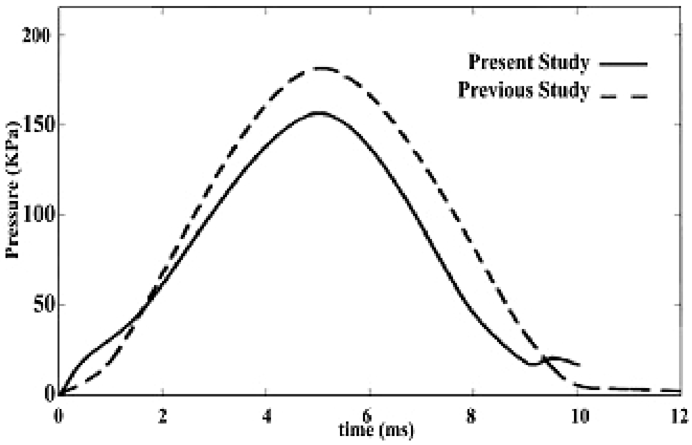

Figure 8: Comparison of the coup pressure obtained by present model and that of ref. [18].

5. CONCLUSIONS

In this paper, two analyses were carried out in which the impact of a protected head and that of an unprotected head were simulated by realistic models of the head and the helmet using the conditions of standard tests. Various parameters such as brain stresses and strains, kinematic parameters such as *HIC*, velocity changes, etc were determined and compared. The results show drastic variations in severity of head injury from the first analysis (the protected head) to the second analysis (the unprotected head). Comparison of the results with similar studies shows that properly developed finite element models by a geometrically realistic human head can lead to acceptable results in impact simulations of the head protected by a helmet. Due to limitations in computer computational capacity, average number of elements and a simple model for the brain were used. In future works, by a realistic brain model and increasing number of elements of the components, more precise results can be obtained.

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