Brain Pressure Wave Propagation during Baseball Impact [†]

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Abstract: The purpose of this study was to examine a mechanism of brain injuries during baseball impact. A baseball helmet was attached to a novel surrogate head, which simulated the intracranial structure of a human head, and baseball impact tests were carried out using a high-speed cannon system. In addition, the baseball impacts were simulated using a corresponding finite element model of the head and helmet. From the results of both the experimental and simulated impacts, the peak acceleration of the brain was greater than that of the skull, which was due to the propagation of pressure waves, in turn reflected in the intracranial space. The peak negative pressure reached the cavitation threshold on a broad area of the brain surface, repeatedly. This phenomenon was different from the brain deformation in other impact conditions such as football and traffic accident cases. Therefore, a new design philosophy for a helmet which reduces the effects of pressure wave propagation may be required to mitigate brain injuries.

Keywords: baseball; brain injury; novel surrogate head; finite element method; pressure wave propagation

1. Introduction

Mild traumatic brain injuries (mTBI) occur in high-speed projectile sports due to collisions of the ball with the batter's helmet [1,2]. In Japan, 794 head injuries occurred in baseball during school activities in 2005–2011, resulting in the highest number of head injuries compared to other sports [2]. Preventative measures, such as helmets, have been developed to protect against these collisions in baseball. However, the helmet performance is currently evaluated through its ability to reduce the resultant peak linear head acceleration [3]. Whilst the helmet can usefully protect against severe head injuries (e.g., skull fractures), it is unclear to what extent it can protect against mTBI. Baseball impacts have previously been conducted on the Hybrid III surrogate head with equivalent finite element simulations, driven by experimental acceleration responses, to examine the brain deformation mechanism and impact reduction effect of helmets [4]. However, the Hybrid III surrogate head has low biofidelity due to its overly stiff skull and hollow construction, and therefore cannot evaluate the intracranial stress wave propagation phenomenon that could be important in short-duration impacts, such as baseball collisions [4]. Authors of the current research have previously developed a novel surrogate head which incorporates detailed intracranial structures and has successfully been used to investigate intracranial brain deformation mechanisms during vehicle collision events [5]. It was

hypothesised that its improved biofidelity (compared to the Hybrid III) would also be useful in a short-duration impact. Therefore, the main aim of this research was to conduct baseball impacts with the novel surrogate head to examine the mechanism of intracranial brain deformation. Secondly, equivalent finite element impacts were simulated to validate the experimental sensor results and to investigate pressure wave propagation during the impacts.

2. Materials and Method

2.1. Novel Human Head Surrogate Capable of Evaluation of Intracranial Brain Deformation Behavior

The head surrogate [5] consisted of a skull, mandible, skin, cerebrospinal fluid (CSF), falx and tentorium parts. In addition, for this research, the novel surrogate head was advanced by embedding accelerometers in the right and left side of the brain. The skull and mandible were made of transparent polycarbonate and based on the 3D CAD model of an adult male (age: 25, height: 173 cm, mass: 65 kg) (Figure 1a). The skull model was divided into frontal and occipital parts to insert a brain part, and the two parts were assembled with low-profile fasteners to prevent the skull from interfering with a helmet. A 6DOF sensor (DTS 6DX PRO: DTS) was mounted inside the nasal bone component. The falx and tentorium, which act to constrain intracranial brain motion, were made of 1mm polyurethane sheet and fixed in the skull model. The brain model consisted of the right and left cerebrum, cerebellum and brain stem (Figure 1a). The brain CAD model was constructed from the adult's MR images, and the brain mold was constructed with a 3D printer (FORTUS250mc: Stratasys). Silicone gel (Sylgard527: Dow Corning), whose dynamic viscoelasticity is equivalent to that of the human brain, was injected and cured in the mold. In addition, two 6DOF sensors capable of measuring linear acceleration and angular velocity were implanted in the left and right cerebrum to measure the intracerebral acceleration during baseball impacts (Figure 1b). The 6DOF sensors were put into low-density foam blocks so as to maintain the density of the brain. The skin is a silicone rubber that satisfies the biofidelity requirements of the THOR dummy head [5]. To assemble the head, the brain model was inserted in the skull, and the intracranial space was filled with water to simulate cerebrospinal fluid. The total weight of the head model, including the neck joint, was 4.2 kg and the principal moments of inertia were $I_x = 0.014 \text{ kg}\cdot\text{m}^2$, $I_y = 0.019 \text{ kg}\cdot\text{m}^2$ and $I_z = 0.018 \text{ kg}\cdot\text{m}^2$.

2.2. Baseball Impact Tests

The novel surrogate head was attached to the Hybrid III surrogate neck and fixed to a projectile impact test apparatus in an upright position, as shown in Figure 1c. Rawlings Coolflo baseball helmets, which satisfy the NOCSAE test standard, were selected. Rawlings ROLB1 major league baseball (MLB) balls were projected by a high-speed pneumatic ball launcher. Balls with closely matched mass, diameter and stiffness (evaluated by the recorded peak force of a 1-meter-high drop test on a rigidly mounted load cell) were selected. The impact velocity was 129 km/h (80 mph), which was the same speed as that of a previous study [4]. The impact location (lateral low) was defined as a point midway along the coronal plane and at the height of the Glabella, where concussion often occurred in MLB games [1]. The triaxial linear acceleration and angular velocities at the skull, right and left cerebrum were obtained, and the raw data was filtered with CFC1000 filter.

Figure 1. Novel human head surrogate.

2.3. Head-Neck Finite Element Model

A finite element (FE) head model based on the same three-dimensional shape as the novel surrogate head was constructed in our previous research [5]. The material characteristics were defined based on human material properties, and the intracranial brain motion was validated against a cadaver study [6]. The volumetric response of CSF was modeled using the Mie–Gruneisen equation, and its material properties were used for water constants with -100 kPa cut-off pressure, in order to simulate the cavitation occurrence in the intracranial space.

The FE model included the three-layered skull, dura mater, falx cerebelli, tentorium cerebelli, CSF, ventricles, cerebrum, cerebellum, brain stem and skin. The total number of elements in the model was approximately 270,000. In addition, a finite element model of a Hybrid III neck (Simplified Hybrid III 50 percentile neck model: LSTC) was connected to the head model by constraining the node group near the occipital condyle joint on the skull of the finite element head model and the upper surface of the neck model. LS-DYNA ver971 was used as the solver.

2.4. Ball and Helmet Finite Element Models

Both the ball and the helmet were modelled with simple material characteristics, i.e. linear elastic bodies, since the main purpose of this study was to investigate the mechanism of brain deformation behaviour, and the requirement for the ball-helmet structure was to reproduce the external force waveform acting on the human head. As with the balls used in the projectile impact tests, the diameter of the ball was 72.3 mm and the weight was 0.144 kg. Young's modulus of the ball model was identified, so that the maximum load in the 1-meter free drop simulation was equivalent to the actual drop tests (Table 1). The three-dimensional shape of the helmet FE model was constructed based on the measurement of the helmet shape used in the projectile tests by a 3D scanner. The components of the helmet model were the shell and the liner. The material property of the helmet shell was based on ABS. A reconstruction simulation of the baseball impact tests using the Hybrid III surrogate head was conducted to identify the elastic properties of the helmet liner, so that the resultant acceleration waveform at the centre of gravity (CoG) of the Hybrid III head was equivalent to the test result (Figure 2) (Table 1). The maximum value and waveform were in good agreement, as shown in Figure 3.

Figure 2. Baseball impact simulation using the Hybrid III head finite element (FE) model to identify material properties of the helmet model.

Figure 3. Comparison of the resultant acceleration of the head's centre of gravity (CoG) when the ball impacts the Hybrid III head with the helmet.

Table 1. Material properties of the helmet parts and the ball.

	Material	Density [kg/m ³]	Young's Modulus [GPa]	Poisson's Ratio
Helmet shell	Elastic	660	2.8	0.35
Helmet liner	Elastic	10	0.010	0.10
Ball	Elastic	650	0.018	0.38

2.5. Baseball Impact Simulation Using the Human Head FE Model

Reconstructive simulations of the impact tests were conducted using the helmet, ball and head-neck FE models (Figure 4). An initial velocity of 129 km/h was input to the ball model, and the ball collided with the lower temporal region of the helmet as shown in Figure 4. Contact definitions were set for balls and helmets and for helmets and skin, respectively. The acceleration responses at the 6DOF sensor locations were obtained. In addition, brain pressure responses at five locations in the brain, from the collision point to the opposite side, were obtained, as shown in Figure 5.

Figure 4. Initial condition of the baseball impact simulation using FE models of the human head, the helmet and the ball.

Figure 5. Five elements where the brain pressure responses were obtained.

3. Result

3.1. Baseball Impact Tests

Figure 6 shows the resultant acceleration responses at the skull and left and right cerebrum measured in a baseball impact test. Figure 7 also shows the angular velocity responses measured at the skull sensor. The acceleration responses of the skull and brain showed oscillatory waveforms with a duration of about 4 ms. The rise times of the resultant acceleration waveform at the right and left cerebrum showed earlier than those at the skull, and the maximum value of the accelerations in the cerebra were greater than the value at the skull. The peak change in angular velocity around the sagittal axis, shown as the *x*-axis in Figure 7, was about 13 rad/s.

Figure 6. Resultant linear acceleration responses at the Brain Left, Brain Right and Skull points, where 6DOF sensors were inserted, in the case of the impact tests.

Figure 7. Angular velocity responses at the skull in the case of the impact tests.

3.2. Baseball Impact Simulation

The resultant acceleration responses at the skull and brain sensor locations obtained by finite element analysis are shown in Figure 8. The acceleration waveforms were multimodal, the rise time at the brain sensor locations showed earlier than that at the skull sensor and the maximum value at the brain sensors was higher than the value at the skull. The maximum value of the maximum principal strain in the brain was relatively small: 0.17. Figure 9 shows the brain pressure responses at the five brain elements shown in Figure 5. Firstly, high positive pressure was generated at element a, closest to the collision site, and then the positive pressure rose in the order of elements: b, c, d and e. In addition, at element e (the opposite side of the collision point), after the positive pressure was generated, the sign of the pressure was reversed. The negative pressure reached -100 kPa, close to the CSF cavitation pressure, while the sign of the pressure was repeatedly changed. In terms of the magnitude of negative pressure, elements a and e (near the skull-CSF boundary) reached the cut-off pressure, repeatedly.

Figure 8. Resultant acceleration responses at the Brain Left, Brain Right and Skull points, where 6DOF sensors were inserted, in the case of ball impact simulations.

Figure 9. Brain pressure responses at five locations in the cerebrum.

4. Discussion

In both experiments and simulations results, left and right cerebrum acceleration responses showed an earlier rise time and higher peak values than those at the skull sensor. This was because of the influence of the pressure wave propagation in the intracranial space. As shown in Figure 9, the pressure in the brain first rose from the side of the collision, propagating from the point closest to the collision to that furthest from the collision. When it reached the contrecoup side (i.e., opposite the collision site) of the skull, the sign of the pressure was converted to negative by a free-end reflection at the boundary between the skull and CSF. After that, the negative pressure propagated in the opposite direction. When the negative pressure wave returned to the skull boundary at the collision site, the sign of the pressure wave was reversed again. Although this pressure wave was gradually attenuated, a negative pressure of -100 kPa was repeatedly generated near the skull boundary. There is a hypothesis that the negative pressure which induces CSF cavitation could cause brain injury [7]. Therefore, the hypothesis of brain injury mechanism during baseball impact might correspond to the cavitation theory.

Recent studies on brain injuries in sports have mainly focused on longer-duration impacts such as player collision in football [8]. Based on the results of reconstruction simulations in the football impact cases and traffic accident cases, the rotational motion of a head showed good correlation with the shear deformation of brain tissue, and various injury criteria based on the head's rotational motion have been proposed [9,10]. However, in the case of the baseball impact, the peak change in angular velocity and maximum principal strain in the brain was lower than in football cases [8].

Rather, there was an increase in brain acceleration due to the pressure wave propagation by the short-duration impact, as described above. The mechanism of intracranial brain deformation in the baseball impact may differ from that of traffic accidents or football cases. Therefore, a new injury criterion and head protection measures might be required, considering the effect of pressure wave propagation in hard ball collision cases, with shorter duration.

5. Conclusions

The purpose of this research was to examine mechanisms of brain injuries during baseball impacts, based on the results of both baseball impact tests using a novel surrogate head and simulations using finite element models. From the experimental and simulation results, the peak acceleration at the brain was greater than that at the skull, which was due to pressure waves propagating and reflecting in the intracranial space. The peak negative pressure reached the cavitation threshold on the brain surface, repeatedly. Additionally, the head's rotational motion was smaller than that of head impact cases in longer-duration events. Therefore, the mechanism of intracranial brain deformation in baseball impact cases is different from head impact cases in other sports. A new injury criterion and a new design philosophy for baseball helmets might be required, reducing the effect of the pressure wave propagation which originates from the short impact duration.

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