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Brain injury risk estimation of collegiate football player based on game video of concussion suspected accident

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Abstract
The collision accident in collegiate football game was simulated based on the game video and the concussive impact on the head was analyzed. First, the collision motion of players was reproduced based on the video by using motion analysis, and the translational and rotational velocities, relative position and contact location of the struck and the striking players' heads just before the collision were calculated. Then the data obtained were input to two helmeted finite element (FE) human head models as the initial condition, and the brain injury risk was evaluated by using the impact analysis. The FE helmet model was validated by a drop test of the helmet in which the head impactor was embedded. In the present study, two concussion suspected accident cases were analyzed; then the concussion was evaluated by ten mechanical parameters generated inside the skull caused by the collision. The injury risk evaluated by multi parameters belonged to the dangerous range that may cause concussion and was consistent with the diagnosis of the medical team doctor. The brain injury risk can be successfully estimated by the reconstructed simulation of the game video and FE analysis. To our knowledge, this study is the first attempt in Japan to estimate the brain injury risk systematically by a combination of game video analysis which is originally introduced for the players' health care and FE analysis by helmeted human head model. In the future, brain injury risk caused by an accident can be evaluated with higher accuracy by analyzing more accident cases.

Key words : Brain injury risk estimation, Sports-related concussion, Game video, Motion analysis, FEM

1. Introduction

American football is known as a contact sport. On-field head impact accident in football game causes traumatic brain injuries (TBI), and repeated TBI may result cumulative damage to cells of the brain although each damage is mild as concussion and may produce rapid catastrophic deterioration (second impact syndrome) or chronic traumatic encephalopathy (CTE) presenting with cognitive dysfunction (Keener A.B., 2016). Concussions may occur in most contact sports and even noncontact sports in which inadvertent head impacts occur. And one poorly understood aspect of TBI is mild traumatic brain injury (mTBI) such as concussion because post-injury sequelae are difficult to address at the cellular level *in vivo* (Slemmer J.E. *et al*., 2002).

In order to estimate the risk of the sports-related mTBI represented by concussion, the real world accidents suspected concussion must be analyzed biomechanically, where collision causing concussion is simulated and impact mechanics is determined. If the impulsive force added to the head in collision can be known, the damage of the head is analyzed by using a FE method in detail. A clear understanding of injury mechanism may help to explain concussion

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risks and lead to further improvement in safety equipment (Viano D.C. *et al*., 2007, Willinger R. *et al*., 2013). However, we usually do not have any effective method to obtain the mechanical information in the accident.

One of the most reliable methods to determine the collision kinematics is the analysis of game video. Records of video camera are becoming more popular and more high-resolution by rapid technological development and lower expense. Recent year, videos are often used during daily practice and games to check the formation design, assignment play and analyze the tactics of the opponent team. The information obtained from the video is not only important for improving the competition ability of the players, but also plays another important role as a tool for confirming what happened to the players when collision accident occurred. In particular, TBI represented by concussion is rather difficult to diagnose visually than trauma of extremities, the video is often required to confirm the motion of the players after the collision. Currently, it is new in memory that the video analysis staff (medical staff) was allowed to reside in the international tournament in World Rugby (WR), and had the privilege of ordering to implement the protocol of concussion testing when a concussion suspected collision occurred, and the analysis was carried out for 10 or more concussion suspected collisions at the last World Cup. Notwithstanding such a background, quite few opportunities are given to reconstruct the game video of accident for estimating the brain injury risk.

In Nihon University American Football Club, the movement of players has been recorded on video from multiple directions during the exercise and the game to keep the health of the players and to develop a game plan. If a player is diagnosed as concussion by the team doctor, the player is permitted to return to the game and daily physical exercise only after passing the concussion diagnosis test and undergoing a series of staged exercise according to the protocol defined in the American Academy of Neurology guidelines. As concussion diagnostic tool, SCAT2 (Sports Concussion Assessment Tool 2) and Cogsports test (computerized cognitive test, CogState Sport Ltd., Australia) are developed. The tools evaluate and manage the concussion by scoring the symptoms, physical signs, impaired brain function and abnormal behavior of the concussion suspected player.

The health care system of the collegiate football player in Nihon University is well organized and advanced in college sports in Japan. Although the activity for health care is achieving the result steadily, the more biomechanical approach is strongly required because TBI research represented by concussion has remarkably been advancing for years and leading to a greater insight of the cause and effect of a mTBI (Duma S.M. *et al*., 2005).

In the previous study (Aomura S. *et al*., 2010 and Zhang Y.L. *et al*., 2011), the pathogenic mechanism of brain injury due to impact was studied. The impact analysis by using FE human head model was reported to evaluate the type and severity of brain injury by mechanical parameters inside the skull. In order to quantitatively evaluate the concussion generated by on-filed head-to-head impact accident, the linear and angular accelerations that occurred during the impact were measured using the acceleration sensors installed in the helmet (Pellman E.J. *et al*., 2003). In the laboratory experiment, in order to measure the acceleration during the impact the collision play was reconstructed by using human dummy models, and the relative velocity and collision location of the head of the players were determined by analyzing the game video (Pellman E.J. *et al*., 2006, Viano D.C. *et al.*, 2007). These reports discussed and evaluated the tolerance of concussion by the indexes such as HIC, SI, GAMBIT or HIP which were calculated by the measured acceleration. Furthermore, impact analysis using FE human head model was performed to know the distribution of the mechanical parameters inside the skull caused by the impact (Zhang L.Y. *et al*., 2004, McAllister T.W. *et al*., 2012). In these reports the tolerance of concussion evaluated by strain, strain rate and von Mises stress of brain was estimated.

In this study, injury assessment system is proposed to evaluate the risk of mTBI in collegiate football player based on the game video of collision accident. Collision accident causing concussion in the game was simulated by using mathematical model of the whole body of striking and struck players, and head kinematics during the impact was determined. The collision motion was reconstructed by using video taken from two directions. The injury risk was evaluated by using FE human head model with helmet, in which the initial conditions such as translational and rotational velocity, collision location, position and posture of the heads of two players were given by whole body simulation. Biomechanical parameters inside the skull obtained showed large enough to cause concussion as diagnosed by medical team doctor. The reconstructed collision play can also be proposed as a promotion animation for preventing the possible dangerous play. The assessment system based on the game video and FE analysis will contribute to improve the understanding of the mTBI quantitatively and to make a good correlation with the concussion diagnostic program based on interview and observation as SCAT2. The detail of the proposed assessment system and the results of two on-field collision cases are reported in this paper.

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2. Injury assessment system and materials

2.1 Injury assessment system

The schematic illustration of the injury assessment system is shown in Fig.1. The system consists of two parts, the motion analysis part by mathematical model of whole body, and the evaluation part of intracranial parameters by FE head model with helmet.

At the first part the motion during the accident is reproduced by the whole body mathematical model and the kinematics of the colliding two players is determined. The motion analysis based on the game video is performed by using human models implemented in MADYMO software. The human models of MADYMO are validated against post-mortem human tests (MADYMO Human Body Models Manual 7.5). In the motion analysis part, it is difficult to apply the detailed FE head model and helmet model to the simulation, because the motion is simulated kinematically by using mathematical whole body model of more than one player. Therefore, although the posture, acceleration and velocity just before collision can be calculated with some reliability, the reliability of the motion and the intracranial response cannot be guaranteed after collision because the model of the helmet and the contact characteristic of the helmet are difficult to represent in the motion analysis software.

At the second part the intracranial mechanical parameters causing the head injury are calculated by FE analysis (LS-DYNA ver.8.0). In the FE simulation, the relative velocities and position of the heads obtained by whole body simulation are input to the helmeted FE human head models as initial condition. Next the head injury risk is evaluated by some indexes previously proposed by other researchers, where these indexes are calculated based on mechanical parameters obtained by FE analysis.

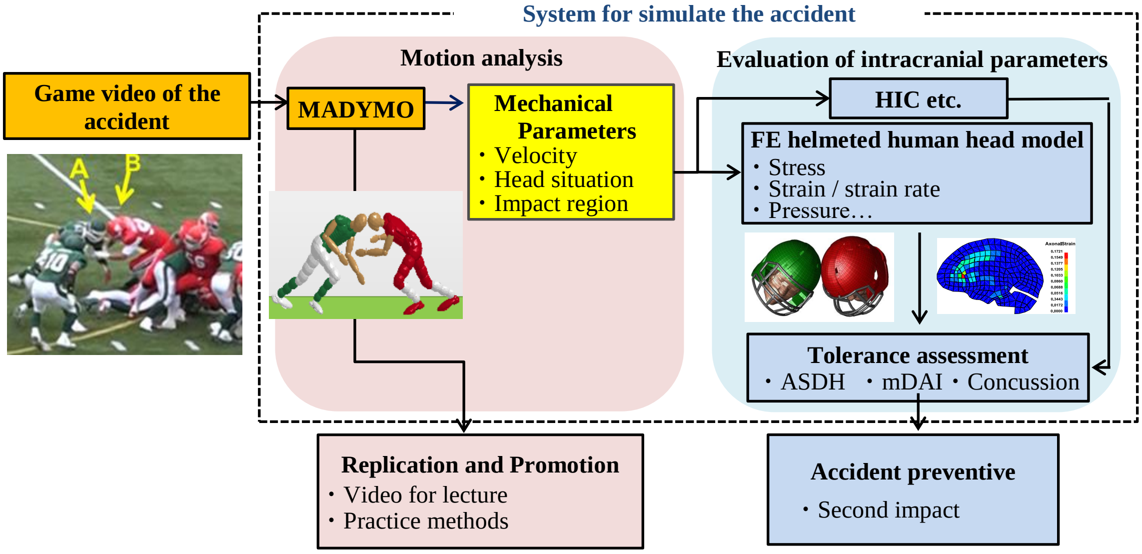

Fig.1 Injury assessment system of TBI based on the accident game video.

2.2 Materials

2.2.1 Finite element human head model

A FE human head model was constructed using sagittal-sectional T1 weighted MRI data of a man's head. The model includes the main anatomical features: scalp, skull, cerebrospinal fluid (CSF), cerebrum, corpus callosum, ventricle, brain stem, falx and tentorium as shown in Fig.2. The three-layered structure of the skull, which consists of an outer table, diploe, and inner table, was also reproduced. In this model, the falx and tentorium are constituted of shell elements, and the others are constituted of hexagon elements. The FE model consists of 89,226 nodes and 74,462 elements with a total mass of 4.2 Kg. The material properties of each part of the model are shown in Table 1. Elastic properties were assigned to the scalp, skull, falx and tentorium, and viscoelastic properties were assigned to the CSF, cerebrum, brain stem, corpus callosum and ventricle.

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2.2.1.1 Model validation by Nahum's experiment

In order to verify the FE head model, the numerical results of the impact test were compared with those results of the cadaver experiment by Nahum A.M. *et al.*, (1977). The outline of the experiment is shown in Fig.3(a). In the experiment, 5 kg iron impactor was hit on the head at 6 m/sec. The time-force history was measured in the experiment as shown in Fig.3(b) and this time-force history data was applied to the head model in numerical calculation. The restraint condition of the head model was free and tied-type contact condition was used between each part of the head model. The experimental pressure and acceleration response obtained by Nahum A.M. and those obtained by numerical calculation with the finite element model in this study are shown in Fig.4. Although slight difference is observed between the experimental cadaver test and numerical results, the head model is sufficiently valid to predict the intracranial pressure and acceleration.

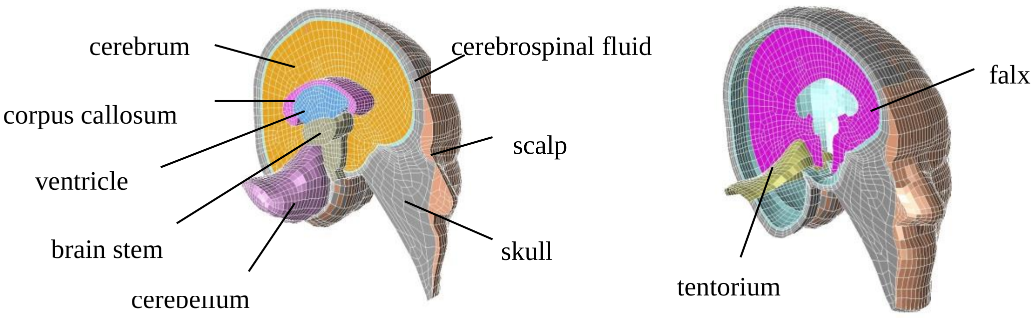

Fig.2 The developed finite element human head model composed of scalp, skull, cerebrospinal fluid, cerebrum, corpus callosum, ventricle, cerebellum, brain stem, flax and tentorium.

Table 1 Material properties of the FE human head model.
(Zhang L.Y. *et al.*, 2002, Willinger R. *et al.*, 2003, Viano D.C. *et al.*, 2005)

	Scalp	Facial bone	Outer/Inner table	Diploe	CSF/ Ventricle	Cerebrum	Brain stem	Corpus callosum	Falx/ Tentorium	Cerebellum
Density ρ [kg/m^3]	1000	1213	1456	850	1040	1040	1040	1040	1130	1040
Young's Modulus E [MPa]	16.7	5000	5000	2320	-	-	-	-	31.5	-
Bulk Modulus K [MPa]	-	-	-	-	2190	2190	2190	2190	-	2190
Short Time Shear Modulus G_0 [MPa]	-	-	-	-	-	0.0125	0.0225	0.041	-	0.01
Long Time Modulus G_∞ [MPa]	-	-	-	-	0.0005	0.0025	0.0045	0.0078	-	0.002
Poisson's Ratio ν [-]	0.42	0.23	0.25	-	-	-	-	-	-	-
decay coefficient [sec^{-1}]					500000	80	80	400	0.45	80

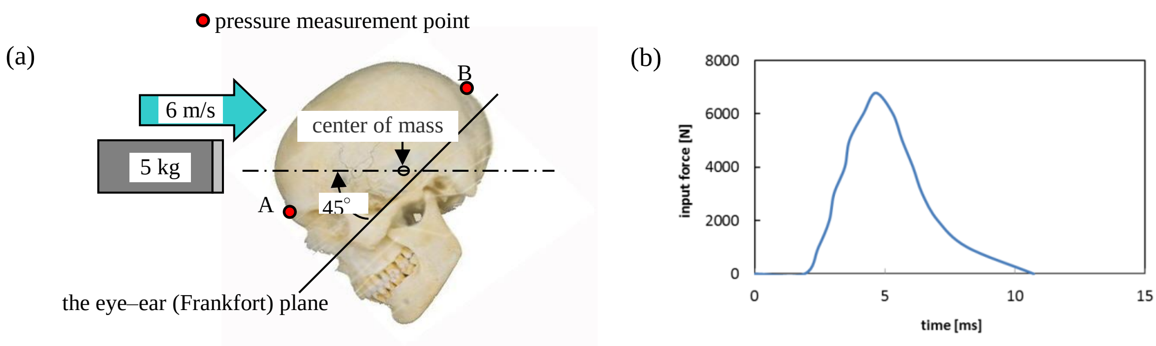

Fig.3 Experiment by Nahum A.M.. (a) The 5 kg iron impactor was hit on the frontal region of the head at 6 m/sec, and intracranial pressures were measured in the frontal (point A) and occipital (point B) region of the head. The acceleration was measured at the center of mass. (b) The input force curve obtained from the experiment.

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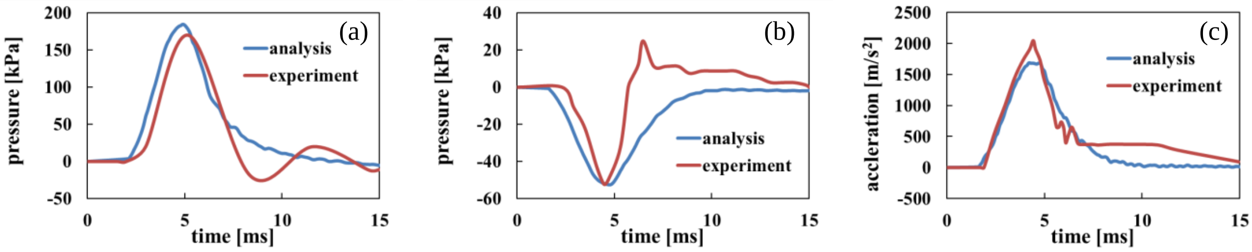

Fig.4 Comparison between the cadaver experiment and numerical calculation, (a) pressure response of frontal region (point A), (b) pressure response of occipital region (point B), (c) acceleration response of center of mass.

2.2.1.2 Model validation by Hardy's experiment

The brain/skull relevant displacements of the model were validated by replicating the cadaver experiment by Hardy *et al.* (2001, 2007). In the experiments, the neutral density targets (NDTs) were implant inside the human cadaver brain and a unique high-speed biplanar X-ray system was used to tracking the NDTs to calculate the brain/skull relative displacements caused by the impact. In this study, the skull was modeled as a rigid material and the liner and angular accelerations of head extracted from the experiment data (Fig.5) were applied to the center of gravity of the head model. The 3 impact experiments from 3 directions, a frontal impact (C383-T3), an occipital impact (C288-T1) and a left temporal impact (C380-T3) were replicated using the head model. The relative displacements of NDTs obtained from each experiment and the relevant nodes of the model are shown in Fig.6. In the figure, the left side column shows the relative displacement obtained from the experiment, and the right column shows the relative displacement obtained from the simulation. Each small curve represents the trajectory of a NDT. The results shown by the displacements of the simulation were smaller than the experiment, the trajectories of the nodes of the model were also tending to follow loop, so the motion patterns are similar to the experiment data. The head model is sufficiently valid to predict the relative displacement of the brain.

2.2.2 FE helmet model

Many effective head protections by helmet have been reported mainly for collegiate and professional football (Daring T. *et al.*, 2016, Johnston J.M. *et al.*, 2015, Post A. *et al.*, 2013), and for other sport, boxing, including head gear (Plant D.J. *et al.*, 2013). In this study, a FE helmet model with a polycarbonate shell and open celled foam (OCF) (Daring T. *et al.*, 2016), the same type which the injured player was used during the game, was developed. The geometry of the helmet was obtained by measuring the dimensions of the helmet. The outer shell is meshed by shell elements with a constant thickness of 4.5 mm and is modelled by an elastic law. The liner is meshed with hexagon solid elements, obtained by extrusion of the outer shell, with a total thickness of 35 mm and the characteristics are derived from experimental drop tests as shown in Table 2. The helmet and the FE helmet model are shown in Fig.7. A drop test of the helmet was carried out to validate the helmet FE model. In the experiment, the helmet in which a head-form impactor was embedded (Matsui Y. *et al.*, 2004) was dropped from 1.0 m height (impact velocity 4.4 m/sec), impacted to a flat anvil at parietal region as shown in Fig.8. The acceleration of the head-form impactor during the impact was measured by a three axial accelerometers (306A06, PCB Piezotronics, Inc.) which was installed in the center of mass of the impactor. Figure 9 shows the FE helmet model colliding on the anvil and the superimposition of the numerical and experimental head acceleration. The numerical result of the drop test shows good agreement with the experimental result.

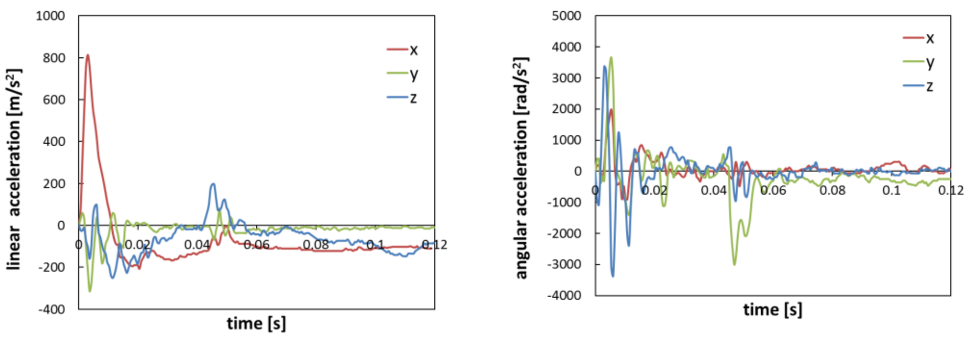

Fig.5 An example of input acceleration for the replication simulation of Hardy's experiment.

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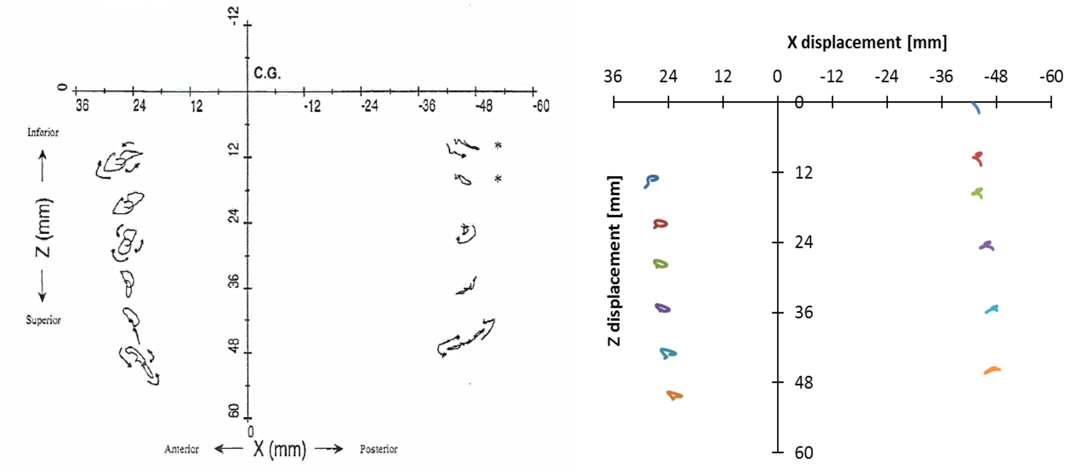
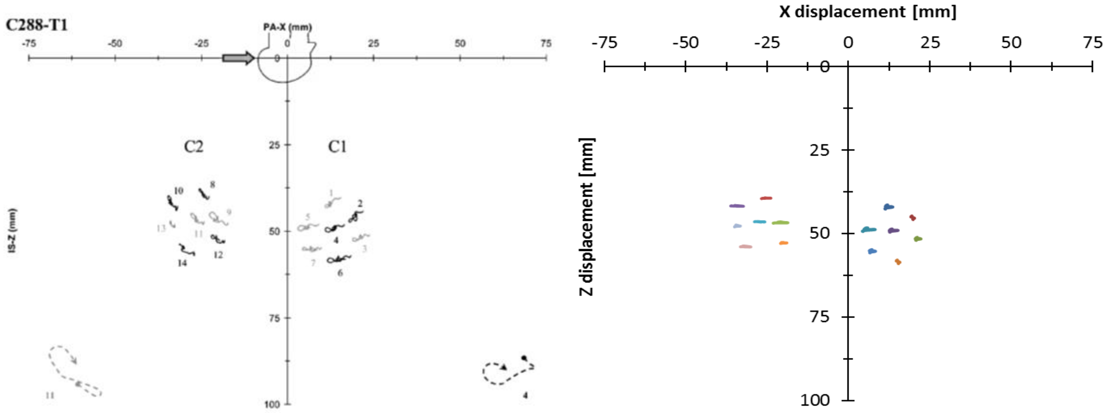
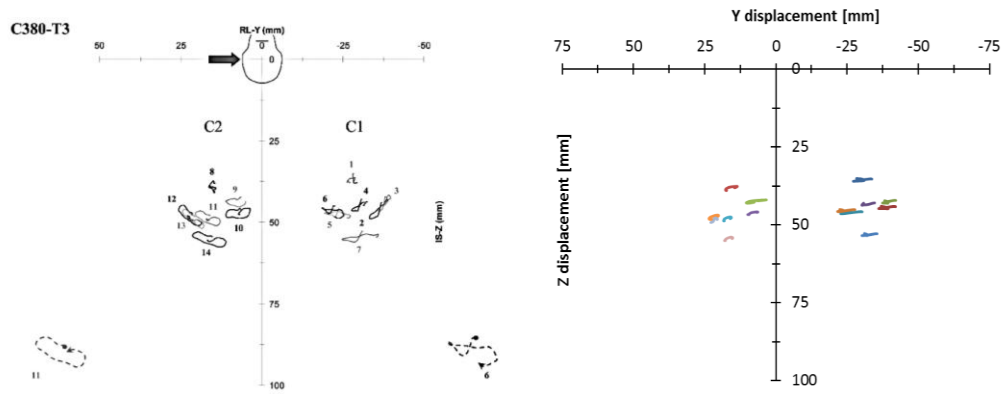


(a) Comparison of sagittal plane brain displacement between experiment (left) and simulation (right) at the NDTs location for frontal impact test C383-T3.



(b) Comparison of brain motion pattern between experiment (left) and simulation (right) at the NDTs location for occipital impact test C288-T1.



(c) Comparison of brain motion pattern between experiment (left) and simulation (right) at the NDTs location for left temporal impact test C380-T3.

Fig.6 Brain motion patterns obtained from the experiments and simulations.

(a)
(b)

Fig.7 Pictures of the helmet referred (a) and FE helmet model (b) front, top and lateral views.

Table 2 Material properties of the FE helmet model (Daring T. *et al.*, 2016).

	Outer shell	Inner form
Density ρ [kg/m^3]	1200	174.3
Young's Modulus E [MPa]	2300	0.08
Poisson’s Ratio ν [-]	0.329	0

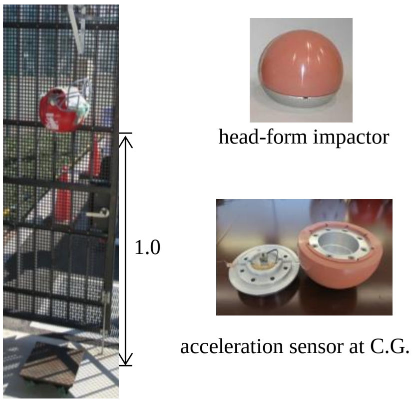


Fig.8 Drop test of the helmet with a head-form impactor.

(a)
(b)

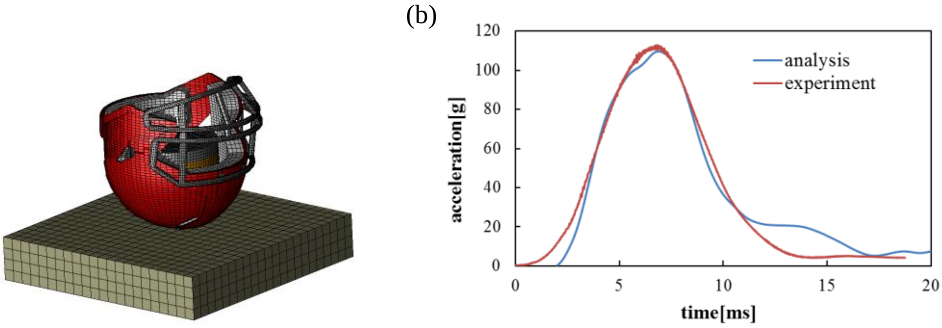


Fig.9 Comparison of the acceleration obtained from simulation and experiment, (a) configuration of parietal impact simulation, (b) superimposition of the numerical and experimental head acceleration.

3. Head-to-head impact accident cases

3.1 Accident case 1

The injured player is a defensive back player and struck the right side of his head. The videos were recorded by 2 video cameras from two directions, the end line and the sideline with 30 fps as shown in Fig.10. The sequence of the colliding motions by the game video are shown in Fig.11, the injured player is wearing red uniform with No.27. The athletic trainer of the team promptly removed the player from the game because of a suspected concussion. Due to concerns for an emergent injury, the player was taken to the emergency room, where an appropriate trauma evaluation

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based on the guideline of American Academy of Neurology 2013 was performed including a head CT, and no remarkable symptom was observed. He was then diagnosed as a concussion by the medical team doctor. The team docotor issued the permission to start the exercise when the symptoms of the player disappeared and the player passed the Cogsport test. The player was eventually instructed to return to game play after making sure that there is no expression of symptoms due to staged exercise. The detail of the time schedule until coming back to the daily exercise based on the guideline is shown in Table 3.

The motion of the players just before and after the collision was reconstructed based on the game videos by using MADYMO software. A series of video frames taking the motion of the colliding players from two directions and the corresponding colliding motion reconstructed by MADYMO are shown in Fig.11. The calculated motion of the players after collision is less precise compared with the motion before collision because the motion of the players after colllision is calculated without helmets. However the motion after collision is still important in order to obtain the consistent motion before and after the collision.

The reconstruction of the motion of the players by MADYMO was carried out as following process;

1. Taking out the frames before and after the collision from the video
2. Calculating a relative velocity between two players from two adjacent frames
   - Measuring the moving distance of the head from the 2 frames, then the velocity is obtained by dividing the distance by frame rate of the video.
   - This process is repeated 5 times and the velocity is obtained by taking the average of them.
3. Input the relative velocity as the initital velocity to human joint of MADYMO model
4. Updating the postures and the velocities of the striking and struck players until the MADYMO motions accorded with the video in each frame

Since two video cameras are not synchronized, the process mentioned above was performed by human viewing and manual work. As the background view was an athletic field specially for American football and straight lines were drawn regularly on it, the distance was able to be estimated exactly. And the kinematics of the players given by MADYMO software is practically reliable on the velocity domain if the posture calculated by MADYMO corresponds closely with the posture of the video on each frame.

The velocities of the head of each player just before collision are shown in Table 4. These results are used as the initial condition of FE analysis. In FE analysis, two helmeted head models are used as shown in Fig.12 and the peak values of translational and rotational acceleration, strain and strain rate and von Mises stress of a struck and a striking player are summarized in Table 5.

Table 3 Daily schedule of the injured player until he is allowed to come back to the exercise (accident case 1).

Date	Exercise
1st day : injured day	·loss of consciousness (-) ·loss of memory (+) ·headache (-) ·CT scan (no findings)
2nd day	·no physical symptoms ·pass Cogsport test
3rd day	·30 min. jogging
4th day	·running 40yds ×10 ·agility ·60% weight training
5th day	·position skill menu ·80%~ weight training
6th day	·normal training after medical clearance
7th day	·game play

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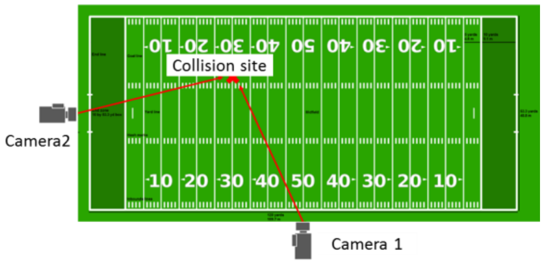

Fig.10 Location of the video cameras on the field (accident case 1).

Camera1

Camera2

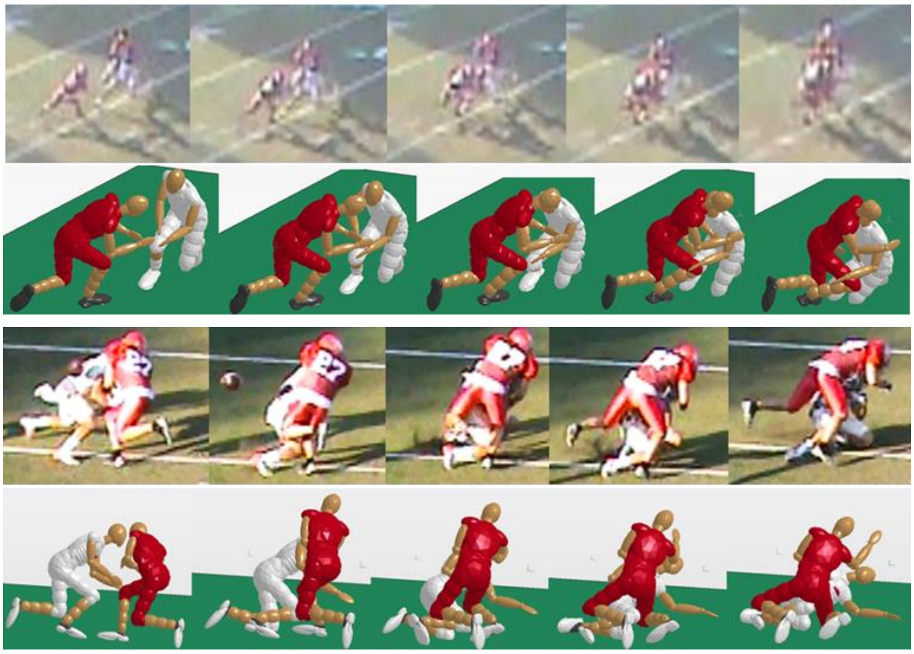

Fig.11 The collision motion of two players reconstructed based on the game video: accident case 1. The upper stillphotographs are from the game video and the lower still photographs are from MADYMO simulation.

Fig.12 FE simulation configuration of accident case 1. The player with a red helmet was injured.

Table 4 Initial condition of FE simulation just before the collision of accident case 1.

Velocity		Struck player (red)	Striking player (white)
Linear velocity [m/sec]	x	-4.2	3.6
	y	1.5	-1.3
	z	0.075	-2.4
Angular velocity [rad/sec]	x	0.35	-0.68
	y	-0.27	1.6
	z	-2.0	-2.1

Table 5 Peak head response of accident case 1. (a_{max} : maximum value of translational acceleration, α_{max} : maximum value of angular acceleration, vMS_{max} : maximum value of von Mises stress)

	a_{max} [G]	α_{max} [rad/sec^2]	Strain	Strain rate [sec^{-1}]	vMS_{max} [kPa]
Struck player (red)	150.8	5539.3	0.31	86.4	10.2
Striking player (white)	134.5	1809.2	0.12	18.5	4.26

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3.2 Accident case 2

The injured player is an offensive lineman and the collision accident occurred just after the game started. The videos were also recorded by 2 video cameras set on the end line and on the sideline with 30 fps as shown in Fig.13. The sequence of the colliding motions during the collision by the game video is shown in Fig.14, the injured player is wearing the red uniform with No.52. He was then diagnosed as a concussion and left the game. He finaly returned to play after 13 days according to the instruction as shown in Table 6. The sequence of the collision motions reconstructed by uisng MADYMO software based on the video are also shown in the Fig.14 and the velocities of the heads just before the collosion are shown in Table 7. These velocities are used as the initial condition of FE analysis. In FE analysis, two helmeted head models are used (Fig.15), the peak of these parameters for a struck and a striking player are summarized in Table 8.

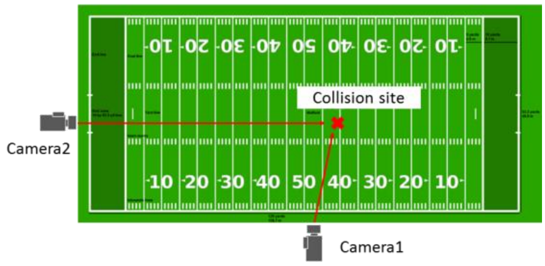
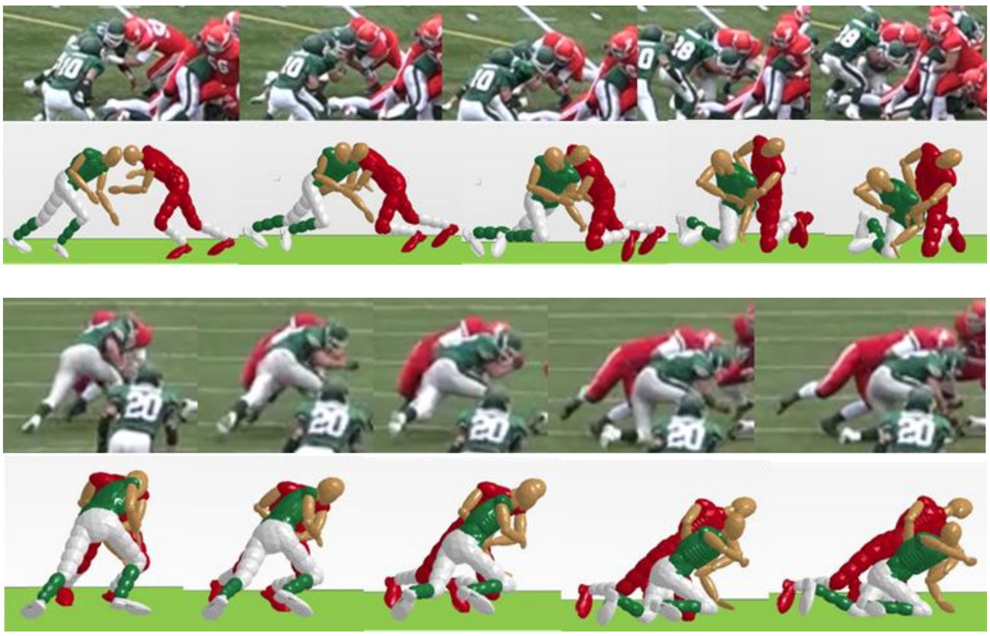


Fig.13 Location of the video cameras on the field (accident case 2).

Camera1



Camera2

Fig.14 The collision motion of two players reconstructed based on game video: accident case2. The upper stillphotographs are from the game video and the lower still photographs are from MADYMO simulation.

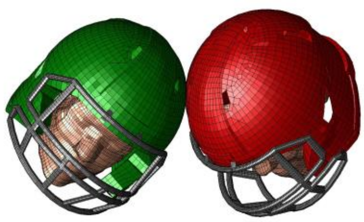


Fig.15 FE simulation configuration of accident case 2. The player with a red helmet was injured.

Table 6 Daily schedule of the injured player until he is allowed to come back to the exercise (accident case 2).

Date	Exercise
1st day : injured day	･loss of consciousness (-) ･loss of memory (-) ･headache (+) ･CT scan (no findings)
8th day	･no physical symptoms ･pass Cogsport test
9th day	･30 min. jogging
10th day	･running 40yds ×10 ･agility ･60% weight training
11th day	･position skill menu ･80%~ weight training
12th day	･normal training after medical clearance
13th day	･game play

Table 7 Initial condition of FE simulation just before the collision of accident case 2.

Velocity		Struck player (red)	Striking player (green)
Linear velocity [m/sec]	x	3.1	-5.7
	y	0.67	0.44
	z	-1.4	-0.50
Angular velocity [rad/sec]	x	-2.9	-0.021
	y	1.3	-0.0051
	z	0.90	-0.77

Table 8 Peak head response of accident case 2. (a_{max} : maximum value of translational acceleration, α_{max} : maximum value of angular acceleration, vMS_{max} : maximum value of von Mises stress)

	a_{max} [G]	α_{max} [rad/sec^2]	Strain	Strain rate [sec^{-1}]	vMS_{max} [kPa]
Struck player (red)	174.2	4888.3	0.25	66.5	9.23
Striking player (green)	165.1	5013.2	0.17	20.9	5.14

4. Discussion

4.1 Reconstruction of the motion of the players based on the video by motion analysis

In this study, a sequence of colliding motions of the football players recorded by two video cameras from two different directions was reconstructed by using whole body mathematical model. The kinematics of the players was determinded and the velocity just before the collision and the location of the collision on the head were calculated.

The velocity of the head calculated by whole body simulator is practically effective and reliable as the initial data of the FE analysis by helmeted FE human head model. In the image processing the velocity of the head can be given just for the composed velocity of the photography direction, on the contrary the velocity for every direction can be caluculated by the whole body simulation, and the response of a head by collision is calculated in more detail. These detailed data makes it possible to obtain higher precision for evaluation of TBI.

The video is recorded with 30 fps, and the interval between each frame is approximately 33 msec. Because the collision time between two heads is almost 10 msec, the momentary movement of the colliding heads may not be recorded on a video. Therefore in this study, at first we reproduced the collision movement of both players by whole body simulation by getting a continuity of the movement before and after the contact recorded on a video. Next we obtained the posture and the translational and the rotational velocities of both heads just before the contact analytically by complementing the movement of both heads before and after the contact. The result of the simulation just before and after the collision was compared with the image of the recorded video in order to evaluate the validity of the motion obtained by the whole body simulation. And the motion reconstructed by the whole body simulation almost agreed with the motion recorded on the video as shown in Fig.11 and Fig.14. As a result, it can be said that the motion reconstructed

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by the whole body simulation is validated and the motion complemented in this study can be considered acceptable.

Although the neck is not constrained in the whole body model explicitly just like a dummy model, the whole body model with the mass has the same motion as the human in the video. And its motion even after the collision did not deviate from the motion of the human in the video, though the precision may be slightly low. In this sense, it can be thought that the influence of the neck constraints is reflected on the movement of the whole body model to some extent kinematically, and the results of whole body simulation can be used as the initial condition of the FE head model.

In a FE analysis by using a FE head model, if the active response of the muscle of the neck is so sufficiently quick as within 20 msec, that is the response time of the head, the neck constraints can give the influence on a head response. From the view point of quick response, we do not have good way to measure the momentary response time of neck constraint and to reflect it to the numerical simulation. However, another result suggests the essential problem that we must not overlook. Although a struck player and striking player collided each other with almost the same velocity, the injury appears in struck player conspicuously. It is possible that the striking player strengthens the constraint of the neck by sensing a collision just before it, but at present we do not have a good way to analysis and to solve it. It seems to be a one of the most important works to be challenged in the future.

4.2 The kinematics of head model

The translational accelerations obtained from the reconstraction analyses are 150.8 g in case 1 and 174.2 g in case 2, the rotational accelerations are 5539.3 rad/s^2 in case 1 and 4888.3 rad/s^2 in case 2. Duma *et al.* (2005) reported the accelerations at the collision obtained by attaching the 6 axis acceleration sensors in the helmet as follows;

・Translational acceleration for causing concussion: 32±25 g (range, 1-200 g)

・Rotational acceleration for causing concussion: : x-axis, 905±1075 rad/s^2 (range, 1-11,348 rad/s^2)
y-axis, 2020±2042 rad/s^2 (range, 1-18,477 rad/s^2)

and Viano *et al.* (2007) reported the accelerations around 6axis by reconstructing the collision motion using Hybrid Ⅲ dummy based on the game video;

・Translational acceleration for causing concussion: 94±28 g

・Rotational acceleration for causing concussion: 6432±1813 rad/s^2

Both accelerations obtained in this study exist in the range of concussion causing zone shown by Duma *et al.* (2005) and Viano *et al.* (2007) and the results can be considered acceptable.

4.3 The risk estimation of causing concussion

Some mechanical parameters have been reported in order to estimate the risk of causing concussion. Gadd C.W. (1966) proposed Severity Index (SI). The SI is set to a limiting value of 1200 in the NOCSAE performance standard for football helmets. Head Injury Criterion (HIC) was proposed by NHTSA, in which the severity of the injury is evaluated by the integral of the maximum translational acceleration on a head. The Generalized Acceleration Model for Brain Injury Threshold (GAMBIT) was proposed by adding the rotational acceleration to the HIC (Newman *et al.*, 2000). Rotational Injury Criterion (RIC) was an adaptation of the HIC in which rotational accelerations were considered instead of the translational accelerations (Kimpara *et al.*, 2012). Brain rotational Injury Criterion (BrIC) was proposed by Takhounts *et al.* (2013) based on the peak rotational velocity in three directions. The injury risk for concussion was also estimated using translational and rotational accelerations (a and α); Duma S.M., *et al*. (2005) developed an In-helmet 6-accelerometer system in order to measure the real-time head accelerations in collegiate football players and Viano D.C. *et al*. (2007) reconstructed the helmet kinematics based on game video using Hybrid Ⅲ dummies in laboratory experiments. The risk curves based on strain, strain rate and von Mises stress calculated by FEM (Newman *et al*., 2000, Kleiven, 2007) were also reported. The injury thresholds expressed by these ten parameters and the risk functions are shown in Table 9.

In this study, the biomechanical parameters inside the skull caused by the head collision were calculated. The mechanical parameters evaluating the TBI of the injured player were calculated by using helmeted FE human head model. The values of SI, HIC, RIC, GAMBIT and BrIC of the injured players calculated based on translational and rotational accelerations are shown in Table 10. In Table 11, the probabilities to cause concussion for struck players evaluaed by multiple parameters are shown for both accident cases. The results accorded with the diagnosis of the medical team doctor. The assessment system is effective for estimating the mTBI caused by on-field head impact accident in collegiate football. The evaluating parameters in which the probability to cause concussion exceeded 50%

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in both the 2 cases are *a*, SI, HIC, GAMBIT, strain rate and vMS.

However, the probability estimated by RIC, BrIC and strain is very low in both cases. Based on the results mentioned above, the probabilities estimated by the parameters based on the translational acceleration: a, SI, HIC and GAMBIT are shown in higher values in rather translational motion dominated collision, and the probabilities estimated by strain, strain rate and vMS are shown in higher values in rather rotational motion dominated collision. These results suggest that it is necessary to evaluate the probability to cause concussion of the accident cases by multiple evaluation parameters instead of evaluating by one of the paramerts. Future, it is necessary to analyze a large number of cases by using the proposed system to establish a method for evaluating the onset of concussion under any situation.

It should be pointed out that the high probability for concussion is also expected even for the striking players in the case 1 and the case 2. However, the concussion was not reported for both striking players. The video shows that the struck players do not expect the collision at all by concentrating the ball, on the other hand, the striking players are aware of collision somewhat by concentrating the other player. Even if the same impact works on the head, the result depentds on whether the player is aware of the collision or not.

Table 9 Equations and risk functions for different mechanical parameters.

Evaluation parameter	Equation of evaluation parameter	Risk function (Newman *et al.*, 2000, Kleiven, 2007)
a [G]		$p=\dfrac{1}{1+e^{(6.057-0.007954a)}}$
α [rad/sec^2]		$p=\dfrac{1}{1+e^{(6.309-0.0009979\alpha)}}$
SI	$SI=\int_0^T a(t)^{2.5}dt$	$p=\dfrac{1}{1+e^{(3.184-0.01093SI)}}$
HIC	$HIC=\left\{(t_2-t_1)\left[\dfrac{1}{t_2-t_1}\int_{t_1}^{t_2} a(t)dt\right]^{2.5}\right\}_{\max}$	$p=\dfrac{1}{1+e^{(2.882-0.01202HIC)}}$
RIC	$RIC=\left\{(t_2-t_1)\left[\dfrac{1}{t_2-t_1}\int_{t_1}^{t_2} \alpha(t)dt\right]^{2.5}\right\}_{\max}/10^4$	$p=\dfrac{1}{1+e^{(7.036-0.00679RIC)}}$
GAMBIT	$GAMBIT=\left[\left(\dfrac{a(t)}{250}\right)^2+\left(\dfrac{\alpha(t)}{25000}\right)^2\right]^{\frac{1}{2}}$	$p=\dfrac{1}{1+e^{(6.892-17.51GAMBIT)}}$
BrIC	$BrIC=\left[\left(\dfrac{\omega_x}{\omega_{xc}}\right)^2+\left(\dfrac{\omega_y}{\omega_{yc}}\right)^2+\left(\dfrac{\omega_z}{\omega_{zc}}\right)^2\right]^{\frac{1}{2}}$	$p=1-e^{-\left(\frac{BrIC}{0.602}\right)^{2.84}}$
strain		$p=\dfrac{1}{1+e^{(3.387-9.155strain)}}$
Strain rate [sec^{-1}]		$p=\dfrac{1}{1+e^{(3.854-0.06423strainrate)}}$
vMS[kPa]		$p=\dfrac{1}{1+e^{(2.228-0.2652vMS)}}$

a(t), *α(t)* are the linear acceleration and the angular acceleration of the head respectively, *T* is the time duration of the head impact, t_1, t_2 are the time period for evaluating the maxmium value of HIC or RIC, ω_x, ω_y and ω_z are maximum angular velocities about X-, Y- and Z-axes respectively, and $\omega_{xc}=66.25\,rad/\sec$, $\omega_{yc}=56.45\,rad/\sec$ and $\omega_{zc}=42.87\,rad/\sec$ are the critical angular velocities in their respective directions.

Table 10 Head response of two concussion cases calculated based on acceleration.

Struck player (injured)	SI	HIC	RIC	GAMBIT	BrIC
Case 1	681.3	605.1	576.0	0.604	0.246
Case 2	1111.9	1001.1	410.8	0.697	0.247

Table 11 Probability of causing concussion based on different mechanical parameters for head response (%)

Struck player (injured)	a [G]	α [rad/sec^2]	SI	HIC	RIC	GAMBIT	BrIC	strain	Strain rate [sec^{-1}]	vMS [kPa]
Case 1	**99.7**	31.4	**98.6**	**98.8**	4.21	**97.6**	7.61	37.3	**84.5**	**61.5**
Case 2	**99.9**	19.3	**99.9**	**99.9**	1.41	**99.5**	7.61	25.5	**60.2**	**55.5**

The number larger than 50% is written in a bold-face.

5. Conclusion

Analyses of head injury based on game video of collision accidents in collegiate football games were performed. First, the collision motion of players was reproduced based on the video by using motion analysis, and the translational/rotational velocities, relative position and contact region of the head just before the collision were calculated. Then the obtained data were input to two FE head models wearing a helmet each as initial condition, and the damage of the head was evaluated by using the impact analysis. The finite element helmet model was validated by using a drop test of the helmet in which the head impactor was embedded. In the present study, two concussions suspected accident cases were analyzed; the concussion was evaluated using 10 mechanical parameters. The results shown that the injury risk were reached dangerous domain for the development of concussion, that consistent with the results that have been diagnosed by the team doctor. It means that the brain injury risk can be estimated by the reproduction simulation of the game video of collegiate football and FE analysis. In the future, brain injury caused by the accident can be evaluated with higher accuracy by analyzing more accident cases.

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