Finite element analysis of the effectiveness of bicycle helmets in head impacts against roads

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Abstract
Severe head injuries can occur in cyclists involved in traffic accidents. In Japan, head injuries accounted for 62% of cyclist fatalities in 2015 (ITARDA, 2016). The purpose of this study is to estimate head injuries for cyclists and quantify the effectiveness of a bicycle helmet by performing finite element (FE) simulations of head impacts against roads. Impacts with and without a helmet over a range of relative head velocities and head impact angles were simulated. A number of possible head injuries were assessed; skull fracture by skull strain, traumatic intracerebral hematoma (ICH) by brain pressure, brain contusion by brain negative-pressure and von Mises stress, and moderate and severe diffuse axonal injuries (DAIs) by von Mises stress. Results showed that without a helmet, the peak values of all metrics exceeded the 50% probability point for head injury in all impacts. The 50% probability points of moderate and severe DAIs were exceeded under impacts of 8.22 m/s at 26.5 degrees and 10.33 m/s at 15.0 degrees for moderate DAI, and 10.33 m/s at 15.0 degrees for severe DAI, without a helmet. All the peak values were reduced when a bicycle helmet was worn, and the largest reduction was found in the skull strain. These results predict that the risks of head injuries due to road impacts may be considerably decreased by helmet use.

Keywords: Bicycle helmet, Cyclist, Head injuries, Impact against road, Finite element simulation

1. Introduction

Bicycles are a popular means of transportation in the urban areas of Japan, as they are convenient for riders of all ages. Their popularity can be attributed to their ease of use and the environmental and health consciousness of Japanese people (JBCA, 2017). Experiences in the Great East Japan Earthquake in 2011 may have increased the popularity further, as the earthquake damaged much of the public transportation system in the affected area. The number of bicycles owned in Japan was approximately 7.2 million in 2013, which was the sixth largest in the world, followed by Netherlands, Germany, Denmark, Sweden and Norway (JBPI, 2013). However, the component rate of cyclist fatalities in Japan in 2014 was relatively high at 15.3%, compared with 12.2% in Sweden, 11.7% in Germany, 6.3% in England, and 4.7% in France (Cabinet Office, Government of Japan, 2016). In the urban areas of Japan, such as Tokyo and Osaka, cyclists often travel on narrow roads with low visibility (Oikawa et al., 2016a). Such situations contribute to the risk for collisions. In 2015, there were 4,117 fatalities, 38,959 serious injuries, and 627,064 minor injuries of road users in Japan, of which cyclists accounted for 14% of fatalities, 22% of serious injuries, and 14% of minor injuries (ITARDA, 2016). Cyclist fatalities were up 6% from 2014. The Japanese government began assessing the safety performance of car hoods in 2005 in an effort to reduce the number of pedestrian deaths (Matsui and Tanahashi, 2004), but there are no effective regulations for cyclist protection at this time. Because of the high rate of cyclist injuries in Japan, countermeasures for cyclist safety
should be considered.

Generally, cyclists fall to the road or ground after impact with a vehicle. In a collision, a cyclist is at risk of sustaining injuries to the head, arms, and legs, and head injuries can be especially severe (Karkhaneh et al., 2013). In cyclist fatalities, the head was the most frequently injured body region (Bambach et al., 2013; Matsui and Oikawa, 2015a). Badea-Romero and Lenard (2013) found that 69% of head injuries occurred from contact with the road or ground rather than with the striking vehicle. Oikawa et al. (2017) compared the rates of cyclist head injuries due to impact with vehicles or roads for age groups of 13–59 years, 60–64 years, 65–74 years, and over 75 years using Japanese accident data from 2009 to 2013. The results indicated that at vehicle speeds of 30 km/h or less, the percentage of cyclists with head injuries due to impact with roads was higher than those due to impact with vehicles for both serious injuries and fatalities for every age group. At vehicle speeds of 31–60 km/h, fatal injuries resulting from head impacts with vehicles were more common than those from head impacts with roads.

One of countermeasures for cyclist head injuries is a bicycle helmet (Bambach et al., 2013; Elvik, 2011; Fahlstedt et al., 2016; Ito et al., 2015; McNally and Whitehead, 2013; Matsui and Oikawa, 2015a; Minowa et al., 2000). The study by Fahlstedt et al. (2016) determined the protective effect of a bicycle helmet by quantifying the risk of brain injuries in single bicycle accident reconstructions with finite element (FE) simulations. Using a FE human head model, Ito et al. (2015) also showed that a bicycle helmet could be effectively prevent skull fractures in a collision with the A-pillar of a vehicle.

In Japan, wearing a bicycle helmet has been recommended but not required by law, and the helmet use by cyclists injured slightly, seriously and fatally by impacts with vehicles was only 8% in 2013 (Matsui and Oikawa, 2015b). For these reasons, quantifying the effectiveness of a bicycle helmet in reducing traumatic head injuries in impact accidents would significantly contribute to cyclists’ recognition of the importance of wearing head protection. Increasing helmet use among cyclists is expected to decrease cyclist fatalities.

The purpose of this study was to estimate head injuries for cyclists and to quantify the effectiveness of a bicycle helmet by performing FE simulations of head impacts against the road with and without a helmet. A range of relative head velocities and head impact angles in vehicle–cyclist collision accidents were simulated. Multiple head injury modes were assessed in the model, including the skull fracture, traumatic intracerebral hematoma (ICH), brain contusion, and diffuse axonal injury (DAI).

2. Methods

FE simulations of head impacts against roads were conducted with and without a helmet to estimate possible head injuries and quantify the effectiveness of the bicycle helmet under different impact conditions. The relative head velocities and head impact angles used in this study were taken from the results of a study using rigid body simulations by Omoda and Kono (2015). A FE human head model developed using sagittal-sectional T1-weighted MRI data of an adult male head by Aomura et al. (2016) was used in this study, since the model enabled the prediction of the intracranial response under head impact conditions. The FE human head model was used to study traumatic brain injury of helmeted American football players (Aomura et al., 2016). In the present study, we estimated the effectiveness of wearing a bicycle helmet using the FE human head model developed by Aomura et al. (2016) and a newly developed FE bicycle helmet model. To evaluate the FE bicycle helmet model, a FE headform impactor model was also developed. This study evaluated the possible risks of a skull fracture by skull strain, traumatic ICH by brain pressure, brain contusion by brain negative-pressure and von Mises stress. The risks of moderate and severe DAIs were evaluated based on the risk curve from Deck and Willinger (2008).

2.1 Materials

2.1.1 FE human head model

A FE human head model developed by Aomura et al. in 2016 consisted of the main anatomical features including the scalp, skull, cerebrospinal fluid, cerebrum, corpus callosum, ventricle, cerebellum, brain stem, falx, and tentorium, as shown in Fig. 1. The skull had the three-layer structure of the outer table, diploe, and inner table. The falx and tentorium were modeled with shell elements, and all other components were modeled with solid hexagonal elements. The FE human head model consisted of 89,226 nodes and 74,462 elements with a total mass of 4.2 kg.
2.1.2 FE headform impactor model

An FE headform impactor model was developed to evaluate the FE bicycle helmet model by comparing with experimental tests performed with a helmet placed on a headform impactor (Oikawa et al., 2016b). A headform impactor weighing 4.5 kg (Matsui and Tanahashi, 2004) represents the head of an adult human. The headform impactor was used to measure the head injury severity in impact experiments against roads. Three damped uniaxial accelerometers (Kyowa ASE-A-500) were installed at the center of gravity of the headform impactor. The headform impactor was covered by a skin simulant material made of polyvinyl chloride (PVC). The complete FE adult headform impactor model consisted of 17,502 nodes and 14,696 elements. The material properties implemented in the model are summarized in Table 1. The headform impactor and the corresponding FE model are presented in Fig. 2. The FE headform impactor model was validated by the results of a drop experiment performed with the headform impactor against road pavement (Matsui et al., 2013), shown in Fig. 3. Asphalt was selected as the material for the road pavement, as it is the material most commonly used for roads in Japan. The material properties of asphalt (Abe et al., 2006) are shown in Table 2. In the experiment, the headform impactor was dropped from 1.50 m height, which produced an impact velocity of 5.42 m/s. Comparison of the simulation and experiment is presented in Fig. 4, which shows good agreement.

![Fig. 1 FE human head model composed of the scalp, skull, cerebrospinal fluid, cerebrum, corpus callosum, ventricle, cerebellum, brain stem, flax, and tentorium (Aomura et al., 2016).](image1)

![Fig. 2 Picture of a headform impactor and the FE headform impactor model.](image2)

![Fig. 3 Headform impactor drop tests against the road (Matsui et al., 2013).](image3)

![Fig. 4 The resultant acceleration of the headform impactor obtained from the simulation and experiment.](image4)

### Table 1 Material properties of the FE headform impactor model.

| Material   | Density $\rho$ (kg/m$^3$) | Young’s modulus $E$ (MPa) | Poisson’s ratio |
|------------|---------------------------|--------------------------|----------------|
| Sphere     | 2700                       | 70000                    | 0.33           |
| Baseplate  | 7870                       | 192080                   | 0.30           |

### Table 2 Material properties of the pavement (Abe et al., 2006).

| Material | Density $\rho$ (kg/m$^3$) | Young’s modulus $E$ (MPa) | Poisson’s ratio |
|----------|---------------------------|--------------------------|----------------|
| Asphalt  | 250                        | 12000                    | 0.35           |
2.1.3 FE bicycle helmet model

A FE bicycle helmet model was developed based on a commercial bicycle helmet. The geometry of the helmet was obtained by a computed tomography (CT) scan of the FIGO G-1 helmet (OGK Kabuto, 2012), which was a bicycle helmet for an adult male. The helmet model consisted of a polycarbonate outer shell and a polystyrene form liner (Milne et al., 2013), which are the components in the FIGO G-1 helmet. The outer shell was modeled with shell elements with a constant thickness of 1.0 mm. The liner was modeled with solid hexagonal elements, which were obtained by extrusion of the outer shell. The material property values for the polystyrene form liner and the polycarbonate outer shell are shown in Table 3. The complete FE helmet model consisted of 5,252 nodes and 4,302 elements. The bicycle helmet and the corresponding FE model are presented in Fig. 5. The FE bicycle helmet model was validated by comparison with a drop experiment of the helmet against road pavement (Oikawa et al., 2016b). The FE headform impactor model was coupled with the helmet model to match the experiment. In the experiment, the headform impactor with the helmet was dropped from 1.50 m height onto the road pavement, which created an impact velocity of 5.42 m/s. The required angle of intersection between the bottom surface of the headform impactor base plate and the horizontal line was determined to be 63.0 degrees so that the side of the helmet could impact the road, as shown in Fig. 6, considering that the most frequent impact locations in helmeted-cyclist accidents were on the side and front of the helmet (Ching et al., 1997). Comparison of the resultant acceleration in the simulation and experiment is presented in Fig. 7, which shows good agreement.

| Table 3 Material properties of the FE bicycle helmet model. |
|-----------------|-----------------|----------------------|
| Density $\rho$ (kg/m$^3$) & Young's modulus $E$ (MPa) & Poisson's ratio $\nu$ |
| Inner liner & 187 & 60 |
| Outer shell & 909 & 1500 & 0.42 |

Fig. 5 Top and lateral views of the bicycle helmet (a) and FE helmet model (b).

Fig. 6 Drop tests against the road for the headform impactor with the helmet (Oikawa et al., 2016b).

2.2 Impact conditions

The horizontal, vertical and relative head impact velocities, head impact angles, and impact locations in the current study were derived from the study by Omoda and Konosu (2015), which performed simulations using Mathematical Dynamic Model (MADYMO) (TASS, 2010) to estimate the head impact conditions on the road for a cyclist impacted by a sedan or a box van. This study simulated head impacts with the road assuming that the cyclist collided with a sedan (Case A) and a box van (Case B) at vehicle impact velocities of (1) 20 km/h (5.6 m/s), (2) 30 km/h (8.3 m/s), and (3) 40 km/h (11.1 m/s) (Omoda and Konosu, 2015), shown in Table 4. Six combinations of relative head velocities and head impact angles were simulated. The sides of the head and helmet were selected as the impact location. All simulations were performed using the FE human head model with and without the FE helmet model with the LS-DYNA (version 8.0) commercial finite element software (LSTC, 2015). The road was assumed to be a rigid surface. The friction coefficient between the head or helmet and the road was set to 0.5, as was the friction coefficient between the head and helmet (Fahlstedt et al., 2016). Fig. 8 shows the impact simulation configuration for Case A–(1) with and without a helmet.
2.3 Head injury risk assessment

The head impact simulations were performed to evaluate the possible risks of a skull fracture, traumatic ICH, brain contusion, and DAI in a head with and without a helmet in accordance with the injury criteria. A skull fracture suggests that significant force was involved in the injury. Patients with any fracture associated with neurologic impairment are at increased risk of intracranial hematomas. Bambach et al. (2013) reported that cyclists with skull fractures had the highest proportion of individuals with serious injury (52.5%). ICHs are collections of blood within the brain that result from coalescence of contusions in traumatic situations. The exact point at which one or more contusions become a hematoma is not well defined (MSD, 2017). Brain contusions can impair a wide range of brain functions, the severity of which depends on contusion size and location. Larger contusions may cause brain edema and increase intracranial pressure (MSD, 2017). DAI causes unconsciousness for 6 hours or more and may lead to serious higher brain dysfunction repercussions or death. DAI is difficult to detect with CT or magnetic resonance imaging (MRI) (Azouvi, 2000; Fork et al., 2005).

The head injury risks were estimated by deformation for a skull fracture (Irwin and Mertz, 1997), the pressure in the brain for traumatic ICH (Gross 1958; Gourin and Shackford, 1997; Shreiber et al., 1997), the negative pressure (Gross, 1958; Gourin and Shackford, 1997; Shreiber et al., 1997) and von Mises stress (Miller et al., 1998) for brain contusion. The risks of moderate and severe DAIs were evaluated by von Mises stress (Miller et al., 1998) based on the risk curve from Deck and Willinger (2008). The head injury criteria are shown in Table 5.

Table 5 Criteria for 50% probability of head injury.

| Head injuries         | Injury risk predictors               | Criteria                  |
|-----------------------|-------------------------------------|---------------------------|
| Skull fracture        | Skull deformation                   | 2.1 (%)*1                 |
| Traumatic ICH         | Brain maximum pressure              | 300 (kPa)*2               |
| Brain contusion       | Brain minimum negative pressure     | -100 (kPa)*3              |
| Brain contusion       | Brain maximum von Mises stress      | 8.6 (kPa)*4               |

*1: Irwin and Mertz, 1997  
*2: Gross 1958, Gourin and Shackford 1997, Shreiber et al. 1997  
*3: Gross 1958, Gourin and Shackford 1997, Shreiber et al. 1997  
*4: Miller et al. 1998

3. Results

Table 6 shows the peak mechanical responses computed with the FE human head model with and without the FE bicycle helmet model. Without a helmet, the peak values of skull strain ranged from 3.4% to 4.9% in Case A and from 3.3% to 4.8% in Case B. With the helmet, the peak values of skull strain ranged from 0.1% to 0.3% in Case A and from 0.2% to 0.3% in Case B, which indicated a 94%–97% reduction in Case A and a 92%–94% reduction in Case B. Every impact without a helmet surpassed the skull fracture 50% probability threshold of 2.1% skull strain (McCalden, 1993), but every impact with the helmet was far below the threshold.

For traumatic ICH, the peak values of brain pressure in the head without a helmet ranged from 1207 kPa to 1974 kPa in Case A and from 1235 kPa to 2203 kPa in Case B. With a helmet, the peak values of brain pressure ranged from 826
kPa to 1514 kPa in Case A and from 851 kPa to 1145 kPa in Case B, which indicated a 23%–46% reduction in Case A and a 25%–50% reduction in Case B. Every impact with and without a helmet surpassed the traumatic ICH 50% probability threshold of 300 kPa (Gross, 1958; Gourin and Shackford, 1997; Shreiber et al., 1997).

The risk of brain contusion was assessed by the negative pressure and von Mises stress. The peak values of negative-pressure in the head without a helmet ranged from -1146 kPa to -1879 kPa in Case A and from -1084 kPa to -1705 kPa for Case B. With a helmet, the peak values ranged from -688 kPa to -1168 kPa in Case A and from -709 kPa to -933 kPa in Case B, which indicated a 33%–53% reduction in Case A and a 25%–45% reduction in Case B. Every impact with and without a helmet surpassed the brain contusion 50% probability threshold of -100 kPa (Gross, 1958; Gourin and Shackford, 1997; Shreiber et al., 1997). The peak values of von Mises stress in the head without a helmet ranged from 13.6 kPa to 27.5 kPa in Case A and from 14.0 kPa to 33.3 kPa in Case B. With a helmet, the peak values ranged from 9.9 kPa to 23.7 kPa in Case A and from 10.4 kPa to 19.7 kPa in Case B, which indicated a 14%–27% reduction in Case A and a 26%–46% reduction in Case B. Every impact with and without a helmet surpassed the brain contusion 50% probability threshold of 8.6 kPa (Miller et al., 1998).

Risks of moderate and severe DAIs with and without a helmet were derived from the risk curve presented by Deck and Willinger (2008). The results for moderate and severe DAIs are shown in Fig. 9(a) and Fig. 9(b), respectively. Case A–(1), (2), and (3) resulted in a 1%, 3%, and 72% probability of moderate DAI without a helmet and 1%, 1%, and 30% with a helmet, respectively. Case B–(1), (2), and (3) resulted in a 1%, 29%, and 95% probability of moderate DAI with a helmet and 1%, 1%, and 7% with a helmet, respectively. Case A–(1), (2), and (3) resulted in a 1%, 1%, and 18% probability of severe DAI without a helmet and a 1%, 1%, and 7% with a helmet, respectively. Case B–(1), (2), and (3) resulted in a 1%, 6%, and 51% probability of severe DAI without a helmet and a 1%, 1%, and 2% with a helmet, respectively. The results predict the effectiveness of wearing a helmet for reducing the probability of both moderate and severe DAIs.

![Fig. 9 Risks of moderate and severe DAIs with and without a helmet.](image)

4. Discussion

This study conducted FE simulations of head impacts on a road to estimate possible head injuries and quantify the effectiveness of a bicycle helmet. Six cases covering a range of relative head velocities and head impact angles were simulated to evaluate the risks of a skull fracture, traumatic ICH, brain contusion, and moderate and severe DAIs in a head with and without a helmet. In every case, wearing a helmet reduced the probability of head injury. In particular, the reduction in the peak skull strain used to estimate the skull fracture (92%–97%) was much larger than that for other head injuries. A study reported that skull fracture (25%) was the most frequent head injury, followed by subarachnoid bleeding (16%), and subdural hematoma (12%) in cyclists involved in traffic accidents (Oikawa et al., 2016b). Bambach et al. (2013) indicated that head injury risk reduction was significantly associated with helmet use, with a reduction up to 74% overall, including a 78% reduction in skull fracture, 72% reduction in intracranial injury, 74% reduction in concussive injury, and 80% reduction in open head wounds. If helmet use can be made common for cyclists by increasing their awareness of the helmet’s effectiveness, the incidence of head injuries can be significantly reduced.
Table 6  Comparison of peak values computed with the FE human head model with and without the FE helmet model.

| Case   | Case No. | $V_{rel}$ (m/s) | $\alpha$ (°) | No helmet | With a helmet | Reduction (%) |
|--------|----------|-----------------|--------------|-----------|---------------|---------------|
|        |          |                 |              | (a)       | (b)           | {(a) – (b)} / (a) * 100 |
| Peak skull strain (%) for skull fracture | | | | | | |
| Case A | A–(1)    | 3.22_36.0       | 3.4*         | 0.1       | 3.3           | 97            |
|        | A–(2)    | 4.75_36.4       | 4.3*         | 0.2       | 4.1           | 94            |
|        | A–(3)    | 8.22_26.5       | 4.9*         | 0.3       | 4.6           | 94            |
| Case B | B–(1)    | 4.33_46.4       | 3.3*         | 0.2       | 3.1           | 94            |
|        | B–(2)    | 6.83_15.9       | 3.7*         | 0.3       | 3.4           | 92            |
|        | B–(3)    | 10.33_15.0      | 4.8*         | 0.3       | 4.5           | 94            |
| Peak brain pressure (kPa) for traumatic intracerebral hematoma | | | | | | |
| Case A | A–(1)    | 3.22_36.0       | 1207*        | 826*      | 381           | 32            |
|        | A–(2)    | 4.75_36.4       | 1931*        | 1052*     | 879           | 46            |
|        | A–(3)    | 8.22_26.5       | 1974*        | 1514*     | 460           | 23            |
| Case B | B–(1)    | 4.33_46.4       | 1235*        | 932*      | 303           | 25            |
|        | B–(2)    | 6.83_15.9       | 1691*        | 851*      | 840           | 50            |
|        | B–(3)    | 10.33_15.0      | 2203*        | 1145*     | 1058          | 48            |
| Peak brain negative-pressure (kPa) for brain contusion | | | | | | |
| Case A | A–(1)    | 3.22_36.0       | -1146*       | -688*     | -458          | 40            |
|        | A–(2)    | 4.75_36.4       | -1879*       | -879*     | -1000         | 53            |
|        | A–(3)    | 8.22_26.5       | -1748*       | -1168*    | -580          | 33            |
| Case B | B–(1)    | 4.33_46.4       | -1084*       | -810*     | -274          | 25            |
|        | B–(2)    | 6.83_15.9       | -1245*       | -709*     | -536          | 43            |
|        | B–(3)    | 10.33_15.0      | -1705*       | -933*     | -772          | 45            |
| Peak brain von Mises stress (kPa) for brain contusion | | | | | | |
| Case A | A–(1)    | 3.22_36.0       | 13.6*        | 9.9*      | 3.7           | 27            |
|        | A–(2)    | 4.75_36.4       | 17.3*        | 13.4*     | 3.9           | 23            |
|        | A–(3)    | 8.22_26.5       | 27.5*        | 23.7*     | 3.8           | 14            |
| Case B | B–(1)    | 4.33_46.4       | 14.0*        | 10.4*     | 3.6           | 26            |
|        | B–(2)    | 6.83_15.9       | 23.5*        | 12.7*     | 10.8          | 46            |
|        | B–(3)    | 10.33_15.0      | 33.3*        | 19.7*     | 13.6          | 41            |

* Head injury criteria at 50% probability obtained
While reductions in peak values were achieved in all the cases when a helmet was used, the tolerance limits for traumatic ICH and brain contusion were still exceeded. The peak values of the skull strain in the helmeted cases were the only outputs that fell below the injury threshold. Therefore, the commercial bicycle helmet might be effective for preventing skull fractures but not internal brain injuries (Deck and Willinger, 2008; Fahlstedt et al., 2016; Kurt et al., 2016). This might be related to performance tests performed on bicycle helmets, which were tested only under linear acceleration in which helmets were dropped from 1.50 m (5.42 m/s) and 1.06 m (4.57 m/s) heights (JHMA, JIS T 8134, 2007), and from 1.72 m (5.80 m/s) and 1.17 m (4.79 m/s) heights (CPSA, 2017). Helmet design improvements, such as an airbag with the potential for preventing cyclists from traumatic brain injuries, may be required (Kurt et al., 2016).

As an injury assessment criterion, the Head Injury Criterion (HIC) has been widely accepted to assess the severity of potential head injuries in road accidents. For example, pedestrian head protection performance tests for vehicle hoods have been evaluated by the HIC (Matsui, 2011). When Oikawa et al. (2016b) conducted impact experiments against the A-pillar of a vehicle and a road using a headform impactor with and without a helmet, the helmeted cases resulted in a 60% HIC reduction for impacts against a vehicle and an 86% HIC reduction for impacts against the road. Considering the results obtained from this study, those reductions would correlate with skull fractures, not brain injuries. HIC is related to the Wayne State Tolerance Curve (WSTC), which was developed from the early research work that determined the tolerance level of the head to skull fracture loads and its relationship to brain injury (Eppinger et al., 1999). HIC measures the head injury risks due to linear head acceleration. Therefore, it might not be able to fully assess brain injuries. Newman (1986) and Chamouard et al. (1986) revealed that severe skull fractures could occur even when small HIC values were obtained. For bicycle helmet impacts, Ching et al. (1997) showed that the most common location for helmet damage was in the front of the helmet (nearly 50%), followed by the sides (nearly 15% on each side), which could cause both linear and rotational head accelerations (Milne et al., 2013). Assessing the risks of head injuries beyond skull fractures for such complex loading conditions would require the use of HIC and other alternative parameters (Deck and Willinger, 2008; Aomura et al., 2016), such as deformation, pressure, negative pressure, and von Mises stress in the brain.

The limitation of the present study was that the FE human head model (Aomura et al., 2016) did not incorporate material properties of the head of older people, even though the Young's modulus of the skull was set to be 8750 MPa of a 65-year-old human (Irwin and Mertz, 1997). Analyses using actual accident data by Matsui and Oikawa (2015a) showed that at vehicle speeds of 30 km/h or less, there was a greater percentage of fatal head injuries in cyclists aged 60 years or older caused by head impacts against roads than against vehicles; 64% by impacts with roads in sedan–bicycle accidents and 81% in box van–bicycle accidents. In contrast, at vehicle speeds of 31–60 km/h, fatal head injuries by head impacts with vehicles were more common for cyclists aged 60 years or older than those resulting from head impacts with roads; 62% by impacts with vehicles in sedan–bicycle accidents and 63% in box van–bicycle accidents. This study simulated head impacts with the road at a range of relative head velocities and head impact angles, assuming that the cyclist collided with a sedan and a box van at vehicle impact velocities of 20 km/h (5.6 m/s), 30 km/h (8.3 m/s), and 40 km/h (11.1 m/s). In all the cases simulated without a helmet, the tolerance limit for 50% probability was exceeded for all the head injuries. However, it is not clear if the results are applicable to older people. One study investigated the head injury mechanism using age-specific human head/brain FE models, and compared the difference between models representing humans in their 20s and 70s (Yanaoka and Dokko, 2014). The age-specific characteristics in their models included the shape, size, and stiffness of the brain matter and blood vessels. Therefore, further simulations should be conducted when the FE human head model used in the current study is modified to have material properties of older people by considering such age-specific characteristics.

This study evaluated the risks of moderate and severe DAIs based on the risk curve from Deck and Willinger (2008). Based on the Abbreviated Injury Scale (AIS), moderate DAI is classified as $2 \leq AIS \leq 3$, and severe DAI is classified as $AIS \geq 4$. The FE simulations showed the effectiveness of wearing a helmet for reducing both moderate and severe DAIs. By wearing a helmet, the moderate DAI risk was reduced from 72% in Case A–(3) and 95% in Case B–(3) to 30% and 7%, respectively, and the severe DAI risk was reduced from 51% in Case B–(3) to 2%. In real world collisions involving cyclists, head injuries occur from the combination of translational and rotational acceleration (Deck and Willinger, 2008). DAI is sustained when the inertial force, generated by forced rotational movement, is applied to the head at the time of the impact (Maxwell et al., 1993; Gennarelli et al., 1998). In this study, impact conditions contained no rotational acceleration, so additional research is required to fully quantify the effectiveness of wearing a helmet for reducing traumatic brain injuries including DAI.
5. Conclusion

FE simulations of head impacts against roads were conducted with and without a bicycle helmet to estimate possible head injuries and quantify the effectiveness of the bicycle helmet under different impact conditions. In this work, a FE bicycle helmet model was developed. The head injury risks were evaluated by skull strain for a skull fracture, brain pressure for traumatic ICH, brain negative-pressure and von Mises stress for brain contusion, and von Mises stress for DAI. The results showed that without a helmet, the peak values of skull strain, brain pressure, brain negative-pressure, and von Mises stress exceeded the threshold values corresponding to 50% injury probability. The 50% probability points of moderate and severe DAI s were exceeded under impacts of 8.22 m/s at 26.5 degrees and 10.33 m/s at 15.0 degrees for moderate DAI, and 10.33 m/s at 15.0 degrees for severe DAI, without a helmet. While all the peak values were reduced in the helmeted cases, the skull strain was reduced the most. These results predict that the risks of head injuries due to road impact may be considerably decreased by helmet use.

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