INVESTIGATION ON EFFECTS OF WHOLE-BODY KINEMATICS DURING COLLISION ON PEDESTRIAN INJURIES

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ABSTRACT

Recently, pedestrian safety performance of vehicles has been improved by the modification of regulations and new car assessment programs (NCAPs). In particular, safety performance of the bonnet has been improved in terms of head protection by reducing HIC. According to the accident statistics, however, pedestrian fatalities account for a high percentage, and the causes of death include not only head injury but also thoracic and pelvis injuries. Therefore, pedestrian protection technologies need to include protection of these body regions in addition to the head. In order to reduce the number of pedestrian fatalities, this study aimed to investigate the effect of the whole-body kinematics on injury reductions of pedestrians.

In a collision between a bonnet-type vehicle and a crossing pedestrian, the whole-body moves in a chain reaction starting from the input to the legs, subsequently transmitted to the pelvis, the thorax, and the head. Therefore, it is expected that controlled pedestrian kinematics from the time of collision will have an effect on the injury to various body regions. In this study, the GHBMC 50th percentile male model and the vehicle model with general bonnet type was used to simulate car-pedestrian collisions. A model composed of spring and shell elements was affixed to the vehicle model to apply controlled loads to the center of gravity of the pedestrian model by changing the stiffness characteristics of the model, and the relationship between the whole-body kinematics of the pedestrian model and the injury values was investigated at a collision speed of 40 km/h.

The results confirmed that the angular velocity of the upper body around the center of gravity was reduced by the early input to the pedestrian pelvis, effectively reducing thoracic input and the head injury value.

Input to the pelvis depends on the input through the legs and the external force from the vehicle. Since the vehicle used in this study had a low bonnet height, there was little external force from the vehicle to the pelvic region, potentially diminishing the effect of restraining the center of gravity. Since this study used a specific collision speed and a pedestrian size, it is necessary to consider the influence of these factors in a future study.

This study clarified that pedestrian kinematics control technology may be one of the effective measures to further reduce pedestrian fatalities.

INTRODUCTION

The number of Traffic accident fatalities in Japan is are trending downward, fatalities in 2021 decrease by about 40% compared to that in 2011 [1]. However, the number of pedestrian fatalities accounted for 35% of overall fatalities in 2021, this rate has been continued to be the highest among the fatality groups for over 10 years. In terms of evaluation of pedestrian safety performance of vehicles, the head (HIC) and lower limb(the bending moment of the bones and the elongation of knee ligaments) injury evaluation using subsystem impactors are performed. Accidents in the real world include not only these injuries, but also brain injuries due to head rotation during collisions, thoracic injuries and pelvic injuries. However, current sub-system impactor is incapable of evaluating brain injury caused by head rotation and in addition, there is no evaluation method for thoracic and pelvic injuries. In many pedestrian accidents, the collision phenomenon starts with the input from the vehicle to the pedestrian's legs, upper body rotates toward the vehicle due to occur movement of the waist close to the pedestrian's center of gravity. Rotation of the upper body becomes rotational energy that causes thoracic injuries, and furthermore, it becomes a factor that causes brain injury due to head rotation. Finally, the head injury occurs when the head collides with the vehicle. In a pedestrian accident, the injury region changes over time due to the input from the vehicle and the pedestrian whole-body kinematics. (Figure 1). Thus, the evaluation by entire body of a pedestrian is useful for investigating the mechanism of real-world pedestrian accidents. Nakamura et al. [2] investigated the effects of the input to the body regions on the whole-body trajectories and the brain injury by using the dummy model and a production car model with the restraint surface connected to the center of gravity of the vehicle model by the spring element. As the results of the study, it was found that the input to the pelvis influenced to the kinematics of the whole-body pedestrian and the brain injury. But, in order to reduce the number of pedestrian fatalities, it is also necessary to investigate the injury measures of the thorax and the pelvis. Since the pelvis of the dummy was rigid, the evaluation by using a human body model is needed to investigate in detail. In order to reduce the number of pedestrian fatalities, this study aimed to investigate the effects of the whole-body kinematics on the injury reductions of pedestrians.

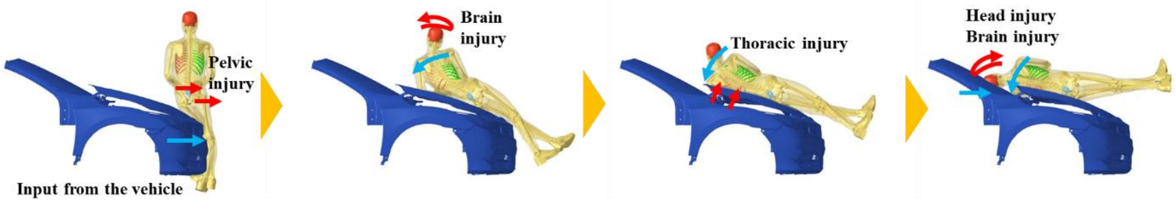

Figure 1. Pedestrian Collision Flow

METHODS

Verification of effects from waist restraint close to the pedestrian’s center of gravity

In this study, a pedestrian human body model (HBM) capable of studying the whole-body kinematics was used. For verification, the 50th percentile male model of the simplified Global Human Body Models Consortium (GHBMC) ver1.4 pedestrian model. All simulations were performed by the LS-Dyna R9.2.0. As the pedestrian kinematics is thought to be influenced by depend on the restraint of the center of gravity, a sedan type vehicle model which exerts little force on the waist close to the center of gravity was selected in this study. In order to change pedestrian whole-body kinematics, a waist restraint surface was added to a standard sedan model vehicle. The restraining surface and vehicle were connected by a spring element with load-displacement properties. (Figure 2) (Table 1) The effect of changing pedestrian kinematics was verified by comparing models with and without waist restraint surface. Vehicle models with different restraint characteristics were made to collide with the pedestrian model at 40 km/h from the left side.

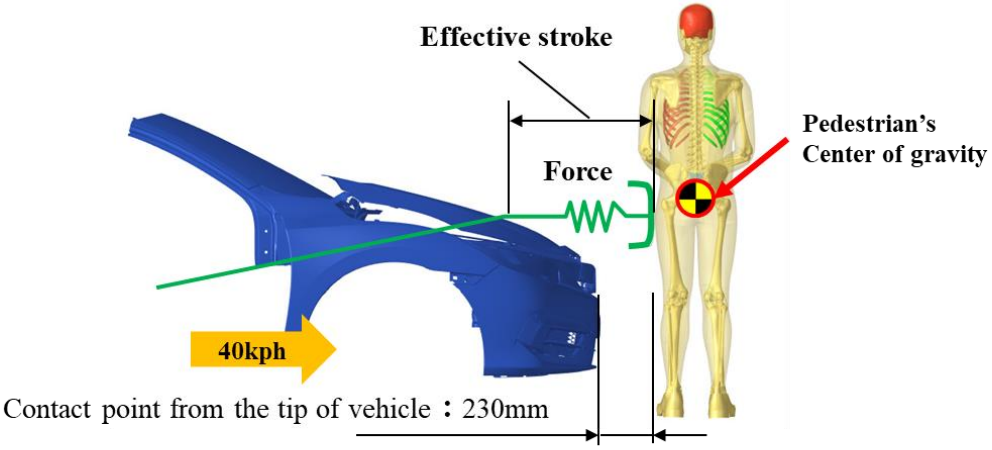

Figure 2. Structure of vehicle model and evaluation method of pedestrian kinematics

Table 1.
Restraint characteristics

	Spring characteristic		
	Force (N)	Effective stroke (mm)	Energy (J)
Original	N/A	N/A	N/A
Case (a)	2500	390	731
Case (b)	3000	390	878
Case (c)	3500	390	1024
Case (d)	4000	390	1170
Case (e)	4500	390	1316

Evaluation items

The effects of changing pedestrian kinematics were evaluated by the trajectory of the pedestrian, angular velocity around the pedestrian’s center of gravity and injury criteria. Trajectories of the pedestrian were measured at the Head, T4, T8 and Pelvis relative to the vehicle models. The angular and translational velocity of the pedestrian's center of gravity were calculated based on the relative displacement of the waist and the T1 displacement in lateral and vertical directions. HIC was used as the injury criteria for the head injury. The brain injury was evaluated by using Convolution of Impulse Response for Brain Injury Criterion (CIBIC) developed by Takahashi et al. [3] In the past study, the multiple peaks of CIBIC were seen in both pre-impact (head swing) and impact (the head collides to the vehicle) phase in the simulation of the collision between the AM 50th percentile male pedestrian model and the vehicle models [4]. Thus, CIBIC was also evaluated in both pre-impact and impact phases in this study. In addition, in pre-impact phase, the relationship between CIBIC in pre-impact and the angular velocity around the pedestrian's center of gravity was calculated in order to investigate the effect of the angular velocity on CIBIC. In the impact phase, the relationship between the injury measures (HIC and CIBIC) and the impact velocity of the head were also investigated. The head impact velocity was calculated by the velocity of the head in x, y and z direction relative to the vehicle model.

The input to thorax was evaluated by the acceleration of the ribs of the left side. Pelvic injury was evaluated by the sacrum force and pubis force measured at the sagittal plane. Evaluation items are described in Figure 3.

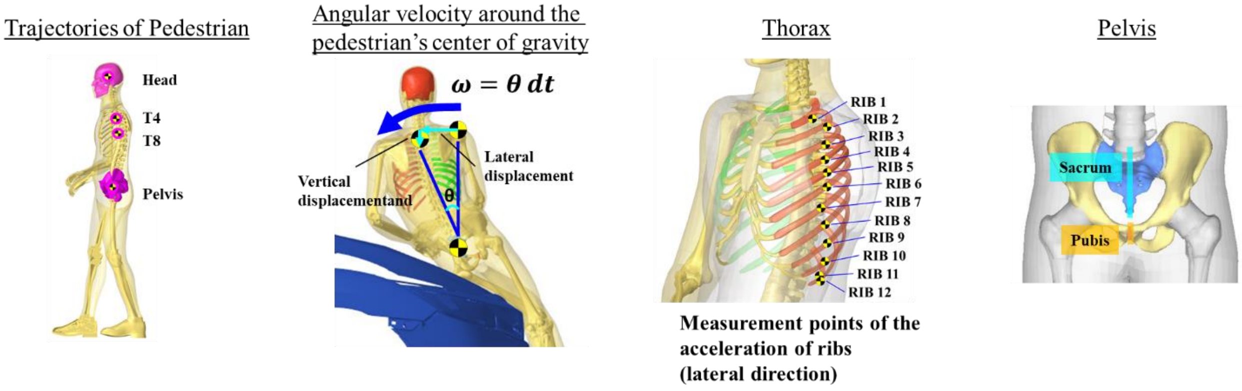

Figure 3. Evaluation items

RESULTS

Comparison of trajectories, angular and translational velocity

Figure 4 shows the trajectories of the landmarks (the head, T4, T8 and pelvis) of the pedestrian human models collided with the restraint characteristic vehicle models relative to the vehicle models. The wrap around distances of the head, T4, T8, and pelvis were decreased with increasing the input force to the waist of the pedestrian model. To analyze the pedestrian kinematics, figure 5 and 6 show the time histories of the angular velocity and translational velocity of the center of gravity of the pedestrian, respectively. The angular velocity at around 50msec increased and the angular velocity at around 100msec decreased as the input to the waist from the vehicle increased (Figure 5). The time of the beginning of increasing and the peak values of the translational velocity showed almost the same tendency as those of the time histories of the angular velocity (Figure 6).

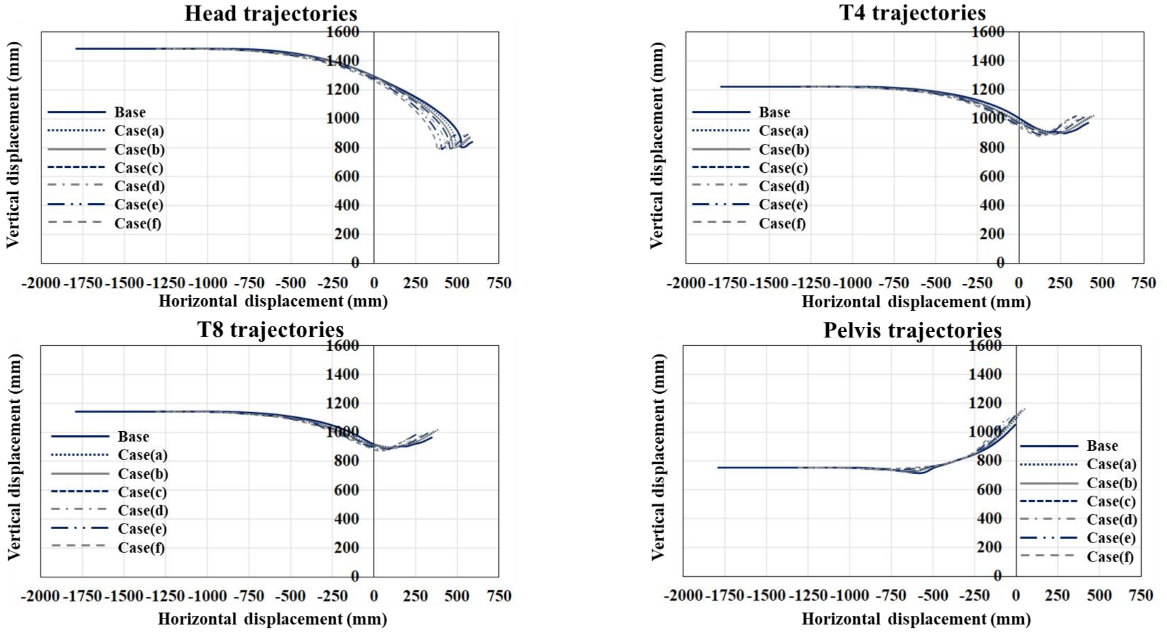

Figure 4. Trajectories of pedestrian with different restraint characteristics

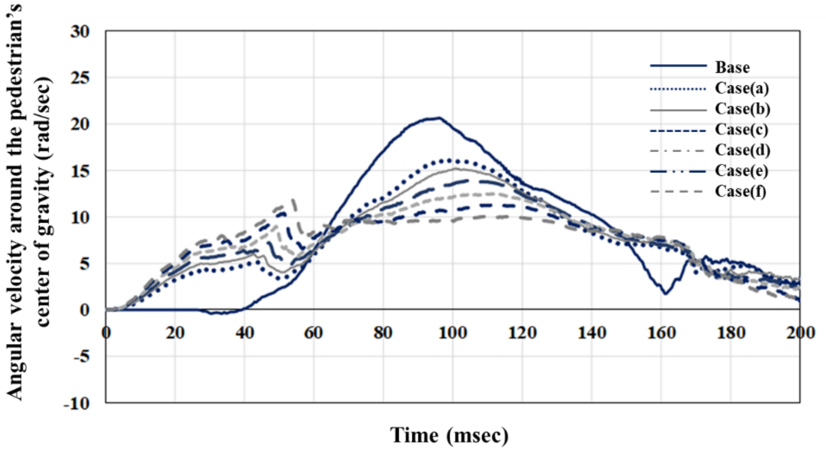

Figure 5. Time history of Angular velocity around the pedestrian's center of gravity

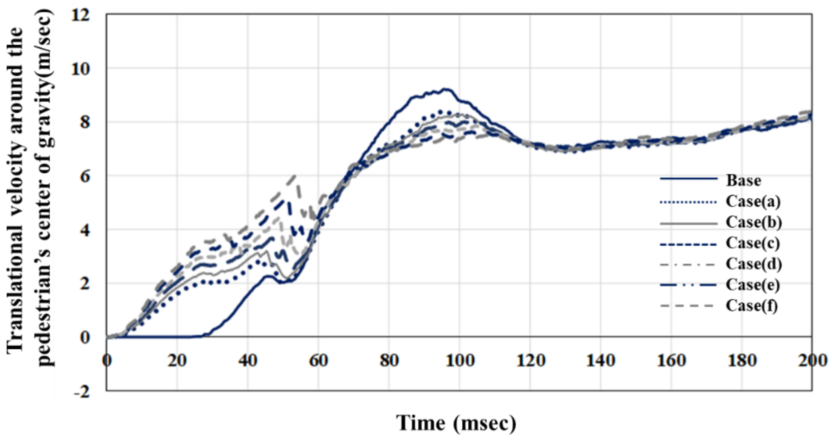

Figure 6. Time history of translational velocity around the pedestrian's center of gravity

Comparison of injury levels

Figure 7 shows the time histories of CIBIC calculated in all cases. The peak value of the strain in the brain occurred during both the pre-impact and impact phase in all cases.

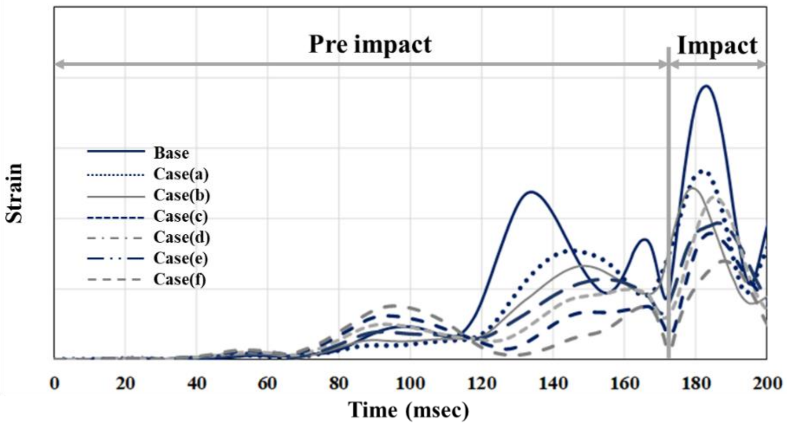

Figure 7. Comparison of CIBIC time histories

Figure 8 and 9 show the comparison of CIBIC in pre-impact by normalizing these values by that of the base model and the correlation between normalized CIBIC in pre-impact and the peak value of the angular velocity of the center of the gravity of the pedestrian model, respectively. Normalized CIBIC in pre-impact of case (a) was decreased to 64%. Normalized CIBICs in pre-impact were decreased with increasing the input force to the waist of the pedestrian model, showing the minimum value in case (e) (Figure 8). That of case (f) was equivalent to that of case (e). The coefficient of determination between normalized CIBIC in pre-impact and the peak value of the angular velocity was 0.9913 (Figure 9).

Figure 10 and 11 show the comparison of CIBIC in impact by normalizing these values by that of the base model and the correlation between normalized CIBIC in impact and the head impact velocity, respectively. Normalized CIBICs in impact of case (a), (b), (c) and (d) were decreased to about 50% (Figure 10). In addition, they were further decreased from case (d) to case (f). The coefficient of determination between normalized CIBIC in-impact and the head impact velocity was 0.9393 (Figure 11).

Figure 12 and 13 show the comparison of HIC by normalizing these values by that of the base model and the correlation between normalized HIC and the head impact velocity, respectively. The tendency of the comparison of normalized HIC was similar to that of normalized CIBIC in impact (Figure 12). That of case (f) with the maximum input to the pelvis showed minimum value. The coefficient of determination between normalized HIC and the head impact velocity was 0.9748 (Figure 13).

Figure 14 shows the comparison of the peak values of the accelerations of the ribs by normalizing these values by those of the base model. Except for the ribs between rib4 and rib9, the accelerations of the ribs with the restraint waist surface were lower than those of the base Model. In those of the ribs between rib4 and rib9, although they were also lower than those of the base model in most of cases, there are some cases that the accelerations were higher than those of base model. Figure 15 shows the comparison of the peak values of the sacrum force and pubis force by normalizing these values by that of the base model. The normalized sacrum force in case (a) was decreased to 86%. However, it was increased with increasing the stiffness of the waist restraint surface, exceeding that of base model in case (d). The normalized pubis forces showed almost the same tendency.

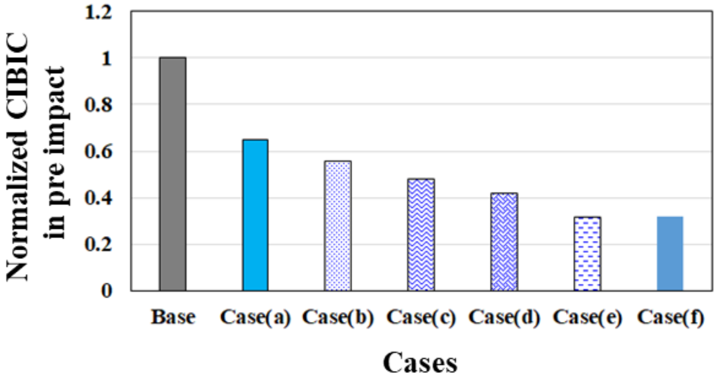

Figure 8. Comparison of CIBIC in pre-impact normalized by result from base model

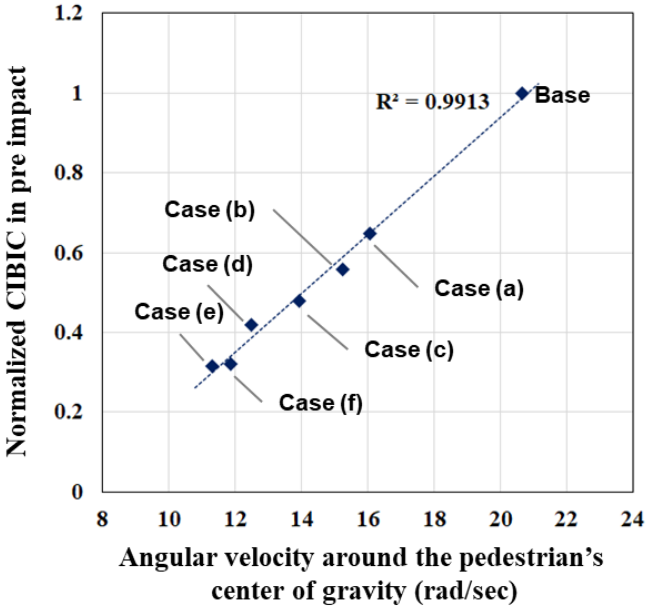

Figure 9. Correlation between normalized CIBIC in pre-impact and the maximum value of the angular velocity of the center of the gravity of the pedestrian model

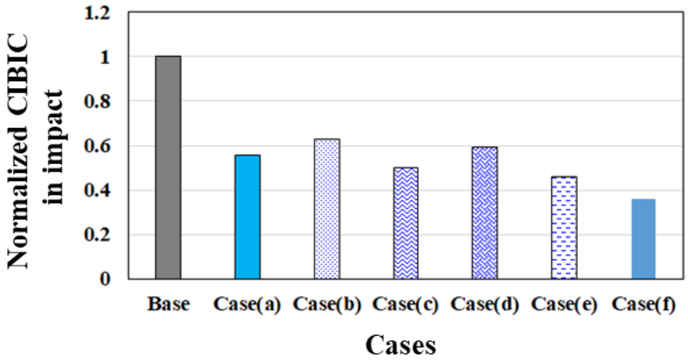

Figure 10. Comparison of CIBIC in impact normalized by result from base model

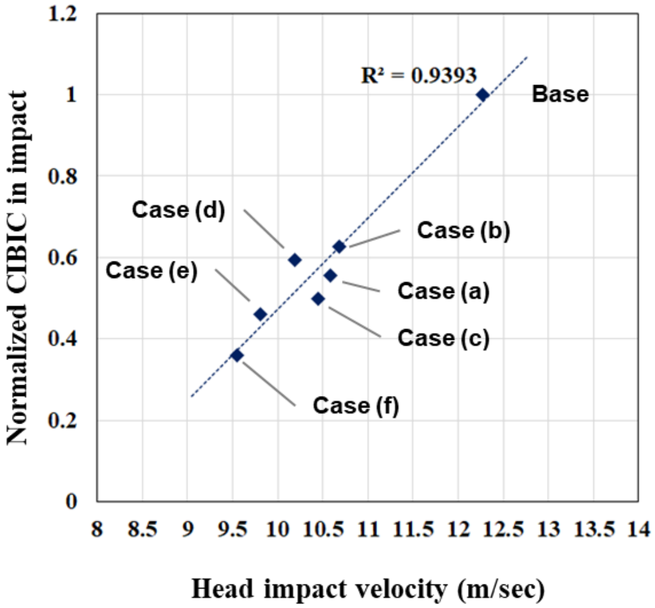

Figure 11. Correlation between normalized CIBIC in impact and the head impact velocity

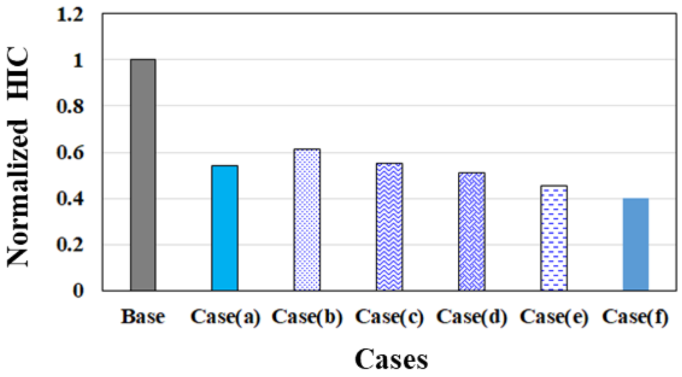

Figure 12. Comparison of HIC normalized by result from base model

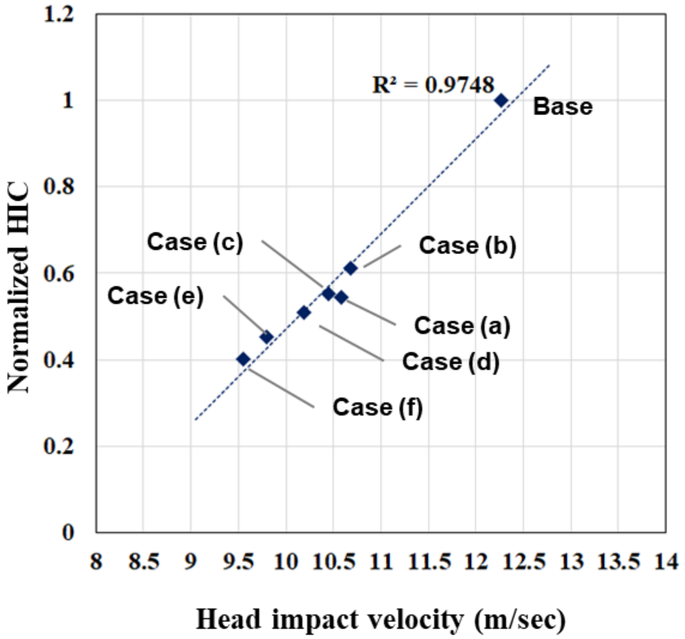

Figure 13. Correlation between normalized HIC and the head impact velocity

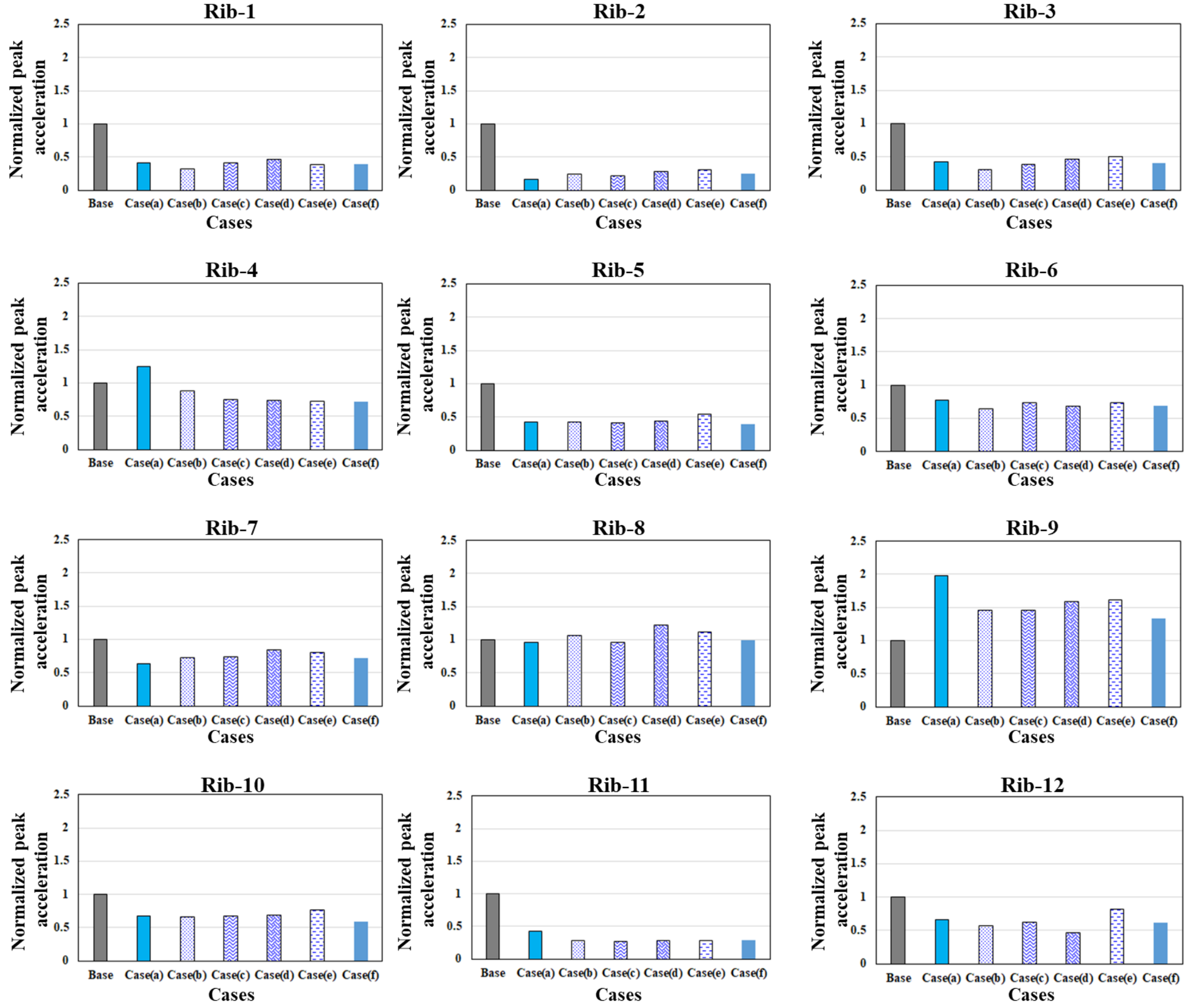

Figure 14. Ratio of the width between each rib and the spine at the timing of the maximum deflection to that of the original conditions from rib 01 to rib 12 of the impact side

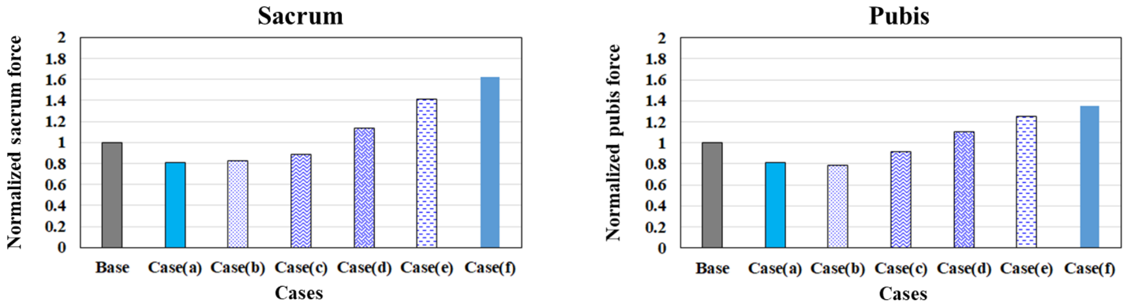

Figure 15. Comparison of the peak values of the sacrum force and the pubis force normalized by result from base model

DISCUSSION

As the results of the simulations that the human body model was collided with the vehicle model adding the waist restraint surface, CIBIC (pre-impact and impact phase) and HIC were reduced, showing the minimum injury values about the maximum input force to the waist. As shown in the time histories of the angular and translational velocities of the center of gravity, the times of these starts were earlier by increasing the input force to the waist, decreasing the slopes of the time history of them. They led to the decrease of the angular acceleration of the upper body. By the influence of the upper body regions kinematics, the angular accelerations of the head were also reduced. Since CIBIC is calculated by the angular acceleration of the head, the values were reduced. By increasing the waist input, the velocity of the center of the gravity with respect to the ground is increased, so the velocity of the center of the gravity with respect to the vehicle is reduced accordingly. Due to the interaction of the upper body regions, lowering the velocity of the center of the gravity leads to lowering the velocity of the head relative to the vehicle, reducing HIC. Reducing HIC means reducing the reaction force from the surface of the vehicle to the head. Since in the impact phase, the head rotated around the shoulder that contacted with the vehicle, the angular acceleration of the head in the impact phase increases with increasing the head velocity. Increasing the input force to the waist leads to decreasing the angular acceleration of the head. Thus, CIBIC was reduced by increasing the input force to the waist. As shown in above, CIBIC in both pre impact and impact phase and HIC was reduced by the larger input force to the waist. Generally, increasing the input force to the waist leads to increasing the risk of the fracture of the pelvis. In the case that showed the minimum value of CIBIC in both pre impact and impact phase and HIC, the sacrum and pubis forces were increased compared to the base model. In the most of cases with the restraint waist surface, the accelerations of the ribs were lower than those of the base model. However, in the region around the arm, those of the ribs were higher than those of the base model. It is thought that the contact between the arm and the thorax influenced the acceleration of the ribs,

A future study needs to focus on the compatibility of the injury measures of these body regions. As one way to do this, the methods the input force to the waist is changed from the concentrated to the distributed force.

CONCLUSIONS

In this study, the effects of whole-body kinematics from the phase of the first contract between the lower body of the pedestrian and the vehicle to that of the head impact were investigated by controlling the input to the pelvis. As the results of this study, the conclusions as shown in below are reached.

The injury measures in the phase of the head swing (CIBIC) and that of the head impact (HIC and CIBIC) can be reduced by controlling the value of the input to the pelvis. Investigation about compatibility of the injuries of the head and the pelvis is needed in order to reduce the injuries of these parts. These results will contribute to the reduction of the number of the pedestrian fatalities. The investigation was conducted by using the standard sedan model vehicle model and the 50th percentile male model in this study. However, the whole-body response is influenced by the geometry of the vehicle and the size of the pedestrian. It is necessary to investigate these influences.

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