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Investigation of lower limb injury under different contact stiffness for drivers during frontal crash

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Lower extremity injuries in AIS2+ were the most costly injuries through the statistical analysis of traffic accidents. This study aimed to investigate the response characteristics of the lower limb with different contact stiffness, in which knee cushion and foot cushion were applied. First, a model with a human body and a car was established, and the muscle function was activated in lower extremity of human model. Second, the deceleration pulse with a peak of 186 m/s^2 was applied to the car to simulate the frontal crash. Then, four sets of simulations with different contact stiffness are conducted to obtain the lower limb responses. Results indicate that the maximum loading of the left and right legs during the impact was 1.29 and 1.22 kN, respectively. Meanwhile, the maximum moment were 28.82 and 52.17 Nm, respectively. The maximum stress of lower extremity was 87.35 MPa, and the maximum tibia index was 0.230. It was demonstrated that the injury risk of the femur in the groups with equipment of knee cushion and foot cushion was low, but the injury risk of the tibia increased at the same time. This study could provide a reference to the study of lower limb injury in a frontal impact.

Key words: *frontal impact, knee bolster, lower extremities injury, contact stiffness, injury outcomes*

1. Introduction

Lower extremity injuries were commonly observed in motor vehicle crashes (MVCs), and were often assessed as Abbreviated Injury Scale (AIS) 2+ injuries. NASS-CDS data from 1998 to 2010 showed that drivers who suffered at least one AIS2+ injury in frontal crashes accounted for 46.3% of all injury cases [30]. In addition, according to UK accident data, AIS2+ lower extremity injuries were the most costly injuries, and they accounted for about 43% of the total cost in frontal and side impacts [17]. AIS2+ injuries in lower extremity were associated with a substantial economic burden on society and every driver. Therefore, the protective effect of restraint systems related to the lower extremity should be paid attention to.

The former research explored the factors that caused lower limb injuries. In MVCs, the most direct factor that causes lower limb injuries was contact loading between the knee and the dashboard. A long-distance between the knee and the dashboard could be a possible approach to reduce the knee impact force, which was concluded from the study with drivers wearing seatbelts [9]. Furthermore, research showed that the distance between the knee and the dashboard was significantly reduced due to the dashboard invasion. Therefore, shortening the movement distance of the knee could cause serious tibial axis injury under frontal impact [25]. Beside increasing the distance between the knee and the dashboard, reducing the contact stiffness of the dashboard could also reduce injury risk. A lower dashboard stiffness may effectively diminish knee shear injury risk, as confirmed by Gok-

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hale [8]. Another study revealed that the patella was fragile when impacted by a rigid interface. And this conclusion was obtained when the patella was relative to the axis location of femur slant or axial impact knee [14]. A study indicated that sharper objects could produce concentrate loading to the lower leg and a high risk of tibia/fibula fracture [4]. From the former research, controlling the stiffness of the dashboard as well as ensuring sufficient space between the knee and the dashboard were two crucial ways to reduce the injury risk of lower limb.

The biomechanical analysis of trauma used dummy and Post Mortem Human Subjects to speculate the injuries in real accidents. However, there's a certain difference between the substitution and the human body. Recently, researchers have shown a growing interest in muscle function during impact. One study found that the bending moment of the femur shaft increased when the lower extremities muscle was tense, resulting in an increased risk of femur shaft fracture [3]. Another study pointed out that as muscle activation levels increased, the risk of lower extremity injuries became higher and the peaks of injury parameters were observed later [6]. Besides, there were different leg postures in real-world traffic accidents, which might be related to the brake pedal that was stepped by drivers for emergency braking [11]. Most of the studies mentioned above have explored the factors that cause lower extremity injuries, including muscle function and sitting postures. In spite of that, the difference between left and right lower limb injuries was not sufficiently studied.

Rigid contact between the knee and the dashboard can cause injuries to knee, femur, tibia, and pelvis. In order to reduce the stiffness of contact surface, the knee airbag (KAB) was widely used. The use of KAB could reduce the risk of serious injuries caused by rigid contact and improve the overall kinematics of driver [10], [23]. In addition, KAB might have a protective effect on knee-to-hip injuries [13], [24]. When studying the protective effect of KAB on the lower extremity, two significant results were obtained. First, if the KAB was placed at the bottom instead of at the rear in an oblique impact, more coverage and less leg abduction was provided [18]. Second, due to the active muscle response, the risk of lower extremity injury increased when the occupant interacted with KAB [19]. However, Ye's research showed that inflated KAB might stretch the knee of small-sized occupants (represented by the 5th Hybrid III female dummy). Moreover, when the knee was stretched, the KAB could not interact with it, resulting in a high risk of lower extremity injury [29].

The knee cushion (KC) was a convenient solution for reducing lower limb injury. Compared to KAB, the passive KC was more friendly to small-sized occupants. With the increasing usage of passive KC, the protective effect of passive KC obtained more attention. Gokhale developed a passive knee bolster and verified its protective effect, and the results indicated that it could improve the kinematics of the driver and reduce the risk of rigid contact between the knee and the parts of the dashboard [7]. A three-piece cellular structural knee bolster could achieve a very low femur load in the NCAP tests [1]. The KC was low-cost and convenient, while KAB was mainly deployed in high-end cars, which was less popular. Meanwhile, KC had nearly similar protective effects as KAB, and it had become another choice for protecting the lower extremity. Additionally, it was common to deploy foot cushion (FC) on the cab floor, but the effect of FC or carpet airbags on lower extremity injuries was not clear. The influence of the KC and FC on lower extremity injuries still needs further research.

The current study used the lower extremity Finite Element (FE) model combined with a vehicle FE model to establish the study environment. The influence of different contact stiffness on lower limb injuries was explored to study the effect of KC and FC on lower limb protection under 40% offset frontal impact.

2. Materials and methods

2.1. Model introduction and validation

The study model consisted of a vehicle and a human body model (HBM) (Fig. 1). The vehicle model included KC and FC. A KC with thickness of 10 mm was placed on knee bolsters, and a FC with thickness of 7 mm was placed on the cab floor. In addition, the material of KC and FC was material No.024 in LS-DYNA, which was *MAT_PIECEWISE_LINEAR_PLASTICITY. Moreover, mass density, Poisson's ratio and yield stress of the materials used in KC and FC were consistent (Table 1).

HBM was placed in the cab of the car model. Furthermore, the sitting posture of the HBM was a usual driving posture as NCAP. The hands of the HBM were placed on the steering wheel. The left foot of the HBM was placed on the footrest while keeping the heel in contact with FC or cab floor. Similarly, the right foot of the HBM was placed on the brake pedal

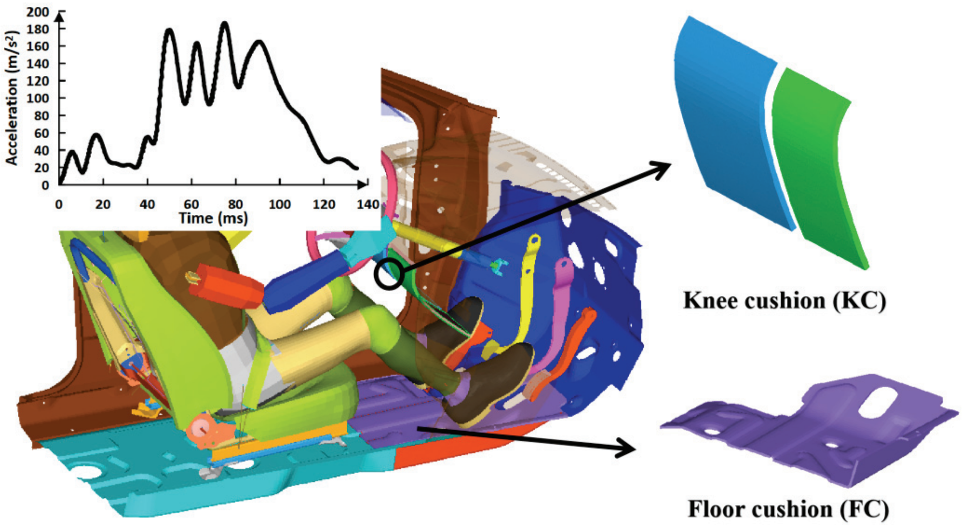

Fig. 1. Study model and simulations conditions

Table 1. Material parameters for KC and FC

Mass density [kg/m^3]	Poisson's ratio	Yield stress [GPa]
1200	0.3	0.045

and the right heel was in contact with FC or cab floor. Moreover, the sliding friction coefficient between knee and KC was set to 0.3 and the sliding friction coefficient between shoe and FC was set to 0.6.

In terms of the HBM, Mo established a lower extremity model with computer tomography (CT) and magnetic resonance imaging (MRI) techniques, and it was based on the data from medium-sized Chinese male volunteers [15]. This lower extremity model contained prominent bones and muscles, such as femur and tibia, 11 thigh muscles, 11 calf muscles and main ligaments. The muscle function was activated in lower extremity of human model. Referring to the previous research, the muscles about the tibialis anterior, gastrocnemius, soleus, rectus femoris and biceps femoris were activated during impact [2], [21]. These muscles were simulated with detailed 3D geometric figures and controllable activation elements. Besides, the mechanical properties of 3D active muscles were verified through volunteer experiments [16]. Therefore, these muscles could simulate the action of lower extremity in a brake, such as hip flexion, hip extension, knee flexion, knee extension, plantar flexion and dorsiflexion. The lower extremity model was verified by recent dynamic experiments and quasi-static experiments. Verification content from isolated materials to sub-segments and finally the whole lower limb, especially considering its passive properties [16].

2.2. Injury index

2.2.1. Force/moment distribution

In order to better characterize the driver's lower limb injury, the left femur, the right femur, the left tibia, and the right tibia were selected as the locations to obtain the section force/moment (Fig. 2). With reference to former research, the cross-sections were taken into account to evaluate the effect of muscle activity on the forces and moments in femur and tibia [12]. Moreover, the contact force between the knee and the knee bolster (or the KC) was measured to analyze the knee injury. The L represented the left leg, and R represented the right leg. For example, the RM-femur represented the mid-section of the right femur.

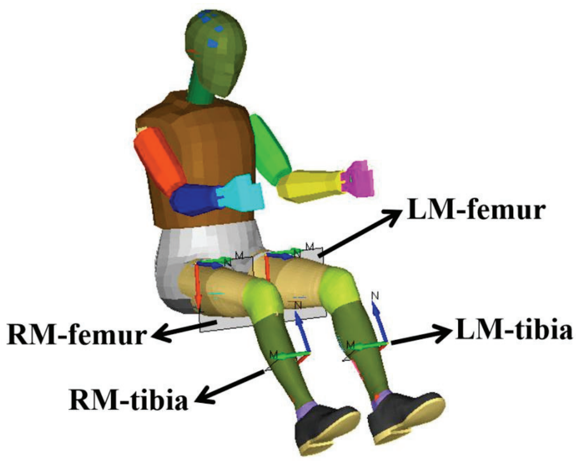

Fig. 2. The measuring locations of force/moment

2.2.2. Stress distribution

As an injury index, the stress distribution could obtain detailed conditions and easily observe the locations of stress concentration. The parts under stress concentration may be cracked or even fractured and they should be paid more attention to when evaluating the injury. The bone measurement locations, including femur and tibia could reflect the stress peaks and concentration locations.

2.2.3. Tibia index

The tibia was a significant weight-bearing bone of the lower limb and had a slender structure. When the knee impact with the knee bolster and the ankle rotated due to the intrusion of the pedal, the risk of tibia injury increased. In this study, tibia index (TI) was used to predict the injury risk of the tibia shaft fractures in impact, and TI consisted of the axial force and the resultant bending moment, as shown in Eq. (1).

$$TI = \frac{F}{F_c} + \frac{M}{M_c}, \tag{1}$$

where F was the axial force and M was the resultant bending moment. The critical value was defined as $F_c = 12$ kN, $M_c = 240$ Nm [20].

2.3. Simulation matrix

In this study, simulation groups with active muscle were designed to investigate the effects of two variables (the functionality as well as elastic modulus of KC and FC) on lower limb injuries (Table 2). In addition, the stiffness of KC and FC changed with the elasticity modulus in this study, while the other characteristics of the KC and FC remained the same. For example, the KC and FC were not used in NE0. In contrast, the KC and FC were equipped in the other three simulations. Moreover, the elastic modulus of the KC and FC varied among all groups. The elastic modulus of the KC and FC was set as 2.8 GPa, 12.8 GPa and 22.8 GPa, and these values referred to those common values used in practical engineering. At last, four groups of FE simulations were carried out. The corresponding relationship between a specific simulation and the two variables in the following text was represented by simulation codes, where N represented the absence of the KC and FC, and Y represented the presence of the KC and FC. Besides, E represented the elastic modulus of the KC and FC.

Table 2. Simulation matrix

Simulation ID	KC and FC	Elastic modulus [GPa]
NE0	N	–
YE2.8	Y	2.8
YE12.8	Y	12.8
YE22.8	Y	22.8

3. Results

According to the mentioned methods, simulations were carried out. Moreover, in a 40% offset frontal impact, the results were obtained using the mentioned methods to analyze the influence of KC and FC on lower limb injuries.

3.1. Force/moment distribution

The duration time of impact contact between the knee and the knee bolster (or the KC) was inconsistent for the right leg in all groups, according to the results. The shortest duration time of the knee and knee bolster impact was 27.7 ms in NE0, while the largest duration time of the knee and KC impact was 42.4 ms in YE2.8. Similarly, the duration time of impact contact between the knee and the knee bolster (or the KC) was observed to be different in all groups for the left leg. However, there was no impact contact in NE0. In addition, the longest duration time of the knee and KC impact contact was observed in YE2.8, which was 20.0 ms, and the shortest duration time was noticed in YE22.8, which was 8.5 ms. The data revealed that the duration time of impact contact decreased as the elastic modulus of the KC and FC increased.

When it came to the contact force between the knee and the knee bolster (or the KC), the impact contact force between the left leg and the KC was smaller than that of the right leg (Table 3). In NE0, there was no impact contact force between the left leg and the knee bolster, while the peak contact force of 1.44 kN was observed in YE2.8. Furthermore, when the KC stiffness rose, the peak contact force between the left leg and the KC decreased. With increasing KC stiffness, the contact force between the right leg and the KC increased (Table 4). In NE0, the minimum contact force of 1.76 kN was obtained for the right leg. And the peak contact force was 2.58 kN, which was captured in YE22.8. The difference in contact force between the greatest and smallest was 0.82 kN.

Table 3. Peak force and peak moment of left lower limb

Measurement target	Simulation ID	F_{max} [kN]	t [ms]	M_{max} [Nm]	t [ms]
LM-femur	NE0	0.56	73.6	25.86	64.4
	YE2.8	1.29	87.9	23.92	79.8
	YE12.8	0.75	93.5	23.64	80.4
	YE22.8	0.55	73.1	28.82	60.5
LM-tibia	NE0	0.73	51.6	18.55	77.9
	YE2.8	0.59	94.2	23.27	88.2
	YE12.8	0.70	56.3	21.94	77.9
	YE22.8	0.88	57.8	19.78	78.0
Impact contact force (L)	NE0	0.00	0.0	–	–
	YE2.8	1.44	87.9	–	–
	YE12.8	0.69	92.6	–	–
	YE22.8	0.03	101.9	–	–

Table 4. Peak force and peak moment of right lower limb

Measurement target	Simulation ID	F_{max} [kN]	t [ms]	M_{max} [Nm]	t [ms]
RM-femur	NE0	0.75	92.3	42.05	81.7
	YE2.8	0.71	83.4	27.22	71.2
	YE12.8	1.07	87.2	49.36	84.0
	YE22.8	1.22	87.2	52.17	87.6
RM-tibia	NE0	0.74	76.3	29.37	79.8
	YE2.8	0.75	94.4	50.50	78.7
	YE12.8	0.76	94.3	40.00	80.7
	YE22.8	0.87	93.0	45.60	80.4
Impact contact force (R)	NE0	1.76	91.9	–	–
	YE2.8	1.79	82.8	–	–
	YE12.8	2.50	86.4	–	–
	YE22.8	2.58	83.7	–	–

When the elastic modulus of the KC and FC was 2.8 GPa, the peak force and peak moment were lower than those of NE0 for the right femur (Fig. 3). It was captured that the peak force of the right femur in YE2.8 was reduced by 0.04 kN, compared to that in NE0. In addition, when comparing YE2.8 to NE0, the peak moment of the right femur was lower by 14.83 Nm. Furthermore, the peak force difference between YE2.8 and YE12.8 was twice as large as the force gap between YE12.8 and YE22.8, measuring 0.36 kN and 0.15 kN, respectively. When compared to the difference between the peak moments of adjacent groups, the difference between YE2.8 and YE12.8 was more significant than the difference between YE12.8 and YE22.8, with values of 22.14 Nm and 2.81 Nm, respectively. The results showed that when the elastic moduli of the KC and FC grew, so did the peak force and peak moment. Meanwhile, the increase in peak force and peak moment was non-linear. For the left femur, the results presented that the peak force of the left femur decreased as the elastic moduli of the KC and FC rose. And the maximum peak force was found in YE2.8, which was 1.29 kN. There was a slight difference in the peak moment of all groups, and the maximum difference of peak moment in all groups was 5.18 Nm (Fig. 4).

For the right tibia (Fig. 3), the peak force increased as the elastic modulus of the KC and FC increased. In addition, the minimum peak force of right tibia in NE0 was observed as 0.74 kN, which was by 0.01 kN lower than the peak force of right tibia in YE2.8. The peak force of the left tibia was similar to the peak force of the right tibia. Both the left and right tibia showed that the peak force increased as the elastic moduli of the KC and FC increased. For left tibia, the minimum peak force was observed in YE2.8, which was by 0.14 kN lower than the peak force of left tibia in NE0 (Fig. 4).

It could be seen that the minimum peak moment of RM-femur and the maximum peak moment of RM-tibia were captured in YE2.8. The correlational analysis revealed that the minimum peak moment in the LM-femur and maximum peak moment in the LM-tibia found a weak regularity.

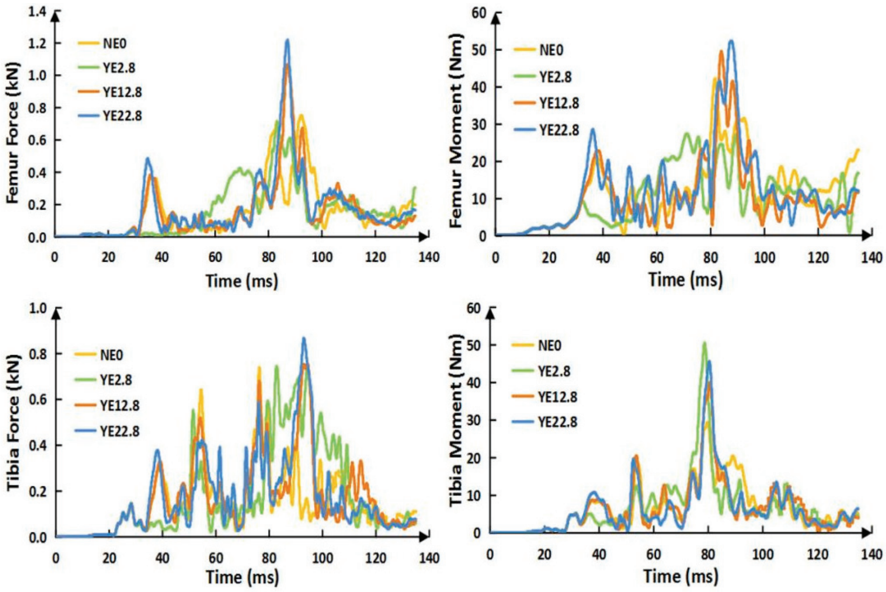

Fig. 3. Comparison of force and moment results in the right leg

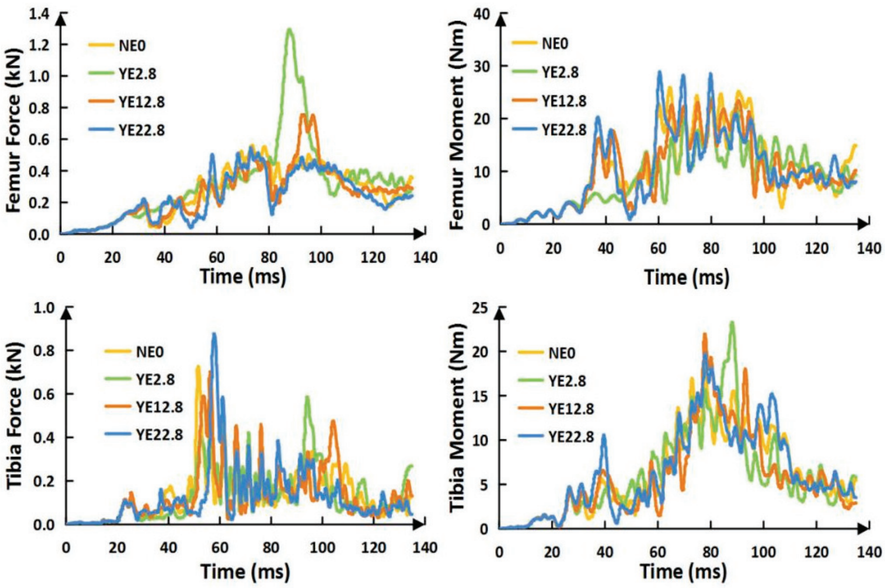

Fig. 4. Comparison of force and moment results in the left leg

3.2. Stress distribution

The results showed that the peak stress of right femur in NE0 was the maximum, compared with that in YE2.8, YE12.8, and YE22.8 (Table 5). The difference between the peak stress of right femur in YE2.8 and NE0 was found to be the largest among them. And the peak stress of right femur in YE2.8 was about 32% smaller than that in NE0. The peak stress of the right femur was shown to increase when the KC and FC elastic modulus rose. In NE0, YE2.8, and YE12.8, the peak stresses of the right femur were found near the intertrochanteric line of the femur. And in YE22.8, the peak stress of right femur was presented in the middle of the shaft (body).

Regarding the left femur, it was observed that the peak stresses in YE2.8 and YE12.8 were greater than that of NE0. It was noticed that the maximum stress of left femur was in YE2.8, which was 25.40 MPa. And the peak stress of left femur in YE2.8 was by about 10% bigger than that in NE0. The peak stress of left femur decreased with the KC and FC elastic modulus. In NE0, YE2.8, and YE12.8, the peak stresses of left femur were found on the popliteal surface. And in YE22.8, the peak stress of left femur was obtained near the intertrochanteric line.

Table 5. Peak stress of lower limb in simulations

Measurement location		Peak stress of lower limb	
		Femur [MPa]	Tibia [MPa]
NE0	L	23.18	39.68
	R	38.61	61.01
YE2.8	L	25.40	42.70
	R	26.34	87.35
YE12.8	L	23.24	42.36
	R	30.30	57.32
YE22.8	L	21.80	43.88
	R	32.53	62.42

YE2.8, which was 0.230. Furthermore, the minimum TI of the right tibia was reflected in NE0, which was 0.137. In addition, compared with the TI in NE0, higher TI was shown in each group with KC and FC. From the results, the regularity of the left tibia was weaker than that of the right tibia. In left tibia, the differences in TI between the groups were small. On the whole, a higher TI was noted in the right tibia than in the left tibia under a 40% offset frontal impact.

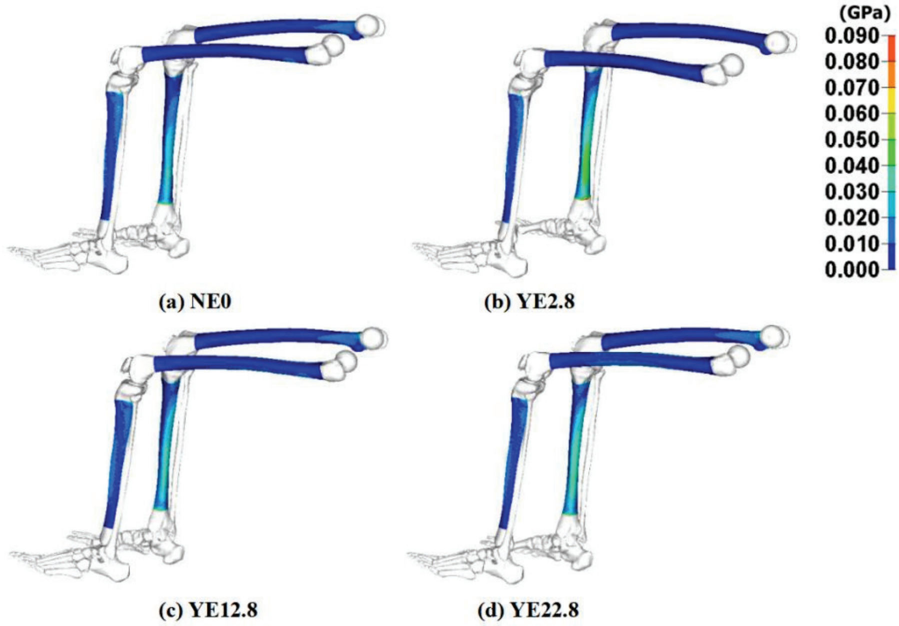

Fig. 5. The stress distribution at the moment of femur and tibia peak stress

In terms of the left tibia, the stress in NE0 was the minimum. With increasing elastic modulus, the peak stress of the left tibia changed from decline to rise and the peak stress in YE12.8 was by about 7% greater than that of NE0. In NE0 and YE12.8, the peak stresses of left tibia were presented near the oblique line of tibia. In addition, the peak stresses of left tibia were noticed near the medial malleolus of tibia in YE2.8 and YE22.8.

The maximum stress in the right tibia, right femur, left tibia and left femur were all found near the medial malleolus (Fig. 5).

3.3. Tibia index

Low TI was captured in the right and left tibia of all groups (Table 6). Moreover, TI did not exceed 1.0 in all groups. In all measurement locations, there were some noticeable differences in TI of right tibia. The maximum TI of the right tibia was reflected in YE2.8, which was 0.230.

Table 6. Tibia index results of simulations

Measurement location	Simulation ID	TI
Right	NE0	0.137
	YE2.8	0.230
	YE12.8	0.183
	YE22.8	0.192
Left	NE0	0.127
	YE2.8	0.126
	YE12.8	0.134
	YE22.8	0.124

4. Discussion

4.1. Injury analysis of right lower limb

In order to analysis of the injury risk of right lower limb, the impact contact force of right knee with KC is obtained in this study. Indicators are used to assess the risk of right femur injury, including maximum axial force, maximum bending moment and maximum stress. Evaluation of the risk of tibia injury includes the above three indicators as well as TI.

The results show that the peak force of the right femur increased with a stiffer KC and FC, and the peak force of the right femur in YE2.8 was the smallest among all simulation groups. This is because the peak force of femur is related to the contact force. It can be observed that the contact force between the right knee and KC increased with a stiffer KC. This is caused by the severe impact between the right leg and KC, and the weak resistance provided by FC. As a result, the stiffness of FC only has a little effect on the change in peak contact force. The change in peak contact force, on the other hand, is strongly influenced by the stiffness of KC. When a softer KC is applied, the severity of the impact between the knee and KC is lighter. On the right tibia, results show that groups with KC and FC obtained a higher peak tibial force than NE0. Furthermore, when stiffer KC and FC are used, the peak forces on the right tibia rise. The minimum peak force of the right tibia is captured in NE0. In NE0, the smallest displacement of the foot and knee is observed at the moment of maximum peak force. It was noticed that the stretched tibial posture causes an increase in the axial force of the tibia [31]. Ye's study got a similar result with this research.

Observation and analysis of the peak moment show that the minimum peak moment is found in the right femur at YE2.8. It may be that the moment of the femur would increase with a stiffer KC and FC. A similar trend was observed in Nie's research [18]. Meanwhile, the peak moment of the right tibia followed the same pattern as the peak force, with the maximum peak moment found at NE0.

The results of the right femur indicate that lower peak stresses are found in the groups with KC and FC. Patel's study observed similar conclusions that the deployment of KAB could reduce the risk of femur injury [22]. Further observation finds that the peak stress of the right femur would increase with a stiffer KC and FC, and the minimum peak stress is observed in YE2.8. The high stresses were obtained from the distal and proximal ends of the femur in van Rooi's study [26], and the results of this research could support his conclusion. Further statistical results reveal that 75% of the peak stress in the right femur is found near the intertrochanteric line and 25% of the peak stress – in the middle of the femoral shaft. The peak stress of the right tibia shows a trend from decline to rise as the stiffness of the KC and FC increases. The high stress was found in both the bottom of tibia and the tip of tibia in Li's study [11], which was consistent with the results of the present research. The peak stresses of the right tibia of all groups are captured in the vicinity of the medial malleolus. The simulation results of the peak stress are all found near the medial malleolus of the right tibia. This is mainly because the HBM tilts to the right under a 40% offset frontal impact. As a result, the contact force between the right leg and the knee bolster or the KC is strong. Due to the intrusion of the brake pedal and active muscle contraction, high stresses were found in the right tibia. In addition, the maximum stress of tibia was greater than the maximum stress of femur, which indicated that the injury of tibia was more serious than that of the femur [28].

The TI did not exceed the tolerance of 1.0, which means TI was low in all simulations. These results could represent a low tibial injury risk in all groups during 40% offset frontal impact. In addition, when the impact was conducted, a higher TI was noted in the right tibia than that in the left tibia. However, Ye's research results pointed out that the TI of the upper tibia was higher than that of the lower tibia [31]. The contact force between the right leg and the knee bolster (or the KC) is greater than that of the left leg. So, the high TI of the upper tibia may be related to the value of contact force. Results show that, compared to NE0, the groups with the application of KC and FC are more likely to obtain a high TI.

Overall, it can be observed that presence or absence of KC and FC differently influences the right lower extremity injury risk in a frontal impact. In the impact, a lower right femur injury risk and a higher right tibia injury risk are captured in groups with KC and FC compared to NE0. Patel's studies [22] and Weaver's studies [27] also found results similar to those obtained in the present study.

4.2. Injury analysis of left lower limb

The indicators for evaluating the risk of injury of the left lower extremity are consistent with those of the right lower limb. In other words, indicators are used to assess the risk of left femur injury, including maximum axial force, maximum bending moment, and maximum stress. Evaluation of the risk of left tibia injury includes the above three indicators, as well as TI. It is observed that when the stiffness of KC increases, the peak contact force between the left leg and KC decreases. It is possible that HBM tilts to the right under a 40% offset frontal impact, making the impact between the left leg and KC less severe. When the stiffness of FC decreases, the resistance provided by FC and the displacement move by knee increases. Thereby, the peak contact force between the knee and the KC increases. In addition, the results also show that the left leg in the NE0 hardly contacts the knee bolster. Because KC is not applied in NE0, the distance between the knee and the knee bolster is large, and the movable displacement of the knee would be prolonged. It was pointed out that a larger distance between the knee and the dashboard could reduce the knee impact force, which was consistent with the results of this study [9].

In addition, the peak force of the left femur decreases with a stiffer KC and FC, and the maximum force of the left femur is found in YE2.8. Moreover, the peak forces of the right and the left femur are both less than 10 kN, which is the injury tolerance defined by FMVSS 208. It's also indicated that the kinematics and dynamics of the femur in the low-to-medium speed impact test are far below the injury tolerance [5]. The peak force of the left and right femurs differs due to the variability in the severity of impact. It seems to be slightly different in the left leg. In the NE0, the driver's knee does not contact the knee bolster. Furthermore, the impact contact force in the YE12.8 is low, while the impact contact force in the YE22.8 is almost negligible. Moreover, the maximum impact contact force is observed in YE2.8. The peak forces of left tibia increase are obtained with a stiff KC and FC. And the minimum peak force of the left tibia is captured in YE2.8. The minimum peak force of the left tibia is observed in YE2.8. This may be because the contact force of the left leg is significantly smaller than that of the right leg. In addition, the distance between the knee and the KC becomes longer when the stiffness of KC decreases. The interaction time between the knee and the KC was prolonged, resulting in a lower peak force [18]. Except for the peak moment of the left tibia in YE2.8, the peak moments of the left femur and left tibia in other simulations are not found in the impact phase. Therefore, the minimum peak moment of the left leg is inconsistent with the right leg.

The peak stresses of left femur decrease along with an increase of the stiffness in KC and FC. And the peak stress in YE2.8 is the greatest among all simulation groups. The peak stress and peak force of the femurs are changing similarly. As a result, the cause of variations in peak stress and peak force could be the same. In the left femur, 75% of the peak stress in left femur is found on the popliteal surface, and 25% of the peak stress is found near the intertrochanteric line. The peak stress in the left tibia shows a pattern from decline to rise as stiffness in the KC and FC increases. In the groups with KC and FC, it can be observed that the peak stress of tibia may be related to the angle between the tibia and the foot. The peak stress of the tibia decreases with the angle. Similar results are not found in the left leg, which may be because that the contact force between the left leg and KC is small. Thus, the difference between the peak stress of left tibia in each group is insignificant.

4.3. Future study

This paper investigated the effect of different contact stiffness of KC and FC on lower extremity injuries. The research results could be applied to various types of vehicles and could provide a reference for the parameter design of passive knee bolster. However, there are many factors that influence injury risk of lower limb in an impact, e.g., the size of driver, pre-collision posture, seat installation position, angle of seat back, use of seat belts and speed of vehicle. This article could be used as a reference to study the mechanism of lower extremity injury. In future research, various factors could be combined for a more in-depth discussion, making study results richer.

This study only focuses on the lower limb responses of the HBM, and injuries of other parts, such as the chest, head, and neck injuries, are not considered, which makes the conclusions of this study limited. Several fields can be applied to improve the accuracy of the research results.

5. Conclusions

Summarizing, this study identified that the impact between the right leg and the knee bolster or KC is more severe than that of the left leg. However, due to the low impact speed, the results of each group have shown that the injury risk to the left and right lower limbs is relatively low. Compared to the injury risk of right femur in NE0, a lower right femur injury risk is found in YE2.8. However, there is an increase in right tibia injury risk in YE2.8. If injuries of other parts (such as the chest, head and neck) are not considered, the use of KC and FC with less stiffness may reduce the risk of femur injury, but the risk of tibial injury might increase. And as the stiffness of KC and FC increases, the increase of injury risk is non-linear. Moreover, it is found that the smaller the contact stiffness of KC and FC, the greater impact the contact force.

The highlights of this research can be summarized as follows. First, the loads loaded in this study are derived from actual car crash experiments. Second, the protective effect of the KC and FC with different stiffness during impact is compared.

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Conflicts of interest

The authors declare that there is no conflict of interest regarding the publication of this paper.

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