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Investigation of Ankle Joint Responses with Emergency Braking During Frontal Impact

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Keywords： active muscle, ankle joint injury, ligament injury, moderate overlap impact.

ABSTRACT

This study aims to provide a reference to the study of lower extremity injury prevention during a frontal impact. First, the boundary conditions of a car model are established. Then, through intrusions and accelerations characters of inner parts to simulate the whole process of a full-scale car accident, the ankle joint injury is determined by the injury index like ligaments kinetics around the joints under emergency braking. Results demonstrate that in the model without active muscle function, the bending angle of the right ankle is increased by 11% compared with that of the left ankle. In the model with active muscle function, the bending angle of the right ankle is increased by 16% compared with that of the left ankle. This study investigates differences in ankle kinetic characteristics due to the functionality of lower extremity active muscle under 40% offset frontal impact.

INTRODUCTION

Ankle injury is a primary source of the lower extremity injuries during the car crash accident (Shin et al., 2013), especially the injury caused by the violent intrusion of the pedals in upward and backward directions during the moderate overlap impact (MOI). The ankle joint is flexible in some directions, such as dorsiflexion, plantar flexion, and a certain range of lateral movement, in which the ligaments around ankle may be largely extended by the excessive movement of the feet. Thus, the relationship between the ankle kinematics and injuries should be deeply investigated to better understand the ankle injury mechanism (Bailey et al., 2017). The connections in ankle joint are the ligaments between foot and tibia. It is known that most lower extremity injuries are captured among drivers due to the great intrusions from pedals and instrument panel (IP) relative to the ankle and knee. The lower extremity injuries are investigated, which showed that ankle joint injury is serious in MOI. In detail, most ankle joint injuries are obtained with fractures in bones and tears in ligaments (Vetter et al., 2020). In order to protect the driver's ankle, the injury outcome characteristics during the impact should be better understood. A study shows that about 67% of drivers would have emergency brakes when they realize the upcoming dangerous situation (Hault-Dubrulle et al., 2009). This means the muscle function activated during emergency braking may influence the injury risk of ankle. Thus, the influence of the active muscle function on ankle joint injuries should be more deeply considered.

Lots of studies have been done to investigate the injury mechanism in the ankle joint. Among the ankle joint injury studies, Funk established a finite element model of ankle joint with linear and nonlinear viscoelastic models of eight types of ligaments. And Funk points out that ligaments are nonlinear viscoelastic (Funk, 2011). In another study, researchers discover the role of each ligament in ankle stability and the effect of ligament sectioning on a range of motion and overall laxity (Palazzi et al., 2020). Meanwhile, it is found that the instantaneous rotating fulcrum of ankle joint, joint surface shape, and ligament geometry are closely related to each other (Mait et al., 2015). Furthermore, change in the angle of the ankle joint is conducted through the study of the lower extremities of nine male post mortem human subjects (PMHS), and it is concluded that the medial ligament would be injured before other parts during the impact (Mait et al., 2018). The index of bending angles and ligament's dynamic

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responses are essential to the ankle injury studies and are not fully considered in the former studies.

The lower extremity injury is estimated to be influenced by the active muscle function heavily due to the previous study in accident cases (Funk, 2011), especially with the driver's action of emergency braking. One of the explanations for the sources of these differences is that the active muscle may influence the stiffness distribution of the joints. Meanwhile, the mechanism and the percentage of the influence are not fully qualified studied. Thus, it is demonstrated helpful to study and compare the difference between the kinetic responses obtained from the active muscle cases and inactive ones (Beeman et al., 2012; Peng et al., 2012). It is also noted that the ankle rotation might be associated with the injury in the joints (Li et al., 2019), which is the index of the lower extremity injury risk. The foot would rotate relative to the heel due to the upward and backward pedal intrusion. Moreover, the ankle joint will be in a dangerous situation (overstretch of ligaments), which is realized as the primary cause of ankle joint injuries. Furthermore, comparing injury outcomes obtained with and without the active muscle can explain the difference and sources between statistical data and PMHS tests.

About the study of active muscle influence on injury outcomes, Nie uses a simplified model to study the outcomes under extreme conditions with active muscle (Nie et al., 2015). According to a previous study, ankle stiffness increases with muscle activation, which implied that hypertonus would generally result in elevated stiffness (Lee et al., 2014). It is speculated that active muscles are a factor that affects the flexibility of the ankle joint, which may affect the injury outcomes. Another study provides evidence that active muscle function could affect structural parameters of the lower extremity (Knaus et al., 2022). Because of the existence of the active muscle functions in the real accident, the injury outcomes in the real accident may be different from the results in cadaver tests. Thus, studying the influence of active muscle functions on injury outcomes has become necessary for driver protection research.

This study is organized as follows. The biomechanical numerical model of the extremity is first established with passive and active muscle functions (Xiao et al., 2020). Then, the extremity model is associated with a car model to establish the crash environment. Third, the MOI boundary condition is built according to the Insurance Institute for Highway Safety (IIHS) regulation. At last, the injury outcomes under two models and compared to analyze the differences caused by the muscle functionalities.

Material and methods

Human body model

The study model included a car, a driver, and restraint systems. The car, driver, and driver-restraint system were all validated to have good biofedility through the comparison between simulation and conference experiment (Mo et al., 2018a). However, the car model is established based on the geometric model of a market-sold car, which includes the body, belt, and airbag. Meanwhile, the driver model consisted of two separate systems. First, a Hybrid III 50th dummy upper body was applied in the study, considering the stability and calculation time of the model. Second, the lower extremity model generated based on computed tomography (CT) scan and magnetic resonance imaging (MRI) of adult males in China was developed and improved. A total of nine dynamic and quasi-static experiments were performed to validate that the lower extremity model can be used in the injury bio-mechanical study (Mo et al., 2018b and Mo et al., 2019). In addition, study also showed that the current driver-restraint system could realistically reflect the dynamic response of the driver during the impact (Mo et al., 2018a).

The current research model is established according to the code of LS-DYNA, in which the explicit algorithm can quickly solve transient large deformation dynamics, large deformation and multiple nonlinear quasi-static problems, and complex contact problems. LS-DYNA has unique advantages in transient computing like automotive safety analysis and has been widely recognized.

The human body model was set in the vehicle according to the prescribed posture of frontal impact regulations. In most vehicles, the right leg is the most affected by the intrusion of the braking pedal. Thus, responses of the right leg had been deeply studied because this leg was in the most complex conditions in the impact. The long lateral ligaments and primary bones of the ankle joint were also involved in the driver model (Fig. 1).

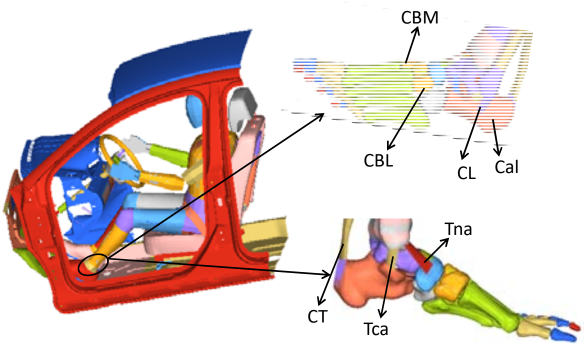


Fig. 1. Study model and main ligaments in ankle model.

The ligaments involved in this study were Calcaneofibular ligament (CL), Tibiocalcaneal part (Tca), Tibionavicular part (Tna) and Calcaneal tendon (CT). And primary bones of the ankle joint involved in this study were calcaneus (Cal), cuneiform bone lateral (CBL), and cuneiform bone medial (CBM). These ligaments were located in the vulnerable locations of the ankle joint. Meanwhile, the contacts among inner parts of driver dummy were automatic single surfaces with a friction coefficient of 0.1, and the contacts between leg and vehicle were automatic surface to surface contacts with a friction coefficient of 0.1-0.3 depending on materials of the contact parts.

The study model contained 189,580 elements and 168,376 nodes. Among them, the finite element model of the driver's lower extremity contains 97 components, with a total of 65,626 elements, 40,155 solid elements, 25,263 shell elements, and 208 spring elements. Furthermore, the mesh structure and quality of the model were controlled. The quality of the elements was determined. The proportion of elements with a Jacobian coefficient lower than 0.6 was less than 3%, and the proportion of elements with a warpage degree greater than 15° was less than 8% (Jiang, 2014).

Active muscle method

In the finite element (FE) dummy model, the discrete elastic element based on Hill's theory was used to simulate muscle units. The relationship between muscle contraction force and contraction speed was shown in Eq. (1).

$$\left(F^{CE}+a\right)\left(V+b\right)=b\left(F_{\max}+a\right) \quad (1)$$

Wherein, F^{CE} was the muscle contraction force, V was the contraction speed, F_{max} was the maximum isometric contraction force, a and b were constants.

Hill's muscle model was mainly divided into three parts. There were contraction element (CE), series elastic element (SEE), and parallel elastic element (PE), respectively (Hill, 1970). The CE represented the contractile protein of myofibrils, actin, and myosin. And the PE was composed of connective tissue around the muscle fibers (the epithelium, the fascia, and the endomysium) and the sarcolemma. The SEE was generally a tendon.

When the model did not contain tendons, the muscle unit can be replaced by a parallel structure where the force *FM* was the sum of the F^{CE} in the CE and the F^{PE} in the PE, which is shown in Eq. (2).

$$F^{M}=F^{CE}+F^{PE} \quad (2)$$

Among them, F^{CE} and F^{PE} were the forces generated by the shrinking units, the elastic unit, and the damping unit, respectively. The force F^{CE} generated by the contraction unit was determined by four factors, which were the degree of activation of the muscle, the length of the muscle, the rate of contraction, and the maximum isometric contraction force, as shown in Eq. (3).

$$F^{CE}=A(t)F_{l}(l)F_{v}(v)F_{\max} \quad (3)$$

Wherein, $A(t)$ was the activation curve of the muscle, $F_l(l)$ was the length curve of the muscle, $Fv(v)$ was the speed curve, Fmax was the maximum isometric contraction force of the muscle. The F_{max}, $F_l(l)$, and $F_v(v)$ were measured by a huge number of experiments on the target muscles (Mo et al., 2019). When the Hill model was applied to simulate the response of the active muscle, $A(t)$ was determined by experimental measurements of the reconstructed physical process. According to the experiments, the $A(t)$ curve was edited to realize active muscle contraction control (Jammes et al., 2017).

Stress distribution

The stress distribution in the ligaments and bones could mainly reflect the risk distribution of the vulnerable area in the ankle joint during the impact. The measurement locations of the long ligaments on the outer side of the ankle joint involved in this study were CL, Tca, and Tna. These ligaments were vulnerable parts of the ankle joint. Their deformation was obvious during the impact, and the microscopic distribution of stress was beneficial to the description of injury characteristics (Xiao et al., 2021). Especially, stress was an injury judgment index of the material failure in the ankle joint. The regional impact loading distribution and the force values were reflected by the stress in the joint from a micro view.

In the current study, the stress measurement locations of lower extremity bone were Cal, CBL, and CBM. Studying these parts could help to represent the ankle joint injury better. The stress distribution from these bones and ligaments near the ankle joint could reflect the forces and deformations of the individual bones. Moreover, they also could represent the most likely lesions, especially in bone locations. The stress distribution was a microscopic injury index and an indicator for explaining the injury mechanism.

Ankle bending angle and ligament elongation

Ankle bending angles and ligament elongations could reflect the injury level under the frontal impact in a full view. Specifically, the values' changing trend of the ligament could reflect the possible reason for the injury types. However, the ligament elongation and joint bending angle might not be changed in the same way. The active muscle function would influence the stretch forces in the muscle groups, which would lead to the change of the bending angle. Thus, the bending angle change during the impact could reflect the change in ligaments elongation of the joint in a macro view. The measurement position of the ankle bending angle involved in this study was the left ankle and the right ankle. In addition, CL, Tca, and Tna are used to measure ligament elongation.

Force/Moment distribution

The ligaments measuring locations for the measurement of force and moment distribution involved in this study were CT, CL, Tca, and Tna. The distribution of force was mainly determined by the distribution of several main ligament forces, usually were long ligaments located in the outer part of the joint. The ligament force changed during the entire foot rotation when a frontal collision occurred. Thus, the forces in the ligaments changed with time process with the change of the entire impact process. Meanwhile, the distribution of the ligament's moment could explain the torsion of each studied ligament. The ankle joint would be twisted when the lower extremity rotated relative to the heel. This twisted was because of the upward and backward movement of the pedal and foot. The torsional environment can cause the ligament rotational moment. The results can represent the change of the whole moment with time went during the impact process.

Simulation Matrix

Two types of models were conducted to investigate the difference and the mechanism of the joint injuries in this study. The corresponding relationship between a specific simulation and with or without the active muscle function in the following text was represented by simulation. In this study, the OA represented the model without the active muscle function, and WA meant a model with the function of active muscle. One simulation was conducted using the model with the active muscle functions and the other without the active muscle function. The active muscle in the study was only applied in the leg, especially the muscles which will act under the emergency brake condition. Meanwhile, the muscle functions in the foot were not significant. A 40% offset frontal impact simulation experiment was performed. And the boundary conditions of the load environment were adjusted according to a reference (Mo et al., 2018).

Results

Stress distribution

In the three ligaments like CL, Tcn, and Tna of the left ankle joint, the maximum stress was captured on CL which belonged to the WA. Furthermore, the peak stress on CL in the WA was significant, and the stress was 0.098 GPa (Fig. 2). And the peak stress on CL in the OA was 0.091 GPa, which was decreased by 7% compared with the model of WA. Besides CT, high stress location was also found on Tca. But the maximum stress of Tca in the two types of models was nearly the same, and there was 0.093 GPa. In the impact, CL and Tca were stretched, and Tna was non-stretched. From the results, it was observed that the stress of Tna was almost zero. The CL and Tca of the left ankle joint had the same distribution of high stress areas in the two types of models. Meanwhile, the high stress areas were concentrated between the ligament and the tibia junction or fibula junction. This phenomenon was because the deflexion angle decreased, and the tibia and fibula leaned forward relative to Cal during the whole impact process. Moreover, the ligaments were moved up and stretched at the tibiofibular junction. On the contrary, the ligament area far away from tibiofibular received relatively little tension, and even the tension force was near zero.

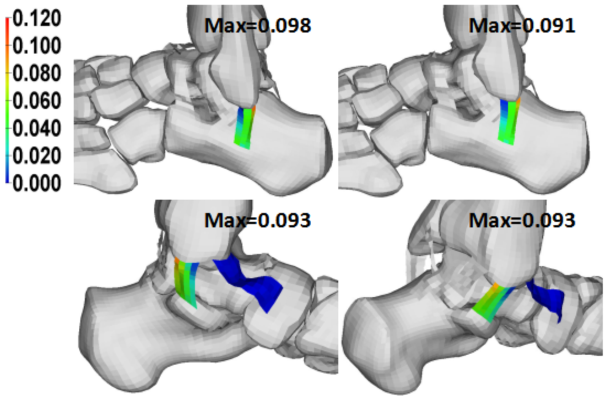


Fig. 2. The stress distribution in ligaments of left ankle /GPa (left: active muscle; right: inactive muscle).

The results obtained from two models with or without active muscle in the right foot were compared (Fig. 3).

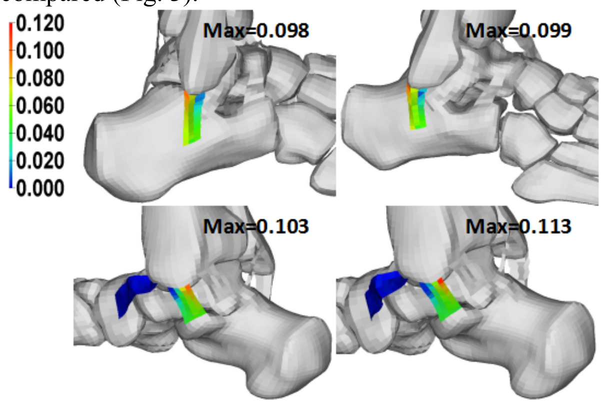
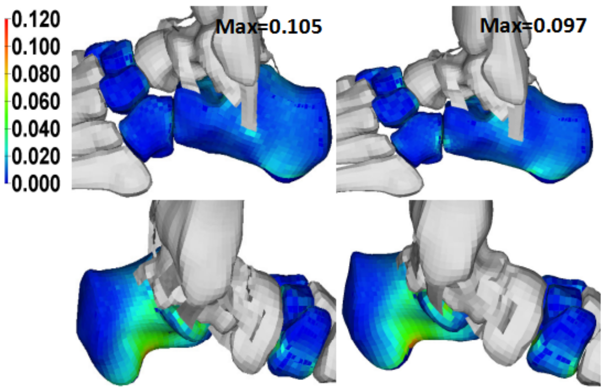


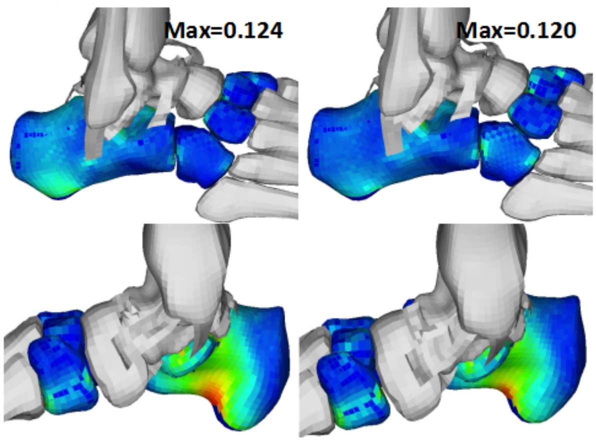
Fig. 3. The stress distribution in ligaments of right ankle /GPa (left: active muscle; right: inactive muscle).

Different from the ankle joint of the left leg, the maximum stress of right ankle ligaments was generated on Tca of OA, and the stress was 0.113 GPa. Meanwhile, the peak stress on Tca in the WA was 0.103 GPa, which was decreased by 9% compared with that without the active muscle function. From the results, it was observed that the stress of CL in the two types of models was nearly the same. And it was found that 0.098 GPa in the WA and 0.099 GPa in the OA, respectively. The stress gap of CL between the two models was only 1%. Similar to the ankle joint of the left leg, it was observed that the stress of Tna was almost zero. The cause of this phenomenon was ligament Tna located in the front of the ankle joint, and this ligament was compressed when the ankle joint rotated upward. As a result, the ligament was convex in space. In this case, the stress was smaller than others. The stress distribution of CL and Tca was the same as the left ankle joint. The stress distribution law of the CL and the Tca were the same as those in the former results.

Regarding the stress distribution of ankle bones, the stress values in the contact area between bones were great because the contact area was the force fulcrum (Fig. 4). In general, it was observed that the peak stress of the right ankle was higher than that of the left ankle. This result of the right ankle was caused by the vast invasion of the brake pedal. In the WA, the maximum stress was generated on right ankle, and the stress was 0.124 GPa. In addition, there was 0.120 GPa found on the right ankle in the OA. And right ankle of the OA was decreased by 3% compared with that of the WA. The stress comparison of the left ankle joint with or without active muscles was more obvious than the stress comparison of the right ankle joint.



(a) left ankle



(b) right ankle

Fig. 4. Stress distribution in bones of ankle /GPa (left: active muscle; right: inactive muscle).

It was captured that the stress of the left ankle joint was 0.105 GPa in the WA, and the stress of the right ankle joint was 0.097 GPa in the OA. And left ankle of the OA was decreased by 8% than that of the WA. Although the bone to bone contact area was protected by soft tissue, stress values were still relatively high. The specific path of force was transferred due to the structure of the foot. Meanwhile, the large internal force area of plantar passes through the sesamoid bone, and most of the forces were transferred to the Cal. Therefore, Cal was in a high stress situation.

Ankle bending angle and ligament elongation

From the measurement results, the initial bending angles of the left ankle and the right ankle were the same, and both were 97˚ (Fig. 5). In general, it was found that the change in the bending angle of the right ankle was more significant than that of the left ankle. The bending angle of the left ankle changed from 97˚ to 78˚ in the two models. And the bending angle of the right ankle changed from 97˚ to 76˚ in the OA. Moreover, the bending angle of the right ankle changed from 97˚ to 75˚ in the WA. During the impact, the bending angle of the left ankle was reduced by 19˚ in the two models. In addition, in the OA, the bending angle of the right ankle was lowered by 21˚. Also, in the WA, the bending angle of the right ankle was lowered by 22˚. The bending angle of the right ankle was increased by 11% in the OA compared to the left ankle. In the WA, the bending angle of the right ankle was increased by 16% than that of the left ankle.

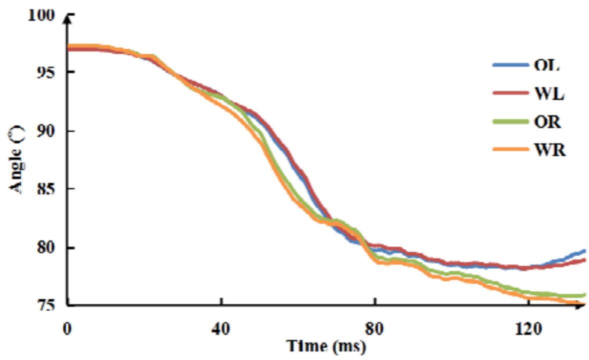


Fig. 5. Ankle bending angle (L: left ankle; R: right ankle; O: inactive muscle; W: active muscle).

The CL and Tca were mainly subjected to tension in the impact, but Tna was mainly subjected to compression. So the ligament elongation of CT and Tca at the ankle joint of the left foot showed the same trend during the impact process (Fig. 6). In addition, CL, Tna, and Tca were found the same trend in the two types of models. Between 0 ms and

40 ms, it was found that both CL and Tca contracted before elongation. The stretch and contraction of the ankle ligaments were both within 6.0 mm, so the observed difference in the effect of active muscles on the elongation of the ligaments was insignificant. During the whole process of simulation, the maximum interpolation of CL in both active and inactive muscle models occurred in 81 ms, and the difference was about 0.39 mm. The maximum interpolation of Tna occurred in 67 ms, and the difference was about 0.30 mm. And the maximum interpolation of Tca occurred in 54 ms, and the difference was about 0.23 mm. Through the data analysis, it could be obtained that the elongation of CL in the WA was generally larger than that of in the OA. On the contrary, the elongation of Tca in the W model was generally smaller than that of in the OA. However, there was no such discovery in Tna. It was captured that with or without active muscles had a small effect on the change in left ankle angle, so the change in ligament elongation was also small.

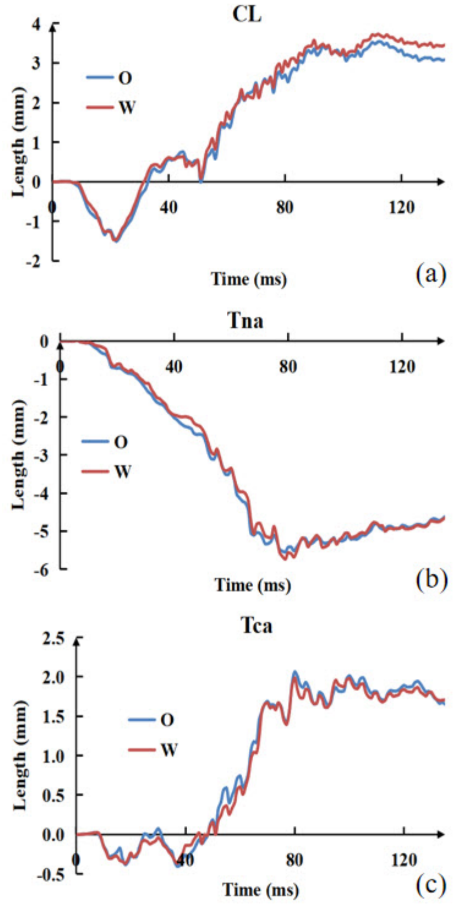


Fig. 6. Ligament elongation in left ankle (O: inactive muscle; W: active muscle).

It could be seen that the elongation of the right ankle ligament had a similar trend to that of the left ankle (Fig. 7). However, it was captured that with or without active muscles had a more significant change in the right ankle angle than that of the left ankle.

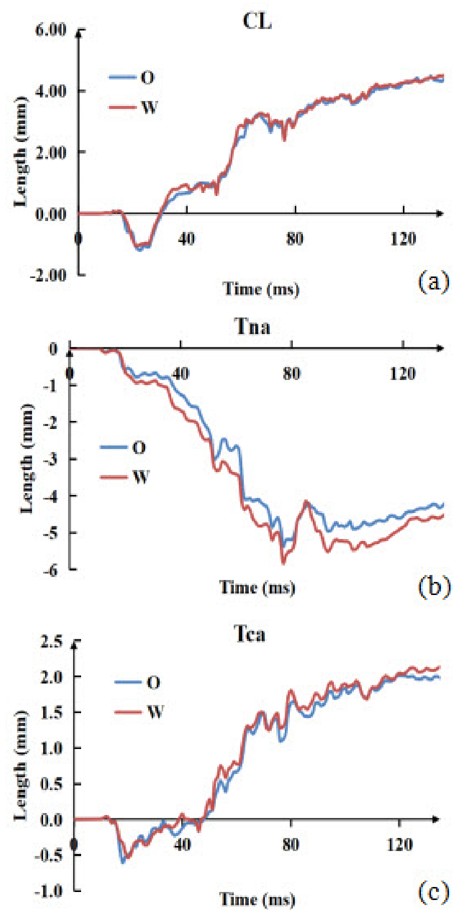


Fig. 7. Ligament elongation in right ankle (O: inactive muscle; W: active muscle).

So the elongation of the right ankle ligament was not the same as the left ankle ligament. Similar to the left ankle, the right ankle was also found that CL, Tna, and Tca had the same trend in the two types of models. And between 0 ms and 40 ms, it was found that both CL and Tca contracted before elongation. In the OA, the elongation of CL ranged from -1.19 mm to 4.47 mm, and it was found to have changed by 5.66 mm. The elongation of CL ranged from -1.05 mm to 4.49 mm in the WA, and it was shown to have changed by 5.54 mm. In the OA, Tna's elongation was discovered to have changed by 5.37 mm. In the WA, Tna's elongation was changed by 5.83 mm. However, Tca's elongation ranged from -0.61 mm to 2.01 mm in the OA, and it was discovered to have changed by 2.62 mm. The elongation of Tca ranged from -0.54 mm to 2.13 mm in the WA, and it was observed to have changed by 2.67 mm. During the whole process of simulation, the maximum interpolation of CL in both active and inactive muscle models occurred in 59 ms, and the difference was about 0.39 mm. The maximum interpolation of Tna occurred in 60 ms, and the difference was about 0.77 mm. And the maximum interpolation of Tca occurred in 18 ms, and the difference was about 0.28 mm. In the right ankle joint, the interpolation of Tna was greater than that of CL and Tca in the two types of models. This may be due to the bending angle in right ankle of the WA being greater than that of the OA. Moreover, Tna was the ligament on the front of the ankle, which might be more affected by the change of the ankle's bending angle than that of the lateral ligament.

Ligament forces

For the drivers, only right side of the pedals was used to brake. The forces of ligaments could represent the extension or compression situation in the ankle joint during the impact. In the right leg, the axial forces in primary ligaments were recorded, which were CT, CL, Tna, and Tca.

The active and inactive muscle models were compared among the ligament's force results (Fig. 8). The maximum of all the peak forces was found in Tca with active muscle function, which was around 0.512 kN. The biggest gap between the two types of models (active and inactive) was captured in CL, which was around 0.068 kN. Data analysis showed that the ligament force of CT and Tna is relatively small and fluctuates significantly over time. The CT was different from other ligaments, and ligaments were connected bone to bone. However, CT was formed by the merging of fibers of the gastrocnemius and the soleus muscles, forming the tendon that inserted into the posterosuperior aspect of the calcaneus. Due to this characteristic of CT, when the force reaches the peak, the force might decrease under the muscle's buffer. There was a rapid increase at 40 ms after the initial car impact in Tna axial forces caused by the pedal intrusion. The cause of these results might be that the ligament of Tna was in the ankle, and the middle part of the ligament was bulged up after intensive compression, which could reduce the axial force. Due to the great length of Tna, its shape changed obviously during the impact, so the changing trends of the force curves in the two types of models were quite different. Because the pedal could not rebound automatically in both types of models, the axial forces of CL and Tca were still stretched after reaching the peak value. So the peak value remained unchanged.

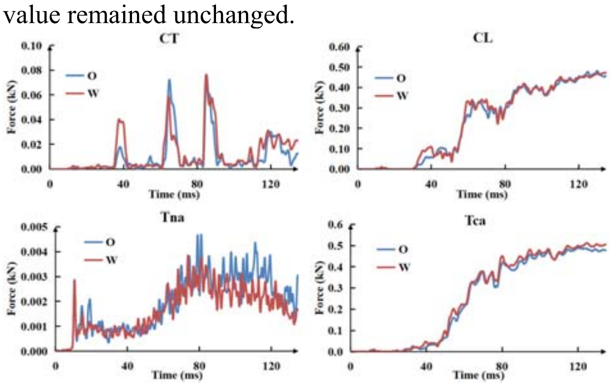


Fig. 8. The forces of ligaments in the right ankle (O: inactive muscle; W: active muscle).

The changing trends in the forces of CL and Tca were the same. The axial force continued to rise at 80 ms after the initial impact. The peak force of CL was 0.481 kN in the OA. And the peak force of CL was 0.473 kN in the WA, which was decreased by 2% compared with that of without the active muscle function. In addition, the peak force of Tca was 0.489 kN in the OA. And the peak force of Tca was 0.512 kN in the WA, which was increased by 5% compared with that without the active muscle function.

Ligament moments

The ligaments moments were captured in the right leg due to that the braking action only happened on this side, and the pedals in the right might heavily affect the lower extremity injury outcomes (Fig. 9). The moment of ligaments around the ankle joint could show the rotation or bending consequences during the crash. There were four measuring locations of ligaments in the ankle (CT, CL, Tna, and Tca). Among these ligaments, the results were compared between the active muscle model and the inactive muscle model. The greatest peak moment was found in CT in the inactive muscle model, and the value of this moment was around 0.824 Nm. The smallest peak moment value happened in Tna with active muscle function, which was around 0.015 Nm. The greatest moment gap between the two types of models happened in CT, which was around 0.412 Nm.

The changing trends in the moments of CL and Tca were the same. It was found that the peak moment of CL was 0.677 Nm in the OA. And the peak force of CL was 0.497 Nm in the WA, which changed 0.180 Nm. Moreover, it was found that the peak moment of Tca was 0.601 Nm in the OA. And the peak force of Tca was 0.612 Nm in the WA, which changed 0.011 Nm. The CT and Tna were the same in the changing trend of ligament force and moment under active or inactive muscle conditions. The factors affecting the moment changes of CT and Tna could refer to the force results.

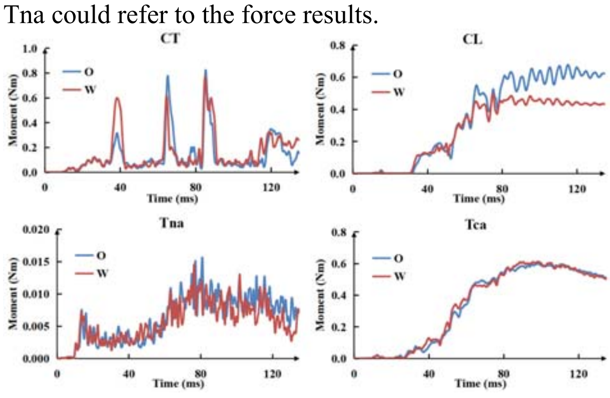


Fig. 9. The moment of ligaments in the right ankle (O: inactive muscle; W: active muscle).

Discussion

Injury difference

The ligaments CL, Tna, and Tca, are all long lateral ligaments around the ankle joint. And CT is the tendon inserted into the posterosuperior aspect of the calcaneus. When the upward rotation of the ankle joint increases, the ligament length change is mostly changed along with axial stretching. The peak stress of the left ankle ligament decreases, and the peak stress of the right ankle ligament increases due to the active muscle function. Moreover, the peak stress of the related bones decreases. Combining the kinematics during the impact and the ankle joint structure, the possible reasons for this phenomenon are as follows. The pedal exerts an upward force on foot due to the invasion of the pedal. But the active muscle function increases the upward bending angle of the ankle joint. Thereby, the long lateral ligament deformation would increase, and the peak stress would increase. The active muscle force reduces the impact of the pedal on the ankle joint reduces the stress on the bones. Due to the movement of the driver, the amplitude of the injury outcomes will fluctuate regularly during the impact.

During the impact, the angle of the ankle changes significantly. Specifically, due to the invasion of the brake pedal, the angle of the right ankle changes more than the left ankle. From the results, the stress, elongation, force, and moment of CL and Tca are significant. Because there are CL and Tca stretched during the impact. However, Tna is mainly compressed during the impact. When the upward rotation angle of the ankle joint increases, the Tna geometric shape changes complex. Therefore, the stress, force, and moment of Tna fluctuate greatly.

Indicators all reflect that active muscle force affects the ankle. However, the effect is not significant. The effect of active muscle force on the ankle will not differ by order of magnitude. First, the elongation of the ligament is in millimeters. In addition, the invasion of the pedal makes the ankle angle change. Compared with the invasion of the pedal, the change of the ankle angle by the active muscle force is slight. The above two factors explain why the results are not significantly different in the two models.

Source of results difference

Judging from the IIHS tests, the layout of the vehicle IP structure may heavily affect the lower extremities injury outcomes to the drivers (Jindal et al., 2018). Particularly, the injury of the lower extremities is always serious in the MOI. The previous injury evaluation method does not consider the influence of ankle rotation on injury outcomes. Thus, the injury evaluation index in the ankle is not fully considered as the serious outcome in the regulation condition. However, ankle injury occupies a great percentage of the injuries in nowadays studies. Thus, the joint injury study along the lower extremities under active muscle should be carefully considered (Karagiannakis et al., 2020; Chen et al., 2020).

Different people may have different muscle reaction times and reaction levels to the same impact. Meanwhile, sometimes the drivers may not be sensitive to the emergency due to gender or mental health. Thus, an EMG (Electromyography) signal may reflect the whole trend but not be applied for all the people (Yasuki et al., 2010), especially for someone drunk or old (respond slowly to the danger). Thus, the detailed EMG singles may require certain people in the follow-up studies.

Differences by biomaterials

With the increase of driver ages, the bone material of lower extremity will change accordingly, though the structure change may not be much. Thus, the influence of the ages on the injury outcomes is small in terms of the change in joint angle and ligament elongation (Gaewsky et al., 2015). However, for previous studies, the bones will be influenced by ages due to the change of material properties (Liu et al., 2021). Thus, if the influence of age is studied, the change of the bone factors should also be involved. Thus, defining the influence of age on the body material is very important to this study. This means that in the continuous study, the influence of the biomaterials parameters should be more accurate to get better results.

Future study

The parameters of the lower extremity muscles function, which is the key point to define a motion for the model (Charles et al., 2019), are used in this study through the activity level is only a unique selected one of the usual responses of the drivers. However, age and health conditions that can be used to conduct the injury difference study can reflect the human emergency responses (Mair et al., 2019). Therefore, more impact conditions with different activity levels should be conducted to make the study results widely applied in biomechanical studies. In the current model, only the muscle groups that work during emergency braking were activated in this study. In addition, the brake pedal cannot rebound automatically, so the calculation results in the later time remain in a stable value.

In crash environments, the muscle function level may differ obviously. The impact velocities vary in real traffic accidents, which means the impact energy will be different. Thus, a more detailed study should be utilized to investigate the influence of impact speed on biochemical outcomes. Moreover, airbags, knee bolsters, and other protective equipment may also influence ankle injuries (Nie et al., 2018), especially since these facilities are usually applied to reduce injuries to the lower extremity in luxury cars. In addition, passengers can be protected from crash impacts by changing the design of the engine room to improve the absorption of crash energy (Kim et al., 2012). Due to the variation of the load paths through the leg, the injury loads are not all involved in the study in different impact conditions, which do not represent the whole loading conditions of the accident environment.

CONCLUSIONS

The driver model with a detailed lower extremity is used under MOI conditions to investigate the different responses of ankle with or without active muscle. The injury differences affected by active muscle can explain the injury mechanism and potential risk of the ankle joint, which can be utilized in the follower studies. According to the current study, the load condition of the lower extremities will change due to the active muscle function. In detail, the peak stress of the left ankle ligament decreases, and the peak stress of the right ankle ligament increases due to the active muscle function. Moreover, the peak stress on the surrounding bones decreases. The elongation of the right Tca increases by 6%, and the compression of Tna increases by 8% with the existence of active muscle. In addition, the greatest peak force of the right leg ankle ligament was around 0.512 kN in the WA. And the greatest peak moment of the right leg ankle ligament was around 0.824 Nm in the OA.

There are three novel bullets in the current study. First, the numerical lower extremities model is a newly established model referring to the data of Chinese figures. Second, both the macro index like bending angles and micro index like stress distribution is used in the study. Third, the function of the active muscle was used in the driver's lower extremities to distinguish the influence on the ankle joint kinetic responses.

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NOMENCLATURE

F^{CE} Muscle contraction force of contraction element

V Contraction speed

F_{max} Maximum isometric contraction force

F^{PE} Muscle contraction force of parallel elastic element

$A(t)$ Activation curve of the muscle

$F_l(l)$ Length curve of the muscle

$F_v(v)$ Speed curve

正面撞擊過程中緊急制動時踝關節響應的調查

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摘 要

本研究旨在為正面撞擊時下肢損傷預防的研究提供參考。首先，建立汽車模型的邊界條件。然後，通過內部部件的侵入和加速特性，模擬了一次實車事故的全過程，通過緊急制動下關節周圍韌帶動力學等損傷指標確定了踝關節損傷。結果表明，在沒有主動肌肉功能的模型中，右腳踝的彎曲角度比左腳踝增加了 11%。在肌肉功能活躍的模型中，右腳踝的彎曲角度比左腳踝增加了 16%。本研究調查了在 40%偏置正面衝擊下由於下肢活動肌肉功能導致的踝關節動力學特征的差異。