SENSITIVITY AND LOAD PATH ANALYSIS FOR THE H-III IN FRONTAL IMPACT

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ABSTRACT

The main objective of the paper is to develop an analysis method of the mechanisms that controls the behavior of the H-III neck, thorax, and lower extremity injuries in a USNCAP and Euro-NCAP frontal impact. The analysis method will be utilized within the engineering design of safety systems to obtain optimal injury values. For this research were conducted in 5 steps.

Step1. Load path analysis based on numerical simulations, crash tests and 6 sled tests of various conditions with extended instrumentation (ex. Angular rate sensors, Rib-Eye). The numerical models were validated with the sled test data, to allow analysis of the load path mechanisms.

Step2. Sensitivity analysis of the safety system and dummy sub-systems with validated models. The sub-system simulation study was conducted in detail for finding out physics of the load paths mechanism and the sensitivity of the injury value characteristics.

Step3. It was going to a systematic approach to injury mechanism through the kinematics. Then, relations between kinematic and physical load paths were characterized.

Step4. Details analyze the effects on each part for various pulse and safety restraint components. Then it will be showed effectiveness guidelines of various safety restraint components.

Step5. 4 sled test for confirmation.

Finally, the study resulted in identification of the mechanisms that mainly affect neck, thorax and lower extremities injury values. Based on the mechanism analysis, design guidelines could be help to safety system design of the target performance.

INTRODUCTION

Recently as the requirements of frontal crash became more strict, especially, the neck and thorax injuries of H-III 5th in passenger side became more challenging than the others. Moreover, the NHTSA's introduction of the new NCAP 5-star rating system [1] starting with 2011 MY vehicles put even higher demands to the safety system development than before. According to this rule, especially, the improvement of the neck injury is very important to achieve a 5-star rating for passenger. But it is difficult to know what is the best improvement method, what is exact the injury mechanism. Because dummy movement depends on many complex variables and limited conditions can be tested. Also the field significance of the neck injury mechanism is appropriately reflected relative to the more prominent roles of the head and thorax [2]. So it is necessary to the analysis the main effect factors and contribution related to dummy injuries by advanced tests and simulations.

On tibia injury, in 1996, the European Community released new 40% offset crash – it is consist of crashing car on a ODB (Offset Deformable Barrier) at 56 km/h - relating to frontal impact vehicle crash. It has increased the importance of low extremities injury such as tibia injury.[3] Furthermore, IIHS of US, EuroNCAP, KNCAP, CNCAP and even Asian NCAP, introduced 64kph 40% offset crash at the same time. So, importance of reducing low extremities injury grown with respect to the other system requirements.[4] One of the main reasons for increasing importance of low extremities injury is the most frequent and costly consequences of those injuries for the survivors of crashes. Therefore, the insurance claims for vehicle occupants whose most serious injury was a fracture of a weight-bearing bone cost $2.06 billion every year [5-6]. In addition to direct medical costs, lower extremity injuries were associated with high incidences of long-term problems that sometimes require additional treatment and interfere with patients’ ability to return to work [6]. So, it is very important that reducing low extremities injuries and developing effective protection system for low extremities injuries.

This paper described about the dummy injury mechanisms for neck, chest and low extremities and their kinematics by using various CAE tools and test methods. These results should be useful to understand the H-III 5th and H-III 50th injuries guide the safety restraints development and interior vehicle package.

To figure out injury mechanism, 6 sled tests were conducted for mid-sized vehicles with 7 cameras and enhanced measurements. These tests were used to obtain data for dummy injury mechanism analysis and also to obtain data set for MADYMO/LS-DYNA model validation. The test simulated 64kph offset and 56kph full frontal impact each 3 times. All the tests were conducted with a 50%tile male driver and a 5%tile female passenger. Furthermore the dummies were installed with enhanced measurement sensors which were 6ch lower neck load cell, 6 belt load cell, 2 belt spool sensors, rib eyes, head/chest/pelvis/foot 3 axis angular sensor each part , 2 angular sensors in the tibia, and foot load cells. (see figure 1)

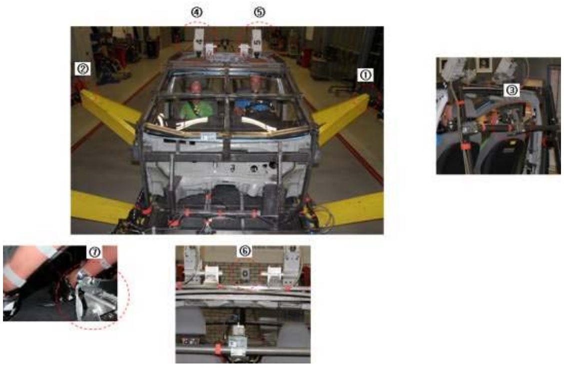

Figure 1. Sled test setup.

LOAD PATH ANALYSIS

Load Path of Head and Neck

Basically, neck injury mainly occurred from various moments by flexion and extension bending and compression force. Furthermore, the Nij criteria which are a function of upper neck forces and moment, play a dominant role in the crash performance star rating under the US NCAP. Therefore, we need to analyze this injury mechanism.

The Nij in US NCAP is defined as equation (1) and Table 1.

$$\text{Nij} = \left[\frac{F_z}{F_{zc}}\right] + \left[\frac{M_{ocy}}{M_{yc}}\right], M_{ocy} = M_y - (F_x \times l) \qquad (1)$$

Table 1.
Summary of Neck injury criteria value

		Sign	50%ile	5%tile
F_{zc}	Tension(N)	(+)	6806	4287
	Compression(N)	(-)	6160	3880
M_{yc}	Flexion(Nm)	(+)	310	155
	Extension(Nm)	(-)	135	67

The dummy neck and upper neck sensor are composed as figure 2 and sign conventions are followed by J211 standard.

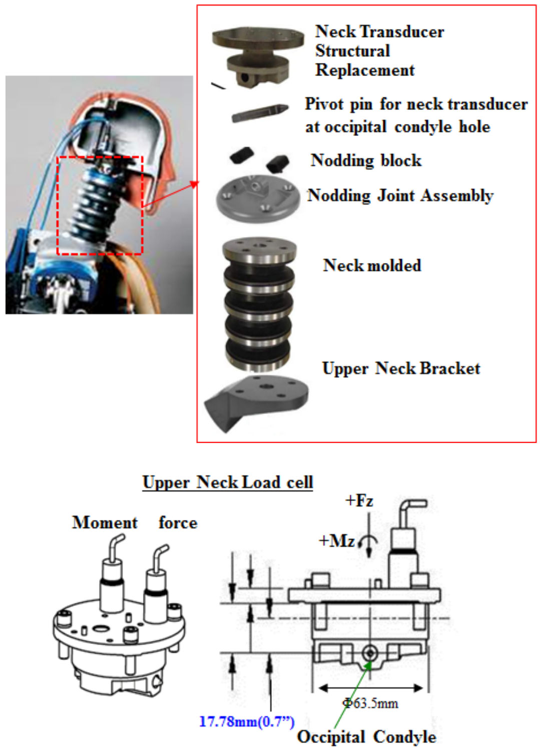

Figure2. Dummy neck and upper neck load cell structure.

The Head/Neck dynamics is mainly affected by airbag and seat belt. However the effects of inertia force cannot be passed over. Therefore the force source of head/neck can be described as figure 3.

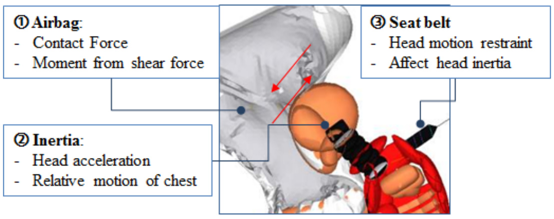

Figure 3. Load path diagram for the head.

Generally, the neck injury pattern in an NCAP test can be divided into 3 phases as figure 4. Phase 1 is seat belt only affected phases, which is occurring before the head contacts the airbag. Phase 2 is affected airbag and belt phase, which is occurring during the principal head loading phase by the airbag. Phase 3 is the rebound phase. So external forces, such as airbag pressure and belt force are gradually tapered off.

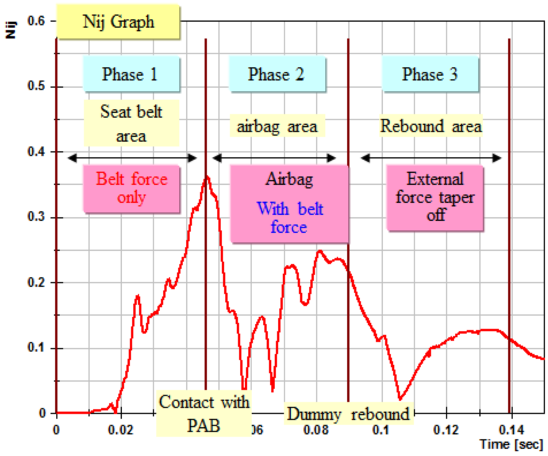

Figure 4. Typical Nij trajectory for 5%tile neck.

M_{ocy} graph is very similar to head angle trajectory. But it has some difference tendency (see figure 5). Because, when head contacts the airbag, an added momentum results from the contact force in x direction. Furthermore it also has a nonlinear characteristic property for nodding block which is composed of rubber.

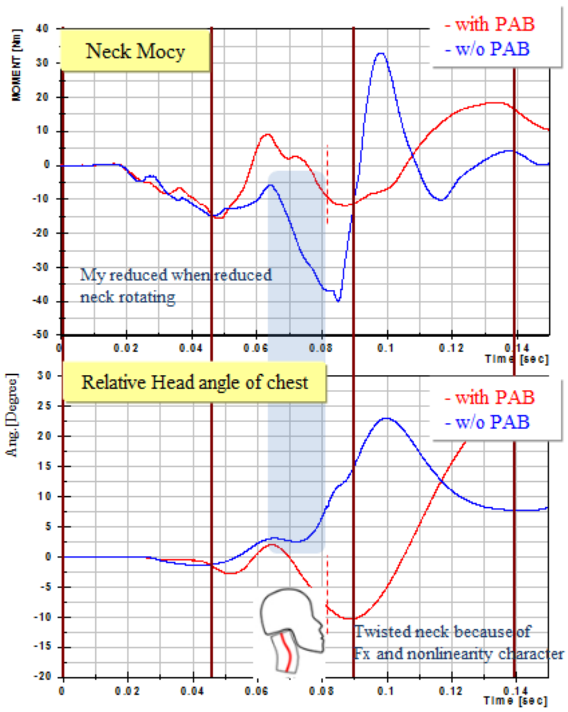

Figure 5. Neck M_{ocy} and relative head angle of chest.

In phase 1, when chest is caught by the seat belt, the head is moved relative forward due to head inertia. At that time, in the neck is occurring a tension force (positive z direction) and an extension moment by this relative head motion of the chest. The tension force (or compression force) of neck is depends on relative head-acceleration of the chest and relative head angle.

In phase 2, there is an additional main interaction area between head and airbag with respect to phase 1. According to airbag shape and pressure distribution a force balance exist among airbag load, head inertia and neck load. In which the neck loads arise from differences in the relative motion of the head with the chest. How three types of airbag design affect the neck loads is explained in figure 7b.

In case of type 2 the airbag generates forces that balance the neck loads such that the head is pushed to follow the thorax motion in the most natural way. In the other two cases the forces do not balances well and the neck gets loaded and will deform. Normally, when chin caught by the airbag or face contacted by asymmetry airbag, it will be visible in the neck load.

In phase 3, it is head and chest rebound phase. Therefore sign of neck moment (M_y) is changed from negative to positive. Sometimes, the head has a hard contact with the headrest, B-pillar or other hard part, so that it causes a high Nij. But, this paper will not deal that kind of special conditions.

In most cases the chest is rebounding earlier as the head. When chest is rebounded, the upper neck is still moving forward with the head as airbag keeps venting. This will results in flexion of the neck. After that, head rebounding will be started. (see the figure 6)

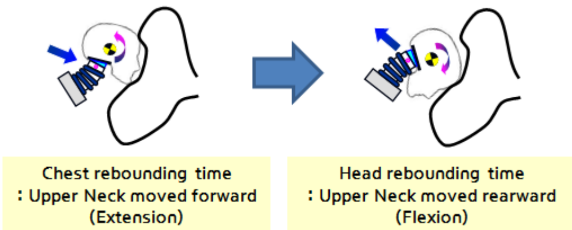

Figure 6. Dummy head movement on phase 3.

The F_z force depends on the vehicles pulse severity. If vehicles pulse severity is high, chest of dummy will be rebounded strongly. In that case, there will be tension force occurred on the neck. On the other hands, if the head rebounds earlier before the chest can be the compression force on the neck.

Finally, the neck injury mechanism was summarized for each phase in Figure 7(a), Figure 7(b) and Figure 7(c).

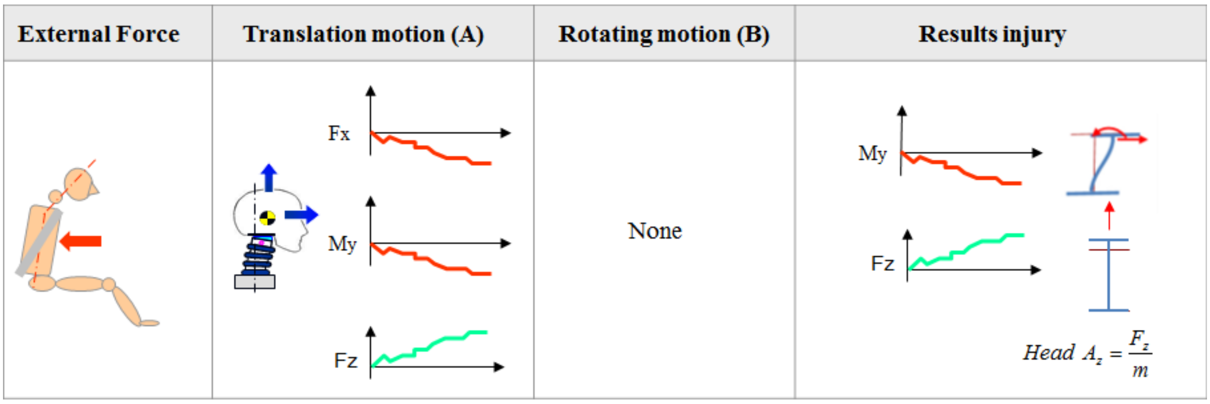

Figure 7(a). Neck Injury mechanism on phase 1.

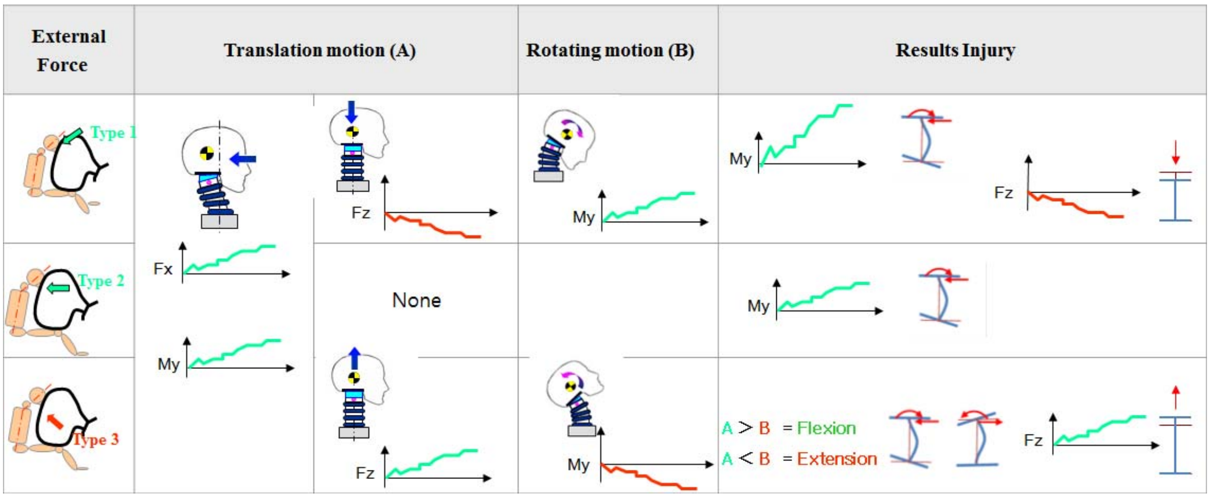

Figure 7(b). Neck Injury mechanism on phase 2 by airbag loading.

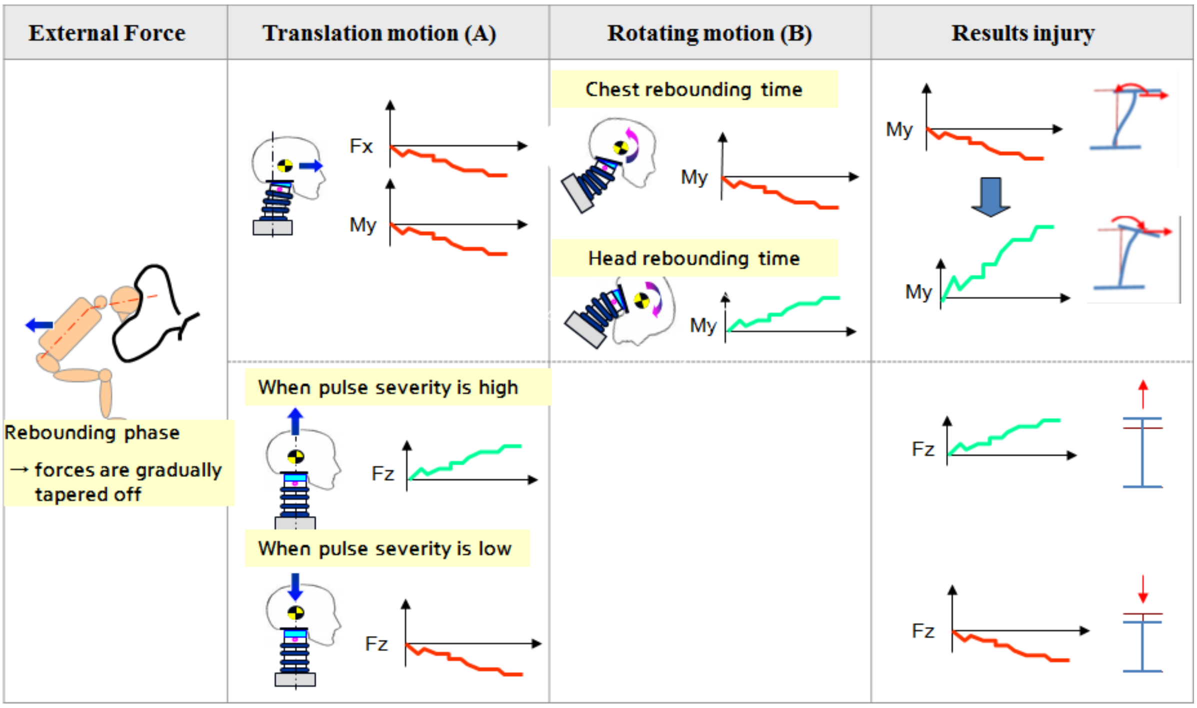

Figure 7(c). Neck Injury mechanism on phase 3.

Load Path of the Chest

Chest injuries are represented by the 3ms peak acceleration and deflection. The chest deflection of the H-III dummy is measured on just one point at the middle of thorax. The thorax injury mechanism is very simple and clear. Because thorax has no joint itself except connecting point between neck and pelvis.(see Figure 8) Therefore, it will be carry out more detail analysis at the sensitivity analysis section.

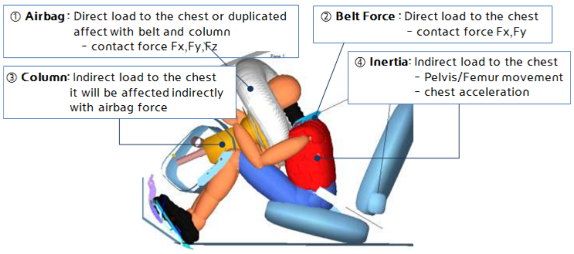

Figure 8. Load path diagram of thorax.

Load Path of Low Extremities

Tibia injury mechanisms are very complicated to understand, quantify and summarize. First of all, when you want to understand the Tibia injury mechanism in detail, the sign convention of the Tibia needs to be clear. Basically, it is followed by J1733 standard. (see Figure 9)

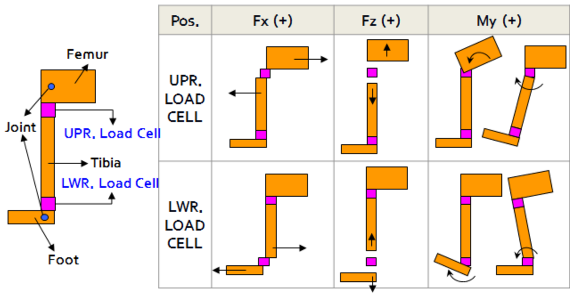

Figure 9. Sign convention of the Tibia.

Tibia has two joints, one is the knee between femur and upper tibia and the other is the ankle between lower tibia and foot. The sign of tibia sensor signals depend on where the external force is applied (below or above the sensor). So, we made a simple tibia model used LS-DYNA. (see Figure 10) Then we could find out exact sign convention depends on the external load position.

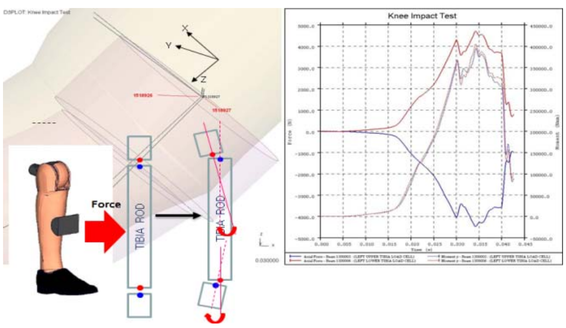

Figure 10. Simple tibia model for checking sign convention depends on external force position.

In case of a force applied by the femur, for example, upper tibia F_x and lower M_y have the same sign, in case of force to upper tibia part(below sensor), there is upper F_x and lower M_y injury occurred with opposite signs. It means that upper F_x and lower M_y occurred simultaneously. Lower F_x and upper M_y either. According to our sled test results, basically upper F_x and lower My had a similar injury pattern. However, when it had an external force, the F_x magnitude increased.(see Figure 11)

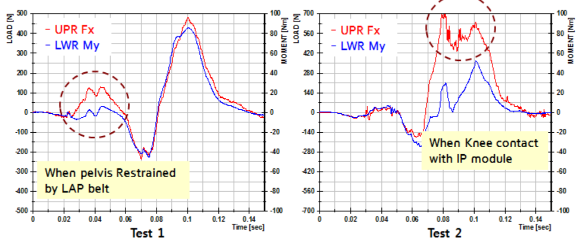

Figure 11. Upper F_x and Lower M_y graph pattern.

During the crash the tibia loading has three phases which are forward, rotation and rebound phase. A summary of the tibia injury mechanism can be found in Figure 12(a),(b),(c) based on the test results analysis.

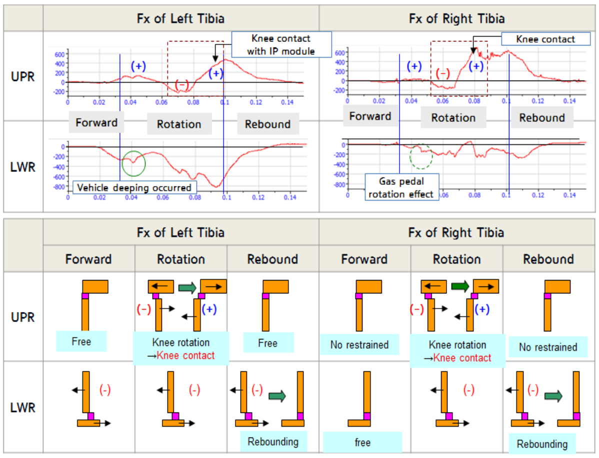

Figure 12(a). Tibia F_x injury occurred trajectory.

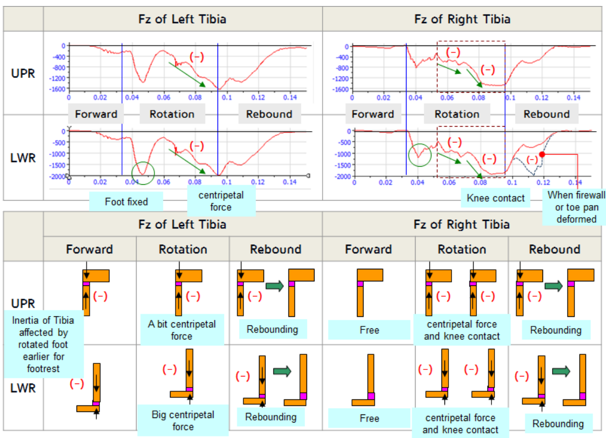

Figure 12(b). Tibia F_z injury occurred trajectory.

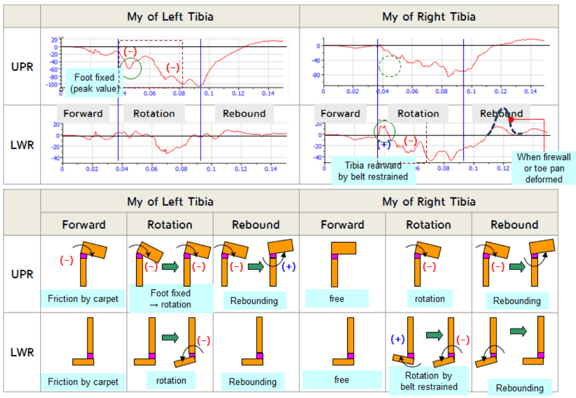

Figure 12(c). Tibia M_y injury occurred trajectory

SENSITIVITY ANALYSIS

This section objective is sensitivity analysis of the dummy response for variation in the principal load paths. Therefore, it was followed below process;

- Use correlated model as reference
- Define principal levels and parameters of the load paths
- Define modelling method to evaluate the variations
- Perform DoE or parameter variation studies
- Analyse the results

The following software were used: MADYMO 7.4.1, Hyper study, and LS-DYNA for this analysis.

Sensitivity Analysis of the Neck

This study was conducted based on passenger side with H-III 5th female dummy. Basically, the pick neck loads occurred before the head is fully loaded by the airbag. However in some cases the during airbag ride down increased extension moment can be occurred. This moment could be caused by shear force and normal contact force between dummy head and airbag. Therefore, it was conducted a DoE by CAE analysis in two difference conditions (with and without airbag) as Table 2(a), (b).

Table 2(a).
Variables for full scale dummy analysis at Phase 1

No	Loading condition	Level	Remark
1	Pretension Force	2	2.0/3.0 kN
2	Pretension damping	3	0,40,80
3	Seat Stiffness	3	16%,100%,183%
4	Buckle torsion angle	2	5.7°,17°

Table 2(b).
Variables for component scale analysis at Phase 2

No	Loading condition	Level	Remark
1	Body Pulse	2	56kph,64kph
2	Airbag pressure	2	10% increased
3	Airbag venting	3	Enlargement
4	head contact height	3	0, 20, 40mm
5	head rotation	2	0.95 and 1.0

According to the results, the first peak neck load was caused by belt pretension force that pushed the dummy into the seat. It means that it could be reduced most effectively by increasing seat stiffness. Secondary a less aggressive (slower) pretensioner or more stalk rotation could reduce neck loads of the Phase-1. (see Figure 13 (a))

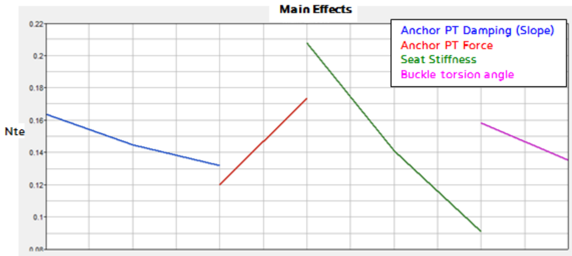

Figure 13(a). Sensitivity analysis results at Phase-1.

In Phase-2, the thorax deceleration by the belt caused the initial extension moments on the neck. The airbag interaction results in a the flexion moments on the neck and it makes the neck most sensitive to the airbag stiffness. Furthermore, the effects of the head position, and travel path were secondary to the airbag stiffness. (see figure 13(b))

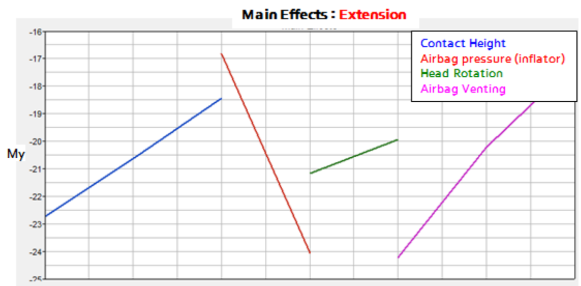

Figure 13(b). Sensitivity analysis results at Phase-2.

Sensitivity Analysis of the Chest

This study was conducted based on driver side of H-III 50th male dummy. This analysis was conducted at a thorax component level with a static status of the thorax. (see Figure 14) Also this research was conducted into two separate parts; one with belt load only and one with airbag and belt loading.

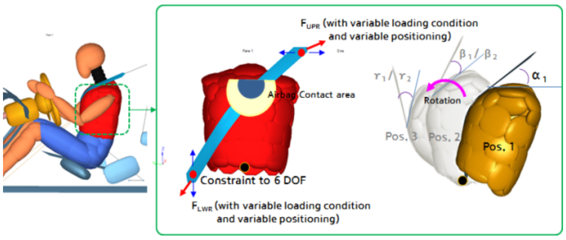

Figure 14. Variables and conditions for chest sub-system analysis

In case of belt force analysis were conducted for three positions according to the three phases; initial stage (phase 1), during airbag ride down (phase 2) and rebounding (phase 3). (see Table 3(a),(b) and Figure 15) But airbag and belt loading area was conducted at phase 2 only. Because airbag does not affect at phase 1 and minimally in phase 3.

Table 3(a)
Variable matrix of belt loading at each condition

No	Airbag Condition	Variables	Remark
1	Airbag Force per Contact increment/ellipsoid	0-150 N	6 level
2	Contact Area Size	5/10	2 level
3	Contact Plane Angle	0,xx	2 level

Table 3(b)
Variable matrix of airbag loading with belt

No	Belt Condition	Variables			Remark
		Case1 (Initial Position)	Case2 (Pos. 70ms)	Case 3 (Pos. 100ms)	
1	Upper Belt Loading – F_{UPR}	2.5 kN, 3.0 kN	←	←	2 level
2	Lower Belt Loading – F_{LWR}	2.5 kN, 3.0 kN	←	←	2 level
3	Upper Angle (D-ring Z position)- α	Ref, +40mm	β_1/β_2	γ_1/γ_2	2 level
4	Lower Angle (Buckle x-rotation)	0, 23°	←	←	2 Level
4	UPR Belt Position – D-ring Y (Belt Fitting)	0 , 50mm (Inboard)	←	←	2 level
5	LWR Belt Position – Buckle Z (Belt Fitting)	0, 30mm (higher)	←	←	2 level

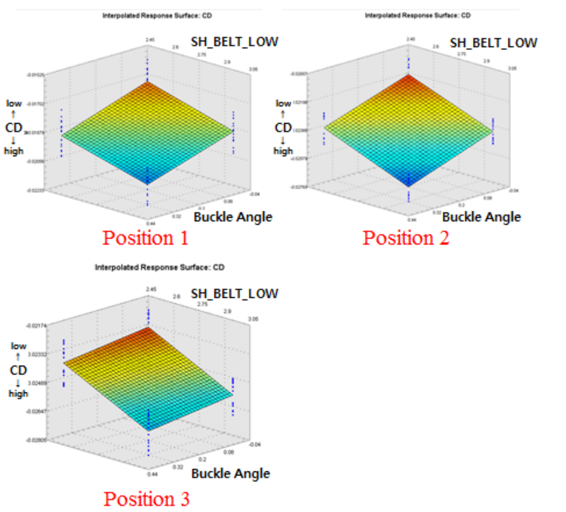

Figure 15. Sensitivity analysis results for belt loading of the chest.

In phase 1 and position 1, the chest deflection was mainly affected by the buckle torsion angle and the upper belt force. These variables continued to affect at position 2 and 3 in Phase 1. But the biggest effect was the lower shoulder belt force followed by the buckle angle and upper shoulder belt force. According to the research, the force component in the compression direction was increased during the ride down as the belt angle changes with respect to the thorax. This means that even though the same belt force at the chest, chest deflection was more increased when chest was in more ride down position.

In phase 2 studies, understandably, the added airbag loading increased the chest deflection. Reducing the load of the airbag on the chest will also reduce the chest deflection. Furthermore the simulation indicates that the chest was more sensitive to the load on the top as a load spread over a wider chest area.(see Figure 16)

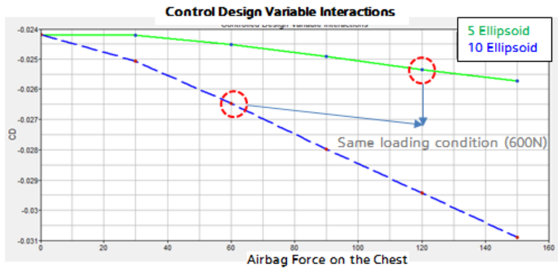

Figure 16. Sensitivity analysis results at airbag loading with belt force.

Sensitivity Analysis of the Tibia

In tibia sensitive study, the full dummy model of LS-DYNA was used to verify effectiveness of the inertia load. Therefore, we were comparing to injury value pattern between basic model and changed condition model at each conditions. (see Figure 17)

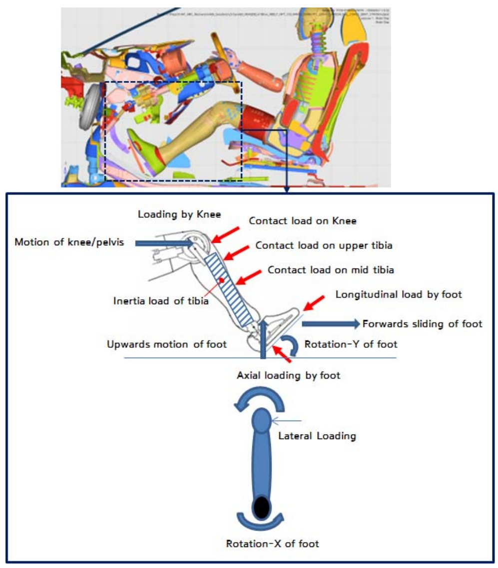

Figure 17. Various loading condition of the tibia.

According to this study, tibia loads were strongly to changes in the tibia mass. This is indicating that tibia loads are caused mainly by tibia deceleration and not by loads coming from upper body. Therefore the controlling of the tibia will improve the tibia loads. Basically, tibia acceleration is controlled mainly by pelvis and foot motion. Also the tibia contact with lower IP generates a load that creates a balance force for the tibia inertia load which is reducing M_y in the tibias. It means that tibia kinematics is controlled by IP, femur and foot. Typically contact force with IP should be low such that they cannot affect much the pelvis motion. So the IP design will not affect much the overall tibia kinematics, but it can be bring more the load balance over the tibia. On the lower side, the foot motion is affected by acceleration pedal, pedal arm, foot stopper, toe board padding, and floor carpet. As soon as the foot starts to rotate the compression force will generate a shear component and moments increases. So, a reduction of M_x moment in the tibia is feasible with stronger and wider acceleration pedal to increases foot support. Lower F_x (results in upper M_y) is influenced by foot stopper, pedal rotation and foot impact with pedal arm or fire wall.

In case of using a wider acceleration pedal (see figure 18), the compression force due to pedal contact increased due to the fact that the heel did not slip off, tibia M_x were reduced as mentioned above. But tibia index injury is reduced due to the fact that tibia index is more sensitive for the moment than the forces on the tibia. (see Figure 19(a),(b))

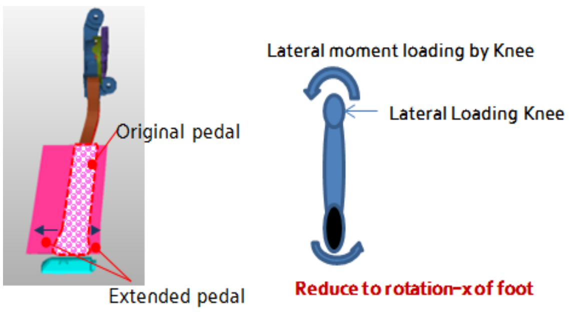

Figure 18. Loading condition of Tibia for reduce X rotation of foot.

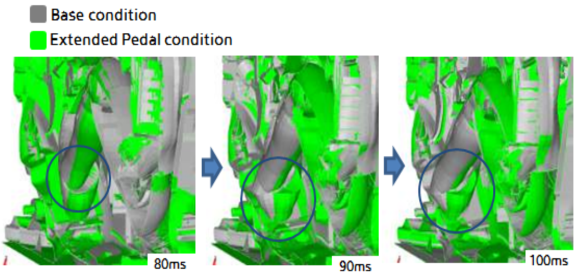

Figure 19(a). Compare to foot movement.

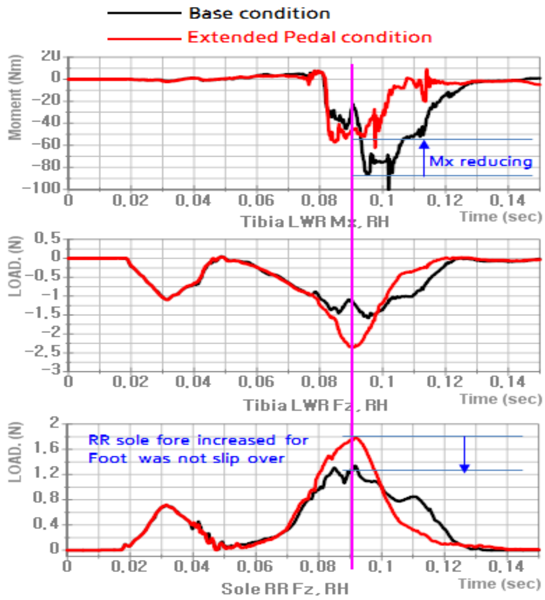

Figure 19(b). Compare graph for different pedal condition

The design objective is to push toes up and allow the foot to travel as much as possible forwards during the higher acceleration phases. Furthermore, padding on the fire wall will avoid a hard contact of foot on the fire wall. In addition, the IP design should allow the resulting tibia motion without interference of any stiff parts.

DISSCUSION AND LIMIATIONS

First, this research was conducted on sled tests and CAE analysis for a target of mid-size passenger car. Although the dummy models have been extensively validated for the standard sensor outputs, the new advanced sensor technology (Rib-Eye and etc.) was used in the project for improving the validation.

Some mechanisms showed to be very sensitive for minimal changes in the system, such as the right foot kinematics. As results the tibia moments and loads were not very reproducible in detail. However the overall mechanism remains similar such that the overall levels of the injury values for the tibias were still comparable.

Second, the results were based only on the analyzed two load cases (50th Driver Euro-NCAP and 5th Passenger USNCAP). The evaluation of other load cases might be needed to have overall balanced system that results in optimal protection that fulfills all the requirements.

CONCLUSION

Improvement Strategy for Neck Injury In phase 1, a combination of a high power anchor pretension force and soft seat characteristics can result in a relative high neck injury value. This can be improved with a less aggressive pretensioner and a stiffer seat cushion such that the dummy thorax is less accelerated.

In Phase 2, the design objective in this phase is to control the head motion with the airbag such that the relative motion to the thorax is minimal. At start of the ride down the thorax deceleration by the belt and the airbag deployment dynamics were causing extension moments ($-M_y$) on the neck. It means that this could be improved by having a more stabilized airbag at start of the initial contact. A softer airbag will decrease the positive contact force ($+F_x$) and increase the relative motion of the head with chest such that an increase in $-M_y$ occurred. The airbag design will influence the moment of the airbag on the head. Especially the use of tethers could influence the initial moment transfer and also stabilize the airbag position. In addition, the airbag generates compression forces on the neck due to the head rotation and airbag volume above the head. These forces could relate to the airbag pressure. Therefore active venting during the crash could help to fine tune the pressure balance of the airbag, and it also could be reduced the overall neck injury values when the peak values occurred after the initial phase.

Improvement Strategy for Chest Injury

The sensitivity study indicates that the reducing the lower belts load with increased dummy rotation should decrease the chest deflection. Reduced travel of the pelvis could be influence to reduce the lower shoulder belt force. Therefore, additional anchor pretensioner will be reducing the pelvis motion and increasing relative rotation of the thorax. In addition, airbag load path to the thorax should be minimal. Also airbag should load the chest preferably above the area of chest sensor as the chest deflection appears to be more sensitive in that area. Furthermore the use of shorter tethers will help to reduce the airbag pressure on the specific chest area. Steering wheel collapsing is crucial to obtain more space to absorb the crash energy and to reduce the airbag load on the chest.

Improvement Strategy for Tibia Injury

Tibia loads were related to the tibia mass. This was indicating that tibia loads were caused mainly by tibia deceleration and not by loads coming from upper body. Therefore, controlling the acceleration of the tibia in lower levels will improve the tibia injury. Tibia acceleration controlled mainly by pelvis restraint and foot motion. Upper tibia or knee acceleration can be controlled by pelvis restraint. Increased contact load of knee will increase pelvis deceleration and tibia loads. Tibia loads by lower IP contact should be needed for minimize M_y moment and keep a load balance in tibias. Also the tibia loads could be improved by controlling the foot motion. The implementation of that will require design changes to pedal construction and padding of the fire wall. Reduction of M_x moment in the tibia is feasible with wider acceleration pedal and stronger pedal. Lower F_x (results in upper M_y) is influenced by foot stopper, pedal rotation and foot impact with pedal arm or fire wall. For example, a guidance plate could improve tibia loads by smoother moving of the foot beyond the pedal arm. In addition, padding on the fire wall will avoid a hard contact of foot on fire wall.

Finally, in this paper, we had been described about injury mechanisms of neck, thorax and lower extremities in detail. This research will help you to understand the injury occurring mechanism of the dummy in frontal crash. Also it will be used Injury predictability guide lines for each restraint system and vehicle conditions.

REFERENCE

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