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Impact injury prediction by FE human body model

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Abstract

The biomechanical simulations as powerful instruments are used in many areas such as traffic, medicine, sport, army etc. The simulations are often performed with models, which are based on the Finite Element Method. The great ability of FE deformable models of human bodies is to predict the injuries during accidents. Due to its modular implementation of thorax and abdomen FE models, human articulated rigid body model ROBBY, which was previously developed at the University of West Bohemia in cooperation with ESI Group (Engineering Simulation for Industry), can be used for this purpose. ROBBY model representing average adult man is still being improved to obtain more precise model of human body with the possibility to predict injuries during accidents. Recently, new generated thoracic model was embedded into ROBBY model and this was subsequently satisfactorily validated. In this study the updated ROBBY model was used and injury of head and thorax were investigated during frontal crashes simulated by virtue of two types of sled tests with various types of restraint system (shoulder belt, lap belt and airbag). The results of the simulation were compared with the experimental ones.

Keywords: thoracic model, ROBBY model, sled test, restraint system, airbag, belts, injury criteria, thoracic injuries

1. Introduction

Head and thoracic injuries are the most frequent injuries during vehicle accidents. Most important injuries to the head are those to the skull and the brain [18]. Since at this time the head of ROBBY model is modeled by virtue of rigid body, the head injury can be predicted only from the value of acceleration of head Center Of Gravity (COG).

Thoracic injuries during the vehicular accidents are usually divided into two types: i) injuries of soft organs (laceration of lungs, rupture of aorta, etc.) and ii) injuries of bony parts (such as fractures of sternum, ribs, etc). Mechanisms of bony injuries can be studied on the postmortem subjects, however the cadaver use in case of investigation of soft organ's injuries is rather restricted because of considerable changes of the structure of these organs [2]. The other possibility to investigate injuries is offered by the correctly validated biomechanical model of human body or its part. This surrogate can help to visualize the process of injury mechanism during the accident. Therefore there arises the significant effort of many specialist to own the real validated model of human body, which would be able to predict injuries during various crash accidents. Injuries are predicted in the model via the injury criteria, which correlates a function of physical parameters (e.g. acceleration, force) with a probability of certain body regions to be injured. Injury criteria are generally proposed and validated on the basis of experimental studies [18]. The standard method for classifying the level of injury to a body region or organ is the Abbreviated Injury Scale (AIS) [1]. The AIS utilizes a numerical rating system

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to assess impact injury severity. It was created on the base of the demand for widely accepted injury scale that could be used by the medical engineering automotive accidents investigation teams to classify the level of injury [19]. Today it is used for the prediction of research and assessments during emergency medical situations. The scale (see Tab. 1) starts with 0, in case when no injury can occur and finishes by six, when the possibility of survival is negligible.

Table 1. AIS ranking codes [18]

AIS Level	Injury severity
0	No injury
1	Minor
2	Moderate
3	Serious (not life threating)
4	Severe (life threating but survivable)
5	Critical (survival uncertain)
6	Fatal

In this study human articulated rigid body model ROBBY [10, 11] with modular implementation of deformable thorax [4, 5] and abdomen [12] was used for the prediction of predominantly thoracic injuries during frontal crashes simulated via two types of sled tests. There was paid attention to thoracic segment since newly created thoracic model [4, 5] was newly embedded into ROBBY model and this was subsequently satisfactorily validated [6]. In whole study there was the used principle of Finite Element Method (FEM). The simulations were performed in the solver PAM-CRASH [17].

2. Description of sled tests

Two types of sled tests (see Fig. 1) according to [20] were performed to predict injuries during the frontal accident. The conditions of tests according to [20] are following. The seat geometry was close to standard mid-size car. The seat pan has a slope of 18 degrees and the footrest a slope of 43 degrees. The feet of the surrogate were strapped to the footrest. The seat was rigid. The subjects were restrained by separated shoulder and lap belts. The shoulder belt ran over the left clavicle and was equipped with a 4 kN force-limiting system.

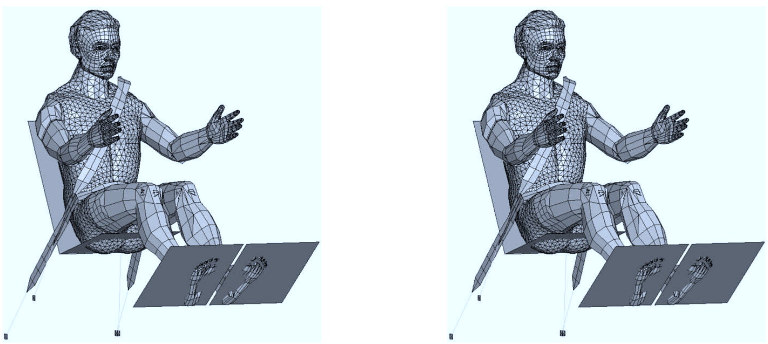

Fig. 1. Setup of sled tests with ROBBY model: 30 km/h test and 50 km/h test

First test was conducted at 50 km/h with a 22 g ([g] … acceleration of gravity) peak sled deceleration pulse and a restraint system composed of lap belt, a shoulder belt with a force limiter and a airbag. This test is referred to as 50 km/h test in the whole study. The second one was performed at a lower velocity 30 km/h with lower deceleration pulse 15 g using a lap belt and a shoulder belt with a force limiter. This test is referred to as 30 km/h test in the whole study. Figure 2 illustrates deceleration conditions of these two sled tests.

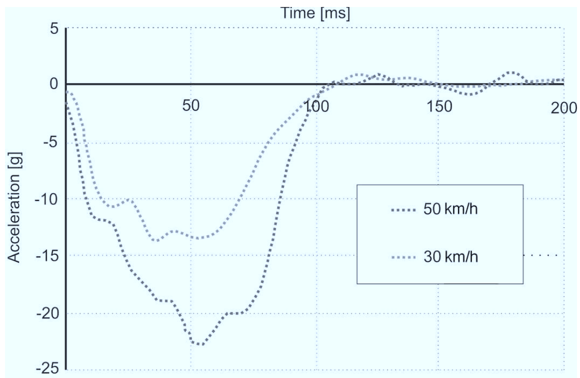

Fig. 2. Sled deceleration time history for 30 km/h and 50 km/h test [20]

3. Injury prediction by ROBBY model

Human body model ROBBY is able to predict the injuries using the standard injury criteria, as HIC (Head Injury Criterion) for head, acceleration for torso, etc. Moreover, since the thoracic and abdominal segment are modeled as deformable, the compression (C) and rate of compression (VC) can be computed. Maximal values of these variables can be used to predict the thoracic and abdominal injuries [2, 18]. Founding impact tolerances of the thorax are summarized in Tab. 2.

Table 2. Frontal and lateral impact tolerances of the thorax summarized in [18]

Frontal impact			Lateral impact		
Tolerance level		Injury level	Tolerance level		Injury level
Deflection [mm]	28	no rib fracture			
			Compression [%] to half thorax	35 33	AIS = 3 25% probability of AIS ≥ 4
Compression [%]	20 40	onset rib fracture flail chest	Compression [%] to whole thorax	38.4	25% probability of AIS ≥ 4
			VC_{max} [m/s] to half thorax	0.85	25 % probability of AIS ≥ 4
VC_{max} [m/s]	1.0 1.3	25% probability 50 % probability of AIS ≥ 4	VC_{max} [m/s] to whole thorax	1.0 1.47	50 % probability 25 % probability of AIS ≥ 4

Last, but not less importantly, the ROBBY model is able to predict rib fractures and the suggestion of the soft organ's injuries is allowed from the stress-strain analysis. The summary of investigated injuries of human body through the use of ROBBY model is described below. Some of the presented results are shown only to the time 150 ms. During this time there can be observed the loading and unloading of human body during such crash situations.

3.1. Head injury prediction

Although the great progress in passive safety, such as the introduction of advanced restraint systems, has been made in the last couple of years to reduce the number and severity of head injuries, there is only one injury criteria in wide use, the Head Injury Criterion. Besides the HIC, the 3 ms criterion and the Generalized Acceleration Model for Brain Injury Threshold (GAMBIT) are often presented [18].

3.1.1. Head Injury Criterion (HIC)

HIC (as a measure of average acceleration) was previously developed by [8] to reduce the number and severity of head injuries. This criterion is described by eq. (1).

$$HIC = \max\left(\frac{1}{t_2 - t_1}\int_{t_1}^{t_2} a(t)\,\mathrm{d}t\right)^{2.5}(t_2 - t_1), \tag{1}$$

where t_1 and t_2 are any two arbitrary times during the acceleration pulse. FMVSS (Federal Motor Vehicle Safety Standards) requires the HIC time interval to be 36 ms and the maximum HIC36 not to exceed a value of 1000 for the 50th percentile male (average adult male). However, this criterion is based on acceleration only [18]. The injuries that are related to impact force can not be predicted using this criterion.

The maximum head accelerations and HIC36 were investigated during the simulations of sled tests with ROBBY model. These values computed by the solver PAM-CRASH [17] were compared with the results of experiments performed with cadavers by Vezin et al. [20], who with his colleagues performed tests with six various cadavers, i.e. Test 1, Test 2 and Test 3 with conditions similar to 30 km/h test; and Test 4, Test 5 and Test 6 with conditions similar to 50 km/h test. HIC values of cadavers obtained during experiments and HIC values of ROBBY computed during simulations, are given in Tab. 3 and visualized in Fig. 3.

Table 3. Maximal resultant head acceleration and HIC36 values of ROBBY model during 30 km/h test and 50 km/h test

	30 km/h test		50 km/h test	
	$a_{max}[g]$	HIC [–]	$a_{max}[g]$	HIC [–]
Cadavers	22.9	66		
	44.0	210		
	34.9	160		
			66.4	620
			57.7	403
			46.7	350
ROBBY	26.5	105	64.0	659

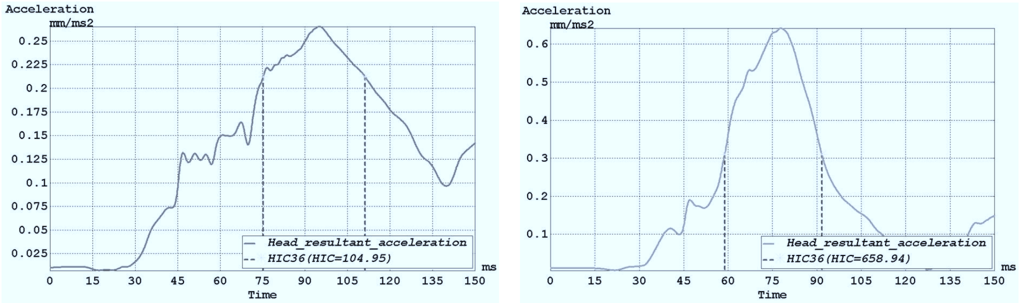

Fig. 3. Acceleration of ROBBY head COG and evaluation of HIC during the 30 km/h test and 50 km/h test

As can be seen from the Tab. 3, HIC value of ROBBY is in case of 30 km/h test in the range of experimental results. In case of 50 km/h test, HIC value of ROBBY model is little greater than HIC values of cadaver, however it is still comparable with the experimental results.

It can be observed in Fig. 3 that the maximal acceleration of head occurs in case of 30 km/h test around 100 ms and in case of 30 km/h test before 80 ms. It is caused by the fact that during the 30 km/h test the body is slowed-down up to shoulder belt, however in case of 50 km/h test the body is slowed early by the airbag.

3.1.2. 3 ms criterion (3ms)

This criterion is defined as the acceleration level obtained for an impact duration of 3 ms [18]. Value of head 3ms criterion should not exceed 80 g [9] according to FMVSS. The investigated head COG acceleration of ROBBY model and by solver PAM-CRASH computed values of 3ms values are visualized in Fig. 4 and summarized in Tab. 4.

From the investigated head criteria it can be concluded that during both tests, i.e. 30 km/h test and 50 km/h test, no serious injury of head arises.

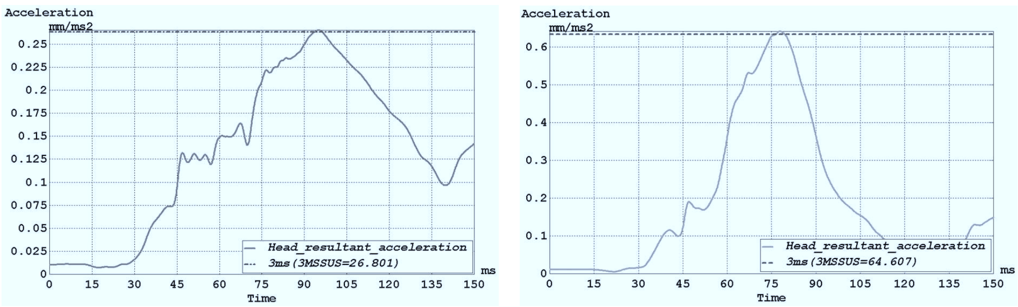

Fig. 4. Accelerations of ROBBY head COG and evaluation of head 3ms criterion during the 30 km/h test and 50 km/h test

Table 4. ROBBY values of head 3ms criterion during the 30 km/h test and 50 km/h test

	3ms [g]	
	30 km/h test	50 km/h test
ROBBY	27	65

3.2. Thorax injury

The injuries of thorax during crash accidents can be investigated through the use of acceleration, force, deflection, energy or combined criteria [3]. Some of them were used to predict thoracic injuries during 30 km/h and 50 km/h sled tests by virtue of ROBBY model.

3.2.1. Acceleration criteria

Acceleration criteria are used only as indicators of the injury [18]. Because the human body is deformable and the loading during a collision is complex, often resulting in localized deformations that may become injuries before an acceleration tolerance is exceeded, therefore the use of peak acceleration for evaluating designs for automobile safety is limited [15]. ROBBY resultant accelerations along the spine are visualized in Fig. 5.

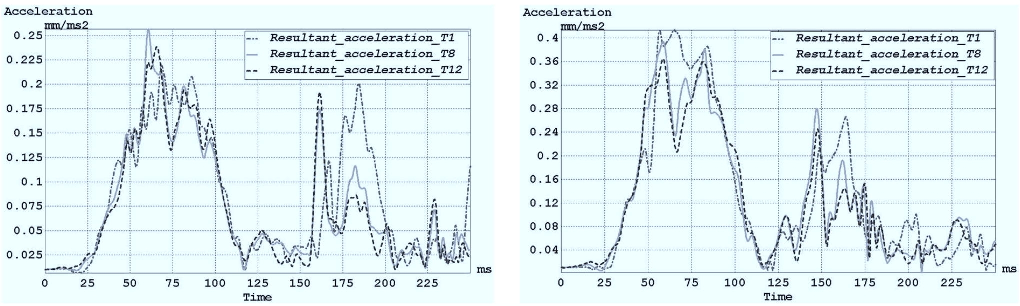

Fig. 5. The resultant accelerations of ROBBY thorax along the spine, i.e. accelerations of vertebra T1, T8 and T12, during the 30 km/h and 50 km/h test

Generally, the human tolerance for severe thorax injuries is considered as peak of spinal acceleration sustained for 3 ms or longer not exceeding 60 g according to FMVSS in a frontal impact [18]. From Fig. 5 it can be concluded, that thoracic accelerations do not exceed the threshold of 60 g. Usually, the thoracic injury is predicted by virtue of vertebrae T8, because the behavior of thorax is usually presented by T8 vertebrae behavior [15]. Therefore the acceleration of vertebra T8 was investigated and 3ms criterion was computed by solver PAM-CRASH.

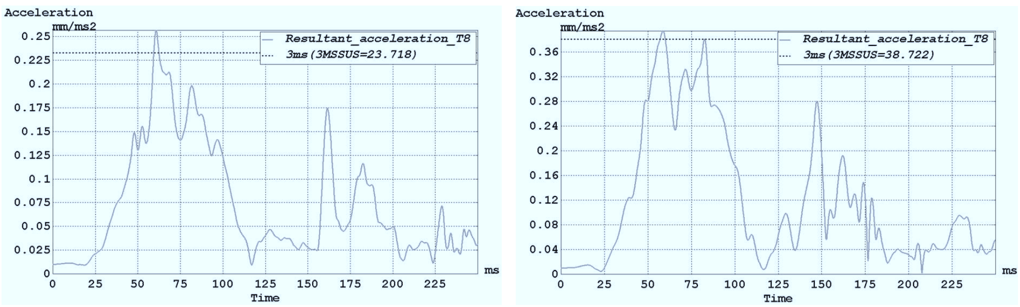

Fig. 6. Acceleration of ROBBY vertebra T8 and evaluation of 3ms criterion for ROBBY for 30 km/h and 50 km/h test

Table 5. ROBBY values of T8 3ms criterion during the 30 km/h test and 50 km/h test

	3ms [g]	
	30 km/h test	50 km/h test
ROBBY	24	39

3.2.2. Chest deflection

Following criteria are based on the chest deformation measurement. It was found that the location of maximum chest deformation depends on the application of the load by the restraint system [2]. The chest deflection, the posterior displacement of the anterior chest under frontal loading, is often used to describe the injury of thorax [15]. During the sled tests the maximum chest deflection is recorded in the anterior part of right lung and in the region of liver, see Fig. 7. So, high deflections caused by the asymmetrical loading of the belt are supposed to be localized along the shoulder belt path [13].

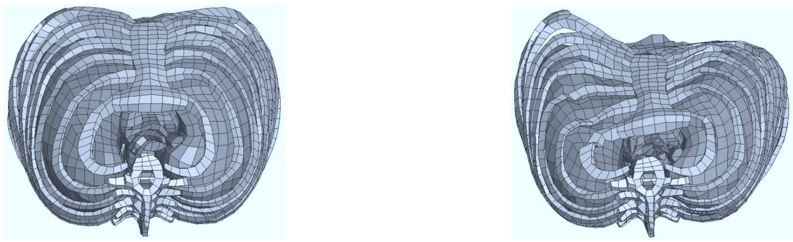

Fig. 7. Not deformed torso and deformed torso at time 75 ms, i.e. at the time of maximal shoulder belt loading

Compared to the load due to the standard belt system only, the distributed load of the airbag causes lesser chest deformation at the lower thorax level, however it does not allow a significant decrease of chest deformation at the upper level of the thorax [7]. To imagine this situation there were performed two types of 50 km/h sled tests — i) with airbag and ii) without airbag. The resultant deflection of upper torso and lower torso are visualized in Figs. 8, 9.

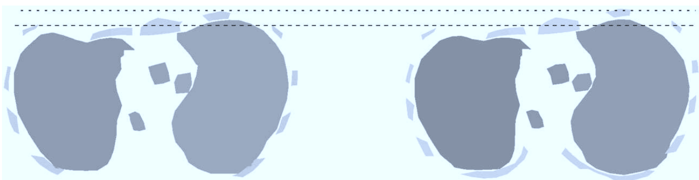

Fig. 8. Upper torso deflection, when a) no airbag, b) airbag was used during the 50 km/h test

Fig. 9. Bottom torso deflection, when a) no airbag, b) airbag was used during the 50 km/h test

From the Fig. 8 it can be concluded that the deflection of upper torso is in case, where the airbag was used comparable with the deflection of upper torso in case of without airbag, i.e. the using of airbag did not have the influence on the deflection of upper torso.

Fig. 9 displayes that the deflection of lower thorax is smaller (approximately by 5 %) in case of the using of airbag in comparison with the case, when no airbag was used.

Compression criteria

Compression criteria are based on the thoracic deflection measurement, see Fig. 10. It was found, that the maximal value of thoracic compression correlated well with AIS [16]. Therefore by definition of compression C as the chest deflection divided by the thickness of the thorax, i.e. $C = \frac{D(t)}{b}$, where $D(t)$ is the thorax deformation and b is the initial torso thickness, the following relation was established

$$\mathrm{AIS} = -3.78 + 19.56 * C_{max}, \tag{2}$$

where C_{max} is maximal value of compression. C_{max} can be considered to be an effective injury criterion for different restraint conditions, as long as the loading is concentrated through the sternum [2].

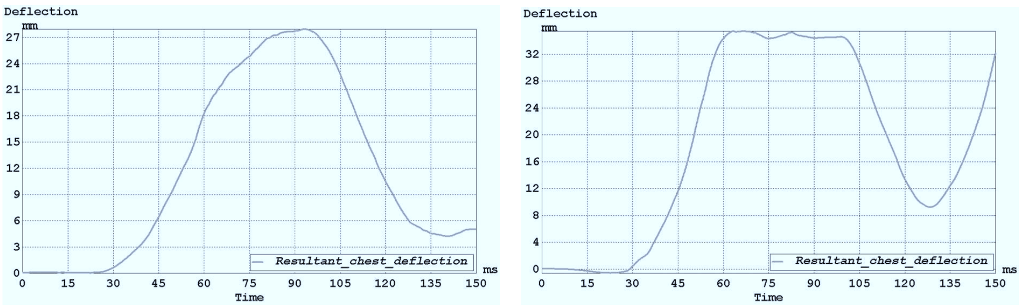

Fig. 10. ROBBY chest deflection during the 30 km/h and 50 km/h test

The maximal values of thoracic compression (see Tab. 6) were computed by virtue of maximal deflection and initial depth of thorax of ROBBY model during both sled tests. Consequently, the AIS values according to eq. (2) were determined from maximal compression values.

Table 6. Maximal value of ROBBY thoracic compression and corresponded AIS value for 30 km/h and 50 km/h test

	30 km/h test		50 km/h test	
	$C_{max}[\%]$	AIS	$C_{max}[\%]$	AIS
ROBBY	17	0	21	0

From the presented results it can be concluded that during both types of sled tests no injury of thorax arises, because the maximal values of compression 17 % and 21 % predict AIS = 0, equal to no injury.

Next criterion, which can be used to predict thoracic injuries is viscous criterion.

Viscous criteria
This criterion is also based on the thoracic deflection measurement. The viscous criterion VC (velocity of compression), also called the soft tissue criterion, is an injury criterion for the chest area taking into account that soft tissue injury is compression-dependent and rate-dependent [18]. This criterion is described by the relation

$$VC = V(t) * C(t) = \frac{\mathrm{d}[D(t)]}{\mathrm{d}t} * \frac{D(t)}{b}, \tag{3}$$

where $V(t)$ is the velocity of the deformation calculated by differentiation of the deformation $D(t)$ and $C(t)$ denotes the compression. b is initial torso thickness.

Hence, VC_{max} is simply the maximum value of VC over the impact time interval:

$$VC_{max} = [V(t) * C(t)]_{max} \tag{4}$$

The threshold level of 1 m/s, which corresponds to 25 % probability of $\mathrm{AIS} \geq 4$ is often chosen as a human thorax tolerance threshold [2].

ROBBY maximal computed thoracic values of VC for both sled tests are shown in Tab. 7.

Table 7. Maximal values of thoracic viscous response of ROBBY model for 30 km/h and 50 km/h test

	VC_{max} [m/s]	
	30 km/h test	50 km/h test
ROBBY	0.12	0.39

From Tab. 7 it is apparent that thorax tolerance level 1 m/s in both cases, i.e. 30 km/h test and 50 km/h test, is not exceeded.

3.3. Rib fracture

Another indicator of level of thoracic injury is the number of rib fractures. It was shown that the number of fractured ribs correlated to AIS value, i.e. the number of fractured ribs corresponds to seriousness of thoracic injury, see Tab. 8. Therefore the number of fractures in the whole thoracic area was defined as the injury indicator [2]. However, the amount of fractures is dependent on individual, i.e. it is influenced by the age, sex, material properties, etc. of specimen. But it can be said that the injury severity generally increases substantially as age increases, because the bones become more sensitive to fractures.

Due to the possibility of model to predict the rib fractures, the number of rib fractures was investigated, see Tab. 9. The threshold of rib fractures was chosen as stress equal to 85 MPa according to [21]. This value corresponds with average behavior of man. The number of rib fractures on the right side and left side was compared with the experimental results according to [20], see Tab. 9.

The number in Tab. 9 denotes the fracture of the x-th rib in the right or left side and the number in the exponent denotes the amount of fractures of x-th rib, i.e. the item $2.^{(2)}$ in the column LF denotes the two fractures of the second left rib and the item — denotes the no fracture of left or right ribs.

The comparison of AIS determined from the number of fractured ribs for cadaver tests and test with ROBBY model is summarized in Tab. 10. From the Tab. 9 it can be seen that during

Table 8. AIS rating for the skeletal and soft tissue injury (RF = rib fracture, HT = hemothorax, PT = pneumothorax) [1]

AIS	Skeletal Injury	Soft tissue injury
1	one RF	contusion of bronchus
2	2–3 RF; sternum fracture	partial thickness bronchus tear
3	4 or more RF on one side; 2–3 RF with HT or PT	lung contusion minor heart contusion
4	fail chest; 4 or more RF on each two side, 4 or more RF with HT or PT	bilateral lung laceration; minor aortic laceration; major heart contusion
5	bilateral fail chest	major aortic laceration; lung laceration with tension PT
6		aortic laceration with hemorrhage not confined to mediastinum

Table 9. Number of rib fractures of cadavers and ROBBY model during the 30 km/h and 50 km/h test (LF = fracture of left rib, RF = fracture of right rib)

	Cadaver tests				Simulation			
	30 km/h		50 km/h		30 km/h		50 km/h	
	LF	RF	LF	RF	LF	RF	LF	RF
Test 1	$2.^{(2)}, 3.^{(2)}, 4.^{(2)}, 5.^{(1)}, 6.^{(1)}, 7.^{(1)}, 8.^{(1)}, 9.^{(1)}$	–						
Test 2	$3.^{(1)}, 4.^{(1)}$	$4.^{(1)}, 5.^{(2)}, 6.^{(2)}$						
Test 3	–	–						
ROBBY					–	–		
Test 4			$3.^{(1)}, 5.^{(1)}$	–				
Test 5			$2.^{(2)}, 5.^{(1)}$	$6.^{(1)}$				
Test 6			–	–				
ROBBY							–	–

30 km/h test and 50 km/h test no rib fractures on ROBBY model was found. This findings correspond to AIS = 0 according to Tab. 8.

It is apparent that during the same impact conditions the injuries of various specimens can be different. From Tab. 10 it can be concluded that the result of 30 km/h test simulation is consistent with experimental result of Test 3 and the result of 50 km/h test simulation is comparable with experimental result of Test 6.

4. Conclusion

In this study the injuries of head and thoracic segment were predicted during the frontal crash simulated through the use of two types of sled tests with various restraint devices (shoulder belt, lap belt, airbag). The head injury was predicted by virtue of the exploration of head COG

Table 10. The value of AIS of thorax for experiments and simulations

	AIS for thorax			
	30 km/h test		50 km/h test	
	experiment	simulation	experiment	simulation
Test 1	3			
Test 2	2			
Test 3	0			
ROBBY		0		
Test 4			2	
Test 5			2	
Test 6			0	
ROBBY				0

acceleration. The values of these acceleration correspond to no serious head injury during 30 km/h test and 50 km/h test. However, it must be said that the investigation of injury by virtue of acceleration is only informative. Therefore there arise the need to develop the deformable FE model of human head in the future, which should allow to examine the head injury more precisely. Because head injuries are still one of the most frequent during crash accidents, there should be paid attention to this body segment.

Subsequently, the thoracic injuries during both sled tests were investigated. Since the new validated deformable FE model of thorax [4, 5] was implemented into ROBBY model, the thoracic injuries could be predicted not only by using of acceleration criteria but also by using the deflection criteria. Moreover due to the ability of model to predict the rib fractures, the seriousness of injury was determined from the number of rib fractures. The results of mentioned criteria advert to no injury of thoracic segment during both types of sled tests. This is in agreement with experiments (see Tabs. 10, 9), which show the different behavior of the human specimens during the same tests. Hence, during the same impact conditions there arise various level of injury. This is caused by the age, sex, fitness, etc. of individual human. However, the results of simulations are same in one case during both sled tests.

Acknowledgements

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References

[1] Abbreviated Injury Scale (AIS) — 1990 Revision. American Association for Automotive Medicine, Morton Grove, IL, (1990).

[2] F. Berthet, APROSYS Review of the thorax injury criteria, Deliverable D5.1.4.A, Part A, 2005.

[3] J. M. Cavanaugh, Biomechanics of Thoracic Trauma, Chapter 16, in Accidental Injury Biomechanics and Preventation, edited by Nahum and Melvin, Springer-Verlag, New York.

[4] L. Číhalová, Biomechanical models of tissue and their implementation into human body, PhD thesis, Plzeň, 2008.

[5] L. Čı́halová, Development of biomechanical deformable thoracic model, Engineering Mechanics Journal 13 (2006) 1 210–2717
[6] L. Čı́halová, L. Hynčı́k, Human body finite element model as instrument for improvement of passive safety, Transactions on Transport Sciences Journal 1 (2008).
[7] J. Crandall, C. Bass, W. Pilkey, et al., An evaluation of thoracic response and injury with belt, airbag, and force limited belt restraint. Proceedings of the NATO-ASI on Crashworthiness of transportation systems structural impact and occupant protection, 17 (1996).
[8] C. Gadd, Criteria for injury potential, Impact Acceleration Stress Symposium, Nat. Academy of Sciences, Washington, Nat. Research Council Pub. 977 (1961) 141–144.
[9] C. Got, A. Patel, A. Fayon, C. Tarriere, G. Walfisch, Results of experimental head impacts on cadavers: the various data obtained and their relation to some measured physical parameters, Proc. 22nd Stapp Car Crash Conference, SAE 780887 (1978) 57–99.
[10] L. Hynčı́k, Human articulated rigid body model (ROBBY1), ESI SA, 1997.
[11] L. Hynčı́k, Biomechanical Human Model, Biomechanics of Man, 2000.
[12] L. Hynčı́k, Deformable Human Abdomen Model for Crash Test Purposes, Applied Mechanics, 2002.
[13] D. Kallieris, A. Rizzeti, R. Mattern, R. Morgan, R. Eppinger, L. Keenan, On the synergism of the driver air bag and the 3-point belt in frontal crashes, Proceeding 39th Stapp Car Crash Conference, (1995) 389–401.
[14] D. Kallieris, et al., Prediction of thoracic injuries in frontal collision, Paper 98-S7-O-04, 16th Technical Conference on the Enhanced Safety of Vehicles, Ontario (Canada), 1998, 1 550–1 563.
[15] R. Kent, Dynamic Response of the Thorax: Restraint-Specific Injury Prediction and Constitutive Description, PhD thesis, University of Virginia, 2001.
[16] C. K. Kroell, D. C. Schneider,A. M. Nahum, Impact tolerance and response of the human thorax, SAE 710851 15th Stapp car crash conference, 1971, 84–134.
[17] PAM-CRASH/SAFE TM, Notes Manual, Version 2007.
[18] K. U. Schmitt, P. Niederer, F. Walz, Trauma Biomechanics, Introduction to Accidental Injury, Springer, 2004.
[19] J. D. States, The abbreviated and the comprehensive research injry scales, Proceedings of 13th Stapp car crash conference, 690810.
[20] P. Vezin, K. Bruyere-Garnier, F. Bermond, J. V. Verriest, Comparison of HYBRID III, Thor- a and PMHS Response in Frontal Sled Tests, Stapp Car Crash Journal 46 (2002) 1–26.
[21] J. Zhao, G. Narwani, Development of a human body finite element model for restraint system R & D applications, TAKATA – Automotive System Laboratory, Inc., paper number 05-0399.