Evaluation of Thoracic Injury Criteria for THUMS Finite Element Human Body Model Using Real-World Crash Data

Evaluation of Thoracic Injury Criteria for THUMS Finite Element Human Body Model Using Real-World Crash Data

Manuel Mendoza-Vázquez, Lotta Jakobsson, Johan Davidsson, Karin Brolin, Merete Östmann

Abstract This study aims to compare the thoracic injury risk predicted by a modified THUMS with the risks predicted by an injury risk curve constructed based on real-world data. Since the injury risk curves for the modified THUMS were developed from reconstruction of post-mortem human subjects tests, it is of interest to investigate their performance in real-world crashes. For this purpose, an AIS2+ injury risk curve was constructed based on selected and representative frontal car crashes from the Volvo Cars' Traffic Statistical Accident Database. Six simulations with three different crash severities and two acceleration pulses for each severity were performed with THUMS in a detailed and representative interior vehicle model. The injury criteria Dmax, DcTHOR, shear stress and first principal strain in the ribs were computed with the modified THUMS and the risks were obtained from its previously developed injury risk curves. These risks were then compared to the risk from the real-world data. All four THUMS criteria predict higher risk compared to the risk predicted by the real-world injury risk curve. The risk estimated with Dmax was closest to the risk estimated by the injury risk curve based on real-world data.

Keywords FE-HBM, field data, injury criteria, thorax, THUMS

I. INTRODUCTION

Injury risk curves (IRCs) relate the probability of injury to different levels of stimuli, for example a deceleration of the vehicle and its occupants, for a prescribed population [1]. IRCs could be obtained from several sources where the stimuli and the injury outcome are known. The first thoracic IRCs developed for crash safety, and scaled for use with crash test dummies, were obtained from post-mortem human subjects (PMHS) tests in the late 1960s [2], relating thoracic injury to chest deflection. Since then, several thoracic IRCs have been developed for different crash test dummies and mathematical models, most based on PMHS tests [3-4] and some based on reconstructions of real-world crashes [5-6].

In recent years, finite element human body models (FE-HBMs) have been developed. Since FE-HBMs have a more detailed representation of the human anatomy and material properties, it is possible to measure and evaluate the risk of injury at the tissue level. Hence efforts have been undertaken to develop thoracic injury criteria and IRCs for FE-HBMs using component tests; i.e. using cortical bone coupon test data [7]. There have also been efforts to develop IRCs based on PMHS sled test data [8]. Other studies have correlated test results obtained in reconstructions of various PMHS tests and injury information from the original PMHS tests in the development of IRCs [9-10]. However, PMHS lack muscular tonus and are usually older than the average age of the population of interest and are therefore more fragile. Consequently, when calculating the risk based on these IRCs, both the lack of muscle tonus and age effect could potentially be higher than the risk of thoracic injury for the targeted population. Therefore, it is of interest to know how accurate the IRCs developed using PMHS data actually predict the risks seen in real-world crashes.

In general there are two approaches to compare IRCs developed using PMHS data and real-world crashes. The first is to reconstruct one or a few crashes in great detail with a crash test dummy or FE-HBM and compare the injury risks computed with the human surrogates to the injury outcome of the reproduced accidents. The other approach is to use crash data from one type of crash and relate the injury outcome in this particular crash scenario to crash severity. In this way an IRC based on real-world data is generated. Examples of such IRCs are

M. Mendoza-Vázquez is a PhD student (phone +46-31-772-82-36, e-mail: manuel.mendoza-vazquez@chalmers.se), L. Jakobsson is an Adjunct Professor, J. Davidsson and K. Brolin are Associate Professors at the Department of Applied Mechanics, Chalmers University of Technology, Sweden. L. Jakobsson is also Senior Technical Leader and M. Östmann is a Senior Safety Engineer at Volvo Cars Safety Centre, Sweden.

provided by Kononen [11] and Stigson [12]. Representative crashes could then be reproduced using a crash test dummy or HBM to compare risks between real-world and crash test dummy or HBM-based IRCs, as in Laituri [13]. A disadvantage of the first approach is that the results might not be generalizable. The second approach has the potential to overcome that disadvantage.

Several criteria have been proposed to evaluate the risk of bone fractures with FE-HBMs. Some of them are based on strains [8, 14] and others on stresses [15]. In some other cases criteria based on deflections have been studied to predict thoracic injury. A previous study [9] compared injury criteria from the global to the material level to predict rib fractures using a modified Total Human Model for Safety (THUMS) [14, 16]. The results of that study indicated that the differential deflection criterion for THORAX (DCTHOR) and rib cortical bone shear stress (τ) were the injury criteria that best predicted the risk of fracture, based on simulations of 25 matched PMHS tests.

The objective of the current study is to compare the thoracic injury risk predicted by a modified THUMS with the risks predicted by an IRC based on real-world data, by reproducing a set of representative real-world crashes. The IRCs used with THUMS were constructed using PMHS test data in a previous study, while the real-world data is selected and IRCs constructed in this study.

II. METHODS

An AIS2+ thoracic IRC was developed from a selected and representative sample of real-world frontal impacts for a specific vehicle type. IRCs, previously developed for a modified THUMS [9] using PMHS data, were used to calculate the injury risk in simulations matching real-world frontal impact situations, using this FE-HBM, a detailed FE model representative of the vehicles in the real-world cases and six representative acceleration pulses. Thoracic injury criteria for the modified THUMS were then evaluated by comparing the injury risk predicted by the modified THUMS at three different crash severity levels to the AIS2+ thoracic IRC based on real-world data. The flow chart for this method is shown in Fig. 1.

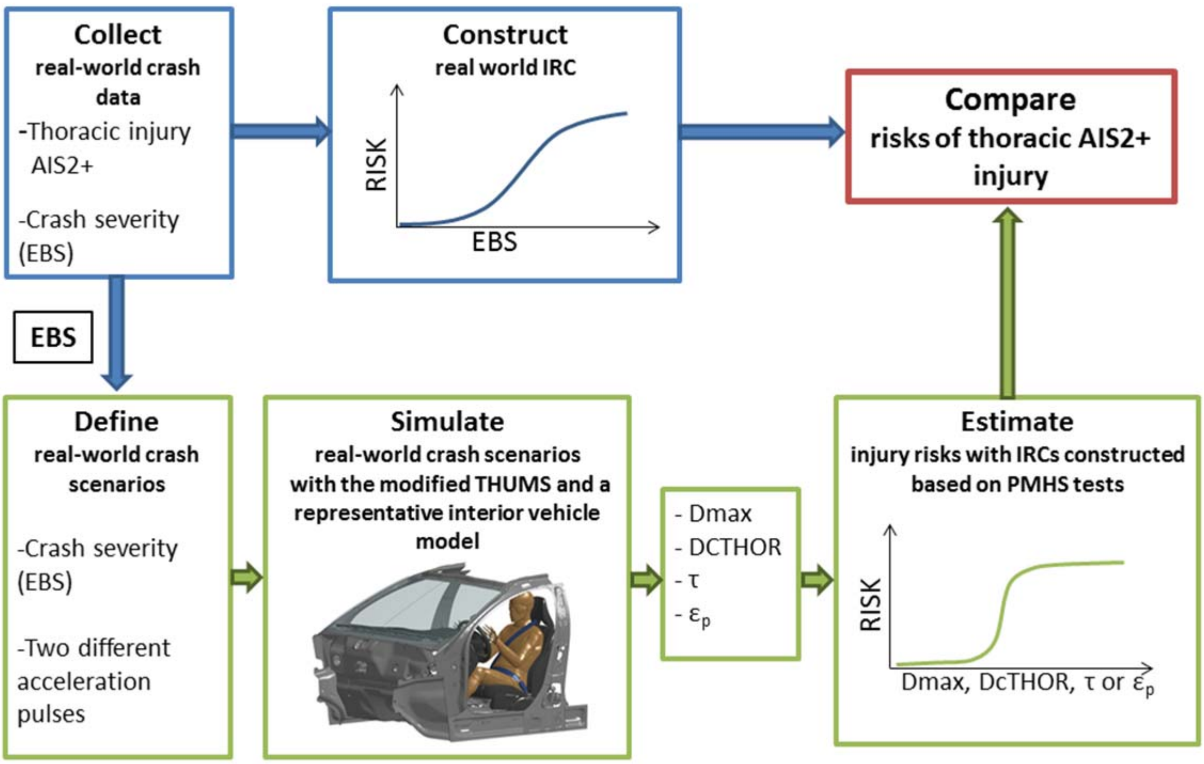


Fig. 1. Flow chart for the method followed in this study

Real-world data

Volvo Cars' Traffic Statistical Accident Database (VCTAD) was used for the thoracic IRC development. VCTAD contains data about Volvo passenger cars in Sweden in which the repair cost due to a crash exceeds a specified level. Inspectors from Volvia (If P&C Insurance), the company with which all new Volvo passenger cars are insured, identify these crashes. Photos and technical details of the vehicles are sent to Volvo Cars' Accident Research Team. A detailed questionnaire is sent to the owner of the vehicle to gather information about the crash, the car and the occupants. With the approval of the occupants, medical records are sent out (when applicable) and coded by a physician within Volvo Cars' Accident Research Team. Injuries are coded according to the Abbreviated Injury Scale (AIS). Based on the photos of the damaged car and information about the crash in the questionnaire, experts from the team code the deformation of the car. This coding is in accordance with the Society of Automotive Engineers (SAE) recommended practice Collision Deformation Classification (CDC) [17]. For frontal impacts, the impact severity is estimated using Equivalent Barrier Speed (EBS) [18]. EBS is calculated based on the CDC coding transformed to the energy matrix for the front structure of the car in question. This energy matrix is derived from laboratory tests for each car model. Based on these data, the EBS is then calculated with the help of the SAS software version 9.3 (SAS Institute Inc., Cary, NC). To date, the database contains a total of more than 42,000 Volvo cars with more than 70,000 occupants involved in crashes from 1976 to 2012. Detailed information about the database is found in [19].

For the purpose of this study, restrained drivers, with known injury outcome, involved in a single frontal impact during 2002-2012 (car year models 1999-2012), with a direction of impact 11-1 o'clock, and a horizontal overlap of 2/3, central overlap of 1/3 or full overlap, were selected. Multiple impact crashes and crashes with rollover events were excluded. The outboard 1/3 overlaps were excluded to avoid crashes with potential high lateral acceleration components. All cars in the sample were fitted with frontal airbags on the driver's side, 3-point seat belts with pre-tensioners and load limiters. Additionally, the airbag and belt function was verified to have adhered to specifications in all cases involving injured drivers. In total, the subset comprises 1,182 drivers, of which 1,007 have complete CDC data enabling an EBS calculation. The median age of the sample with 1,007 drivers was 47 years old. Of these 1,007 cases, 46 drivers sustained MAIS2+ injury, 381 drivers sustained AIS1 as maximum injury and 753 were uninjured. A total of 13 drivers, with a median age of 67 years old, sustained at least one thoracic AIS2+ injury. The distribution of the different Volvo car models included in the sample is presented in Fig. 2. The acceleration vs displacement response to US-NCAP full-width rigid barrier (FWRB) crash test at 56 km/h [20] for four car models included in the sample are displayed in Fig. 3. These four car models represent more than 80% of the cars in the sample and 70% of the cars with a driver sustaining thoracic injury AIS2+.

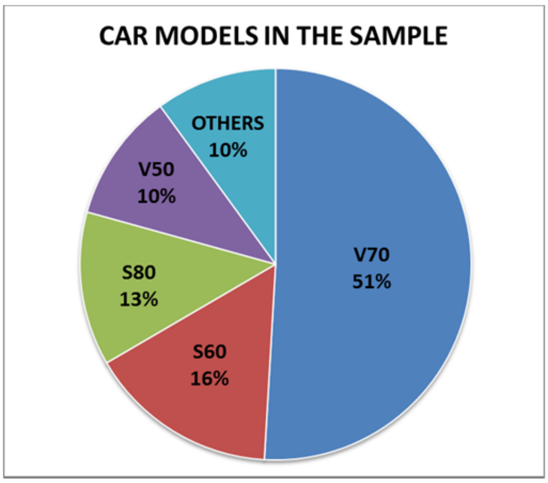


Fig. 2. Distribution of Volvo car models in the sample.

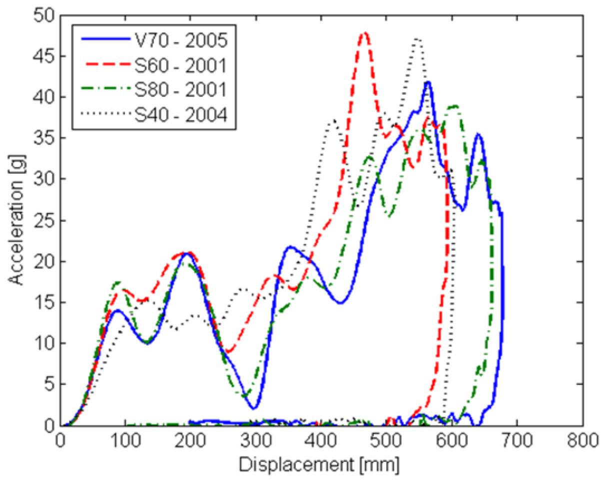


Fig. 3. Acceleration vs displacement for four different Volvo car models in a FWRB test at 56 km/h.

Parametric survival analysis was performed using the software R [21] performed on the data extracted from the VCTAD to establish the thoracic AIS2+ IRC for the real-world data, having EBS as the predictor variable and age as a covariate. All thoracic AIS2+injured drivers were considered as left censored and all non-injured as right censored. Three distributions, Weibull, log-normal and log-logistic, were included in the analysis. The distribution yielding the statistical model with the lowest Akaike Information Criterion (AIC) value and age adjusted to 61 years, the median age of the PMHSs included in the proceeding study on the development of IRCs for THUMS [9], was considered as the IRC based on real-world data. Details of the real-world data IRC construction are described in Appendix A.

FE models

All finite element simulations were performed on LS-DYNA (MPP version 971 R6.1.1, LSTC, Livermore, CA, USA). The pre- and post-processors used were LS-PREPOST (v2.4, LSTC, Livermore, CA, USA), Primer (v10.0, Oasys Ltd., UK) and in-house post processing scripts in MATLAB (R2007b, The Math Works Inc., Natick, MA, USA). The FE-HBM used in this study was a modified Total Human Model for Safety (THUMS version 3) [14], hereafter referred to as modified THUMS. This model was fitted with a finer mesh on the ribcage, without element elimination, and validated against impactor, table top and sled tests as described in [16]. The interior vehicle model was an FE model of one of the most frequent Volvo models in the real-world dataset. The model was developed for and used in-house during the vehicle development and verification process. Based on CAD data, the car body was meshed in-house and assigned the properties according to material test data. For systems such as airbags, belt, seats and the steering wheel, the FE models were developed by the supplier and verified in component or system testing. The complete interior model was validated using physical sled testing. The correspondence was found appropriate for vehicle development and verification. Fig. 4 shows the interior vehicle model and THUMS in the driver position.

Fig. 4. Interior vehicle model with THUMS in the driver position

Reconstruction of collisions and calculation of injury criteria value

THUMS was positioned in the driver seat of the interior vehicle model according to the Euro NCAP positioning protocol [22]. Six different acceleration pulses were applied to the interior vehicle model, two pulses for each of the three chosen EBS values: 30, 50 and 70 km/h. Details of the pulses are presented in Table 1. The six acceleration pulses were obtained from simulations of frontal car-to-car crashes with an overlap of 83%, for type A pulses, and 50% for type B pulses, and initial velocities of 30, 50 and 70 km/h for both vehicles as presented by Wågström [23]. The pulses were considered representative of the same EBS value since the simulations showed similar internal energy for the car structure in both overlap cases. The vehicle models used in the study by [23] featured the same vehicle model as did the interior model used in this study.

TABLE 1
ACCELERATION PULSE CHARACTERISTICS

Simulation	30A	30B	50A	50B	70A	70B
EBS [km/h]	30	30	50	50	70	70
Overlap [%]	83	50	83	50	83	50
Mean acceleration [g]	11.7	11.0	20.9	15.2	26.1	21.3
Duration [ms]	85	89	78	110	88	110

The following injury criteria were calculated with THUMS in each of the six simulations: Dmax [4], DcTHOR [24], maximum shear stress (τ) and first principal strain (ε_p). Dmax was calculated according to Eq. (1)

$$Dmax = \frac{max\big(D(t), UR(t), UL(t), LR(t), LL(t)\big)}{b} \cdot 100 \qquad (1)$$

Where *D, UR, UL, LR and LL* are the time histories of the deflection of each of the points illustrated in Fig. 5 and *b* is the initial chest depth, 230 mm. The combined deflection DcTHOR was computed as in Eq. (2)

$$DcTHOR = Dm + dDup + dDlw \qquad (2)$$

Where

$$Dm = (|UL(t)|_{max} + |UR(t)|_{max} + |LL(t)|_{max} + |LR(t)|_{max})/4$$

$$dDup = \begin{cases} |UL(t) - UR(t)|_{max} - 20 \\ 0\ if\ |UL(t) - UR(t)| \le 20 \\ 0 \text{ if } \min(|UL(t)|_{max}, |UR(t)|_{max}) \le 5 \end{cases}$$

$$dDlw = \begin{cases} |UL(t) - UR(t)|_{max} - 20 \\ 0\ if\ |UL(t) - UR(t)| \le 20 \\ 0 \text{ if } \min(|UL(t)|_{max}, |UR(t)|_{max}) \le 5 \end{cases}$$

and *UR, UL, LR, and LL* are as previously defined.

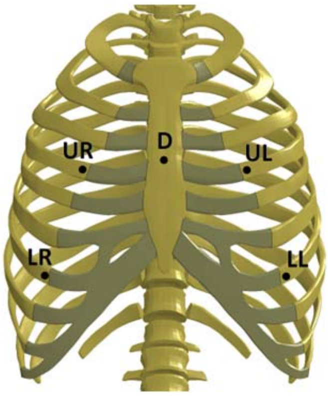


Fig. 5. Measurement sites on the THUMS ribcage. Data from these sites were used to calculate Dmax and DcTHOR

The maximum shear stress (τ) was calculated according to Eq. (3)

$$\tau = max\left(\frac{|\sigma_2 - \sigma_3|}{2}; \frac{|\sigma_3 - \sigma_1|}{2}; \frac{|\sigma_1 - \sigma_2|}{2}\right) \qquad (3)$$

Where σ_1, σ_2, and σ_3 are the principal stresses in the shells representing the rib cortical bone. The first principal strain (ε_p) was computed according to (4). τ and ε_p were calculated for ribs two to ten, and the maximum value from the rib with the second highest τ and ε_p, respectively, was used.

$$\varepsilon_p = max(|\varepsilon_1|; |\varepsilon_2|; |\varepsilon_3|) \qquad (4)$$

Where ε_1, ε_2, and ε_3 are the principal strains in the shells representing the rib cortical bone.

Comparison between real-world IRC and predicted injury risks with THUMS

In a preceding study [9] several IRCs were developed for two or more fractured ribs (NFR2+) for THUMS. Paired THUMS simulations and injury data from tests with 25 PMHS were used to construct IRCs. Fig. 6 to Fig. 9 show the developed IRCs for Dmax, DcTHOR, τ and ε_p for age 61 years old, the median age at time of death of the PMHS used to develop the injury risk curves. The age adjustment was implemented by including age as a covariate in the construction of the IRCs. The NFR2+ IRCs were assumed equivalent to AIS2+ IRCs. These IRCs were used in this study to calculate the injury risk predicted by THUMS in the six different crash simulations. The injury risk predicted by THUMS was then compared to the risk predicted by the IRC based on real-world data.

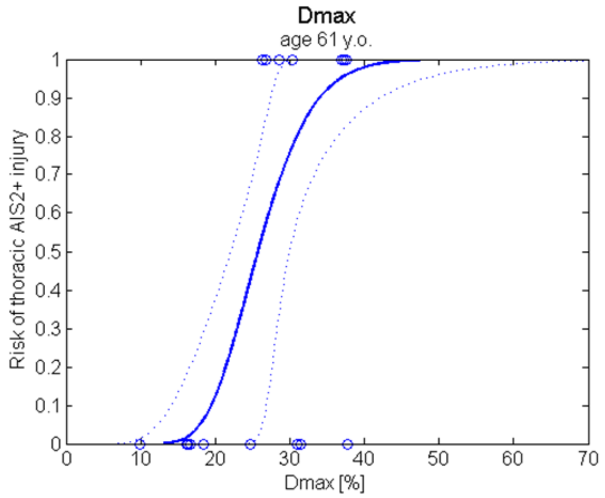


Fig. 6. AIS2+ IRC for Dmax and THUMS that were constructed from matched THUMS simulations and PMHS data [9]

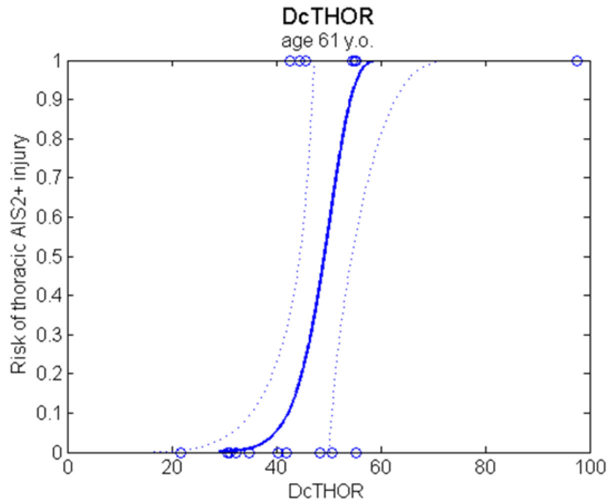


Fig. 7. AIS2+ IRC for DcTHOR and THUMS that were constructed from matched THUMs simulations and PMHS data [9]

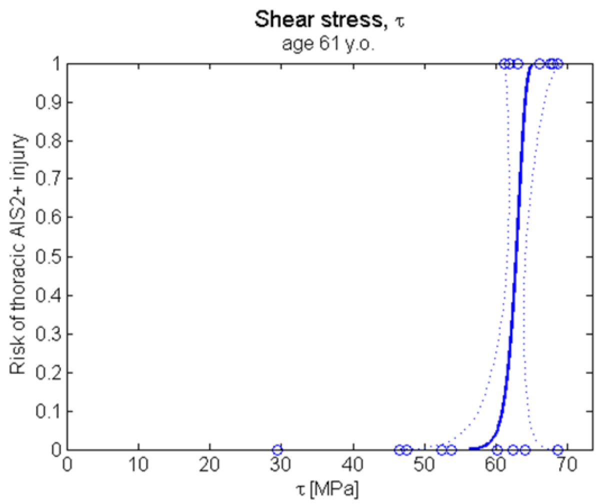


Fig. 8. AIS2+ IRC for τ and THUMS constructed from matched THUMs simulations and PMHS data [9]

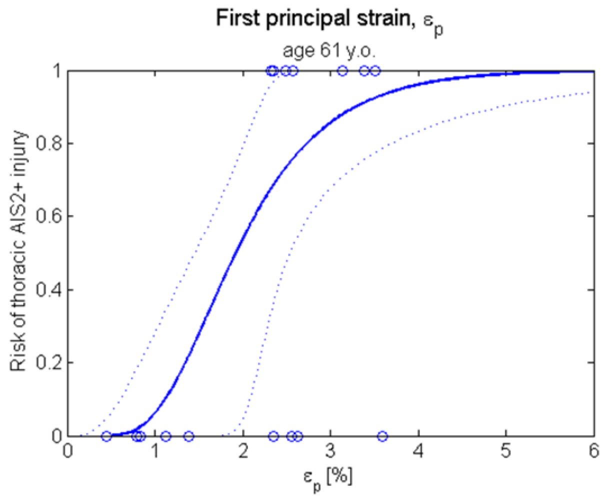


Fig. 9. AIS2+ IRC for ε_p and THUMS constructed from matched THUMs simulations and PMHS data [9]

III. RESULTS

In total 13 drivers sustained a thoracic AIS2+ injury in the subset of real-world data. The IRC based on real-world data and age 61 years old is shown in Fig. 10, along with its 95% confidence intervals. As described in Appendix A, this IRC is based on a Weibull distribution. This curve predicts a thoracic AIS2+ risk of 0.03, 0.25 and 0.71 for EBS values 30, 50 and 70 km/h, respectively.

The Dmax, DcTHOR τ and ε_p values for the simulations with THUMS at the three different EBS values and an acceleration pulse type A, from a car-to-car crash with an 83% overlap, are displayed in Table 2. The Dmax, DcTHOR, τ and ε_p values for the type B acceleration pulse, with a 50% overlap, are introduced in Table 3. The injury criteria values increased as the EBS increased. For EBS 30 and 50 km/h, the injury criteria values were greater for the type A pulse compared to type B pulse, with exception for τ at EBS 50 km/h. The simulations at EBS 70 km/h did not reach a normal termination due to the severity of the crash. The values presented here for EBS 70 km/h were obtained at 80 ms. The values for the injury criteria in these cases, in most of the cases, were already decreasing after reaching a maximum.

TABLE 2

INJURY CRITERIA VALUES FOR THUMS – TYPE A ACCELERATION PULSES

Criterion	EBS 30 km/h	EBS 50 km/h	EBS 70 km/h*
Dmax [%]	19.98	26.77	33.79
DcTHOR [mm]	79.27	108.43	141.25
τ [MPa]	60.87	70.17	71.81
ε_p [%]	2.38	3.51	6.03

*Values at 80 ms, simulations finished prematurely

TABLE 3

INJURY CRITERIA VALUES FOR THUMS – TYPE B ACCELERATION PULSES

Criterion	EBS 30 km/h	EBS 50 km/h	EBS 70 km/h*
Dmax [%]	18.84	23.87	34.86
DcTHOR [mm]	68.63	103.00	138.02
τ [MPa]	59.85	73.08	75.11
ε_p [%]	2.03	4.62	6.48

*Values at 80 ms, simulations finished prematurely

The THUMS injury risks predicted by Dmax are compared to the IRC based on real-world data in Fig. 10. In the same way, injury risks for THUMS using DcTHOR, τ and ε_p were calculated and compared to the injury risks predicted from the real-world based IRC, in Fig. 11 to Fig. 13.

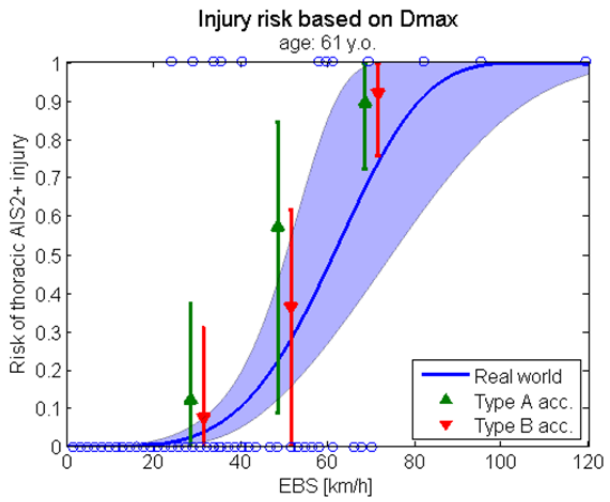


Fig. 10. Injury risk predicted by Dmax and THUMS at three EBS values and two acceleration pulses compared to the real-world data IRC. All data adjusted to 61 years old. Simulations for EBS=70km/h finished prematurely.

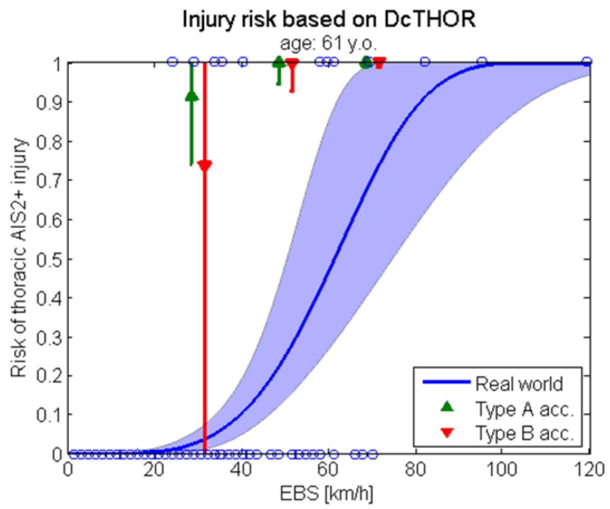
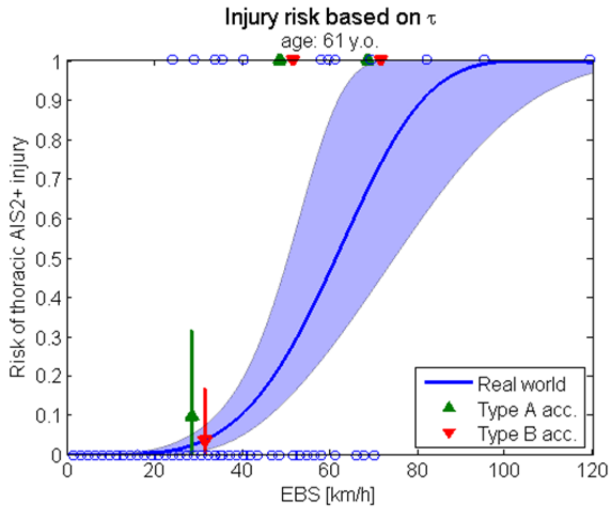


Fig. 11. Injury risk predicted by DcTHOR and THUMS at three EBS values and two acceleration pulses compared to the real-world data IRC. All data adjusted to 61 years old. Simulations for EBS=70km/h finished prematurely.



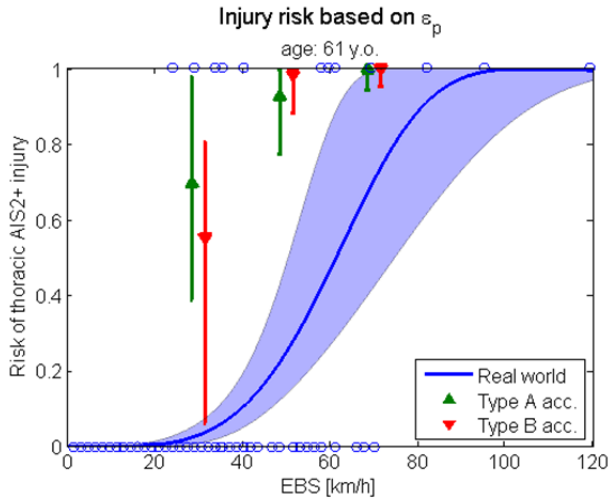


Fig. 12. Injury risk predicted by τ and THUMS at three EBS values and two acceleration pulses compared to the real-world data IRC. All data adjusted to 61 years old. Simulations for EBS=70km/h finished prematurely.

Fig. 13. Injury risk predicted by ε_p and THUMS at three EBS values and two acceleration pulses compared to the real-world data IRC. All data adjusted to 61 years old. Simulations for EBS=70km/h finished prematurely.

All four injury risks calculated with THUMS predicted a higher risk compared to the IRC based on real-world data. For EBS 30km/h, the simulations with acceleration pulses type A showed higher risks than the simulations with acceleration pulses type B. The same trend is observed at EBS 50 km/h for Dmax and DcTHOR. At EBS 70 km/h there was instead a slight increase in risk for the same change in acceleration pulses. The risks predicted by τ and ε_p were already above 0.9 for EBS=50 km/h for both acceleration pulse types.

The criteria τ and ε_p predict higher risks more rapidly than Dmax and DcTHOR; the stress and strain criteria indicated a risk higher than 0.9 at an EBS of only 50 km/h. The injury criterion that was closest to the IRC developed with real-world data was Dmax, the confidence intervals overlap for all tested EBS and acceleration pulses.

IV. DISCUSSION

This study compared the risks of thoracic AIS2+ injuries predicted with a modified THUMS to a set of frontal crash scenarios from a real-world database. The IRCs used with THUMS were constructed using PMHS test data. All four evaluated injury criteria (Dmax, DcTHOR, τ, ε_p) for THUMS predicted a higher risk for thoracic injuries compared to the risk calculated from real-world data. The modified THUMS predicted similar trends as the real-world data with increasing crash severity. This suggests that IRCs created based on PMHS data need to be calibrated to IRC from real-world data sets. However, there are several limitations in this study that contribute to the differences seen in the injury risks. The following discussion presents factors of importance.

An important difference between those datasets is that the IRCs for THUMS were developed from matching PMHS test data and THUMS simulations while the IRCs were based on real-world data that, for obvious reasons, were calculated from live drivers. Foret-Bruno [25] found that PMHSs sustain more rib fractures than live subjects at the same crash severity level. The lack of muscular tonus in PMHS is one of the reasons for this difference. Drivers may, prior to the crash, have braced their arms against the steering wheel and tensed the muscles in their rib cage; as a result the rib cage stiffness can increase by a factor of three [26] and thereby reduce rib cage compression. Bracing the arms against the steering wheel could change the torso kinematics [27] and thereby the rib cage load distribution [28] as compared to un-tensed drivers and PMHSs. Muscle activity could be implemented into THUMS to simulate a bracing action; but this was not within the scope of this study and hence needs to be taken into account in the assessment of the results.

An important difference in the characteristics of the datasets used to construct the THUMS IRCs and the real-world IRCs is the balance between injured and non-injured cases. The PMHS dataset was balanced, with 14 injured and 11 uninjured cases. The real-world data sample consisted of a total of 1,007 drivers, out of which 13 sustained at least a thoracic injury AIS2+. The large number of uninjured cases contributed to the widening of the confidence interval as EBS values increased. A more balanced dataset would reduce the confidence interval and possibly shift the IRC within the current 95% confidence interval.

The IRCs for THUMS developed in [9] were based on 25 PMHS with a median age at time of death of 61 years. The median age of all drivers in the real-world dataset was 47 years, while the median age for the injured drivers was 67. It is known that rib fracture risk increases with age [29] as does thoracic injury risk [30]. This effect was accounted for by adjusting the IRCs for THUMS and the IRCs based on real-world data to 61 years old.

In this study, the NFR2+ was considered equal to AIS2+, but they are not exactly equivalent; NFR2+ counts only the fractures on the ribs, while AIS2+ includes fractures and also injuries to the thoracic organs. In this study, AIS2+ injuries that were reported for young drivers in the real-world data were counted as rib fractures when compared to the THUMS injury risk. This assumption is based on the fact that older drivers are more likely to sustain rib fractures than younger ones [29] and therefore a crash severe enough to generate an AIS2+ injury in the thoracic organs of a young person would likely generate NFR2+ in an older driver. In the real-world dataset used in this study only one driver sustaining an AIS2+ injury in the lungs was assumed to have a NFR2+. The driver's age in this case was 19 years and the EBS was 59 km/h.

The THUMS's IRCs used in this study were developed from 25 PMHS tests. This was one of the reasons for using NFR instead of AIS as the injury measure. The AIS includes injuries to the thoracic organs, but post-mortem changes progress rapidly in soft tissues [31]. In contrast, post-mortem bones can keep their mechanical stable state for months [32]. The injury in some organs, like the lungs, heart and blood vessels, depends also on their physiological response. Reproducing such injuries would require the pressurisation of the PMHS cardiovascular system. Evaluation of rib fractures do not require such preparations, but might be influenced by other post-mortem changes, as previously discussed. The evaluation of rib fractures is then more reliable than the evaluation of thoracic organ injuries in PMHS tests. Furthermore, Wanek [33] found that the number of rib fractures is a good indicator of other thoracic injuries.

The way real-world data were modelled in this study may also explain why THUMS predicted higher risks. In this study, two crash pulses were used to represent a single EBS value. The two pulses were obtained from FE simulations of two frontal car-to-car crashes, one with an 83% overlap and one with a 50% overlap. The pulses were considered representative of the same EBS value since the simulations showed similar internal energy for the car structure in both overlap cases. The maximum difference in internal energy between overlap cases was 9% for an EBS of 70km/h and around 5% for an EBS of 30 and 50 km/h. The different pulses used had different mean accelerations and durations; the risks predicted with THUMS varied with pulse shape. However, these two pulses do not represent all pulses in the real-world data, where there is a large variety in collision objects. Based on deformation extent, the calculation of EBS to some extent takes into account different softer and lighter collision objects. Given that the initial speed of the car and the collision object are kept constant, and mass and stiffness of the collision object variable. The EBS of the car will decrease as the collision object becomes lighter and softer. Using pulses from frontal crashes of two identical cars might generally impose a higher acceleration pulse. Since the shape of the acceleration pulse has an influence on the risk predicted by THUMS, this is likely to influence the results and could make THUMS-predicted injury risks more conservative compared to the real-world frontal impact IRC.

In this study, a THUMS 50th percentile male was used to predict the risk. The male drivers in the real-world dataset represented 77% of the total and 85% of all the drivers sustaining AIS2+ thoracic injuries were males. The use of a male model is justified since it is representative of the majority of drivers in this dataset. Furthermore, Stigson [12] found that age was a more important factor than gender when predicting risk of MAIS2+ in frontal crashes. The average stature of the drivers in the dataset was 177 [cm] and the average weight was 81 [kg]. THUMS has a stature of 175 [cm] and weight of 77 [kg], values that compare favourably to the average of the drivers in the dataset.

The interior vehicle model used in the simulations did not consider intrusion. The small overlap crashes were excluded from the selection. This type of crash generates larger intrusions [23] and therefore the injuries in the drivers in the sample were more related to the deceleration than to the intrusion [34]. A way to account for the intrusions in the model, considering that enough computer resources are available, would be to run the simulations with a complete vehicle model and not only the interior model. The injurious cases were revised to assure that the load on the driver was not aggravated by large intrusions.

The material model in the modified THUMS' rib cortical bone is *MAT_PIECEWISE_LINEAR_PLASTICITY. This material model allows describing the plastic region as a set of linear relations between the stress and strain. The slope of these linear relations should decrease between consecutive segments. As a consequence, once plasticity is reached, the stress values increase at a lower rate and the strains increase at a higher rate. Since element elimination was deactivated, there is no upper limit for the value that strain can reach, neither for the stress. The fact that strain values increase at a higher rate than stress values is seen in Fig. 8 and Fig. 9, where τ shows a sharp boundary between no injury and injury. This effect is also visible in the values shown in Table 2 and Table 3 where ε_p increases faster than τ values between simulations at 50 and 70 MPa.

V. CONCLUSIONS

An AIS2+ IRC was developed based on representative and selected real-world frontal car impacts. Six simulations representing two acceleration pulses at three different crash severities were carried out with a modified THUMS in driver position in a detailed interior vehicle model representative of the car models involved in the real-world crashes. From these simulations, injury criteria were calculated and the injury risk was obtained using the existing AIS2+ IRCs developed using PMHS data in [9]. At this stage, the injury risks calculated using the modified THUMS were compared to the risks predicted by the real-world data based IRC. The results show that all four injury criteria for the modified THUMS (Dmax, DcTHOR, τ and ε_p) predicted higher risks than the risks obtained from the real-world data. Among the four studied criteria, Dmax was the criterion that predicted risks closest to the risks predicted by the real-world data based IRC. The available data in this study indicates that risks predicted by the modified THUMS are conservative and could be used in the design and evaluation of restraint systems. This study also suggests that IRC created based on PMHS data for HBM should be calibrated to IRC from real world data sets and that a methodology for such calibration, including requirements for risk curve generation based on real world data sets, should be a topic of future research.

VI. ACKNOWLEDGEMENT

Funding for this study was provided by VINNOVA – Swedish Governmental Agency for Innovation Systems through the FFI – Vehicle and Traffic Safety research programme and the industrial partners in the project. The work was carried out at SAFER – Vehicle and Traffic Safety Centre at Chalmers, Gothenburg, Sweden. Project partners were Chalmers, Autoliv Research, Volvo AB and Volvo Car Corporation. The simulations were performed on resources at Chalmers Centre for Computational Science and Engineering (C3SE) provided by the Swedish National Infrastructure for Computing (SNIC) and resources at Volvo AB. The authors would like to thank Dan Gustafsson for help with the accident data extraction and Linus Wågström for help with the acceleration pulses, both at Volvo Car Corporation.

VII. REFERENCES

[1] Prasad, P., Mertz, H., Dalmotas, D., Augenstein, J., and Digges, K. *Evaluation of the Field Relevance of Several Injury Risk Functions.* Stapp Car Crash J, 2010. **54:**49-72.

[2] Neathery, R. *Analysis of chest impact response data and scaled performance recommendations.* Stapp Car Crash J, 1974. **18:** p. 459-493.

[3] Petitjean, A., Trosseille, X., et al., *Injury risk curves for the WorldSID 50th male dummy.* Stapp Car Crash J, 2009. **53**: p. 443-76.

[4] Kleinberger, M., Sun, E., Eppinger, R., Kuppa, S., and Saul, R., *Development of improved injury criteria for the assessment of advanced automotive restraint systems*, 1989, NHTSA.

[5] Mertz, H., Horsch, J., Horn, G., and Lowne, R. *Hybrid III sternal deflection associated with thoracic injury severities of occupants restrained with force-limiting shoulder belts*. in *International Congress and Exhibition*. 1991. Detroit, USA: SAE International.

[6] Johansen, H., Trosseille, X., Lesire, P., and Beillas, P., *Estimating Q-Dummy injury criteria using the CASPER project results and scaling adult reference values*, in *International IRCOBI Conference on the Biomechanics of Impact, Sept 12-14*2012: Dublin, Ireland. p. 580-596.

[7] Kemper, A.R., McNally, C., et al., *Material properties of human rib cortical bone from dynamic tension coupon testing.* Stapp Car Crash J, 2005. **49**: p. 199-230.

[8] Forman, J.L., Kent, R.W., et al., *Predicting rib fracture risk with whole-body finite element models: development and preliminary evaluation of a probabilistic analytical framework.* Ann Adv Automot Med, 2012. **56**: p. 109-24.

[9] Mendoza-Vazquez, M., Davidsson, J., and Brolin, K., *Generation and evaluation of thoracic injury risk curves for THUMS in frontal impacts.* Manuscript to be submitted, 2014.

[10] Song, E., Trosseille, X., and Baudrit, P., *Evaluation of thoracic deflection as an injury criterion for side impact using a finite elements thorax model.* Stapp Car Crash J, 2009. **53**: p. 155-91.

[11] Kononen, D.W., Flannagan, C.A.C., and Wang, S.C., *Identification and validation of a logistic regression model for predicting serious injuries associated with motor vehicle crashes.* Accident Analysis and Prevention, 2011. **43**(1): p. 112-122.

[12] Stigson, H., Kullgren, A., and Rosen, E., *Injury risk functions in frontal impacts using data from crash pulse recorders.* Ann Adv Automot Med, 2012. **56**: p. 267-76.

[13] Laituri, T.R., Henry, S., Kachnowski, B., and Sullivan, K., *Initial Assessment of the Next-Generation USA Frontal NCAP: Fidelity of Various Risk Curves for Estimating Field Injury Rates of Belted Drivers*, 2009, SAE International.

[14] Iwamoto, M., Kisanuki, Y., et al., *Development of a finite element model of the total human model for safety (THUMS) and application to injury reconstruction*, in *International IRCOBI Conference on the*

*Biomechanics of Impact, Sept 18-20*2002: Munich, Germany.

[15] Keyak, J.H. and Rossi, S.A., *Prediction of femoral fracture load using finite element models: an examination of stress- and strain-based failure theories.* J Biomech, 2000. **33**(2): p. 209-14.

[16] Mendoza-Vazquez, M., Brolin, K., Davidsson, J., and Wismans, J., *Human rib response to different restraint systems in frontal impacts: a study using a human body model.* International Journal of Crashworthiness, 2013. **18**(5): p. 516-529.

[17] SAE International Surface Vehicle Recommended Practice, *Collision deformation classification (CDC) SAE standard J224*, 1980.

[18] Campbell, K.L. *Energy basis for collision severity*. in *Third International Conference on Occupant Protection*. 1974.

[19] Isaksson-Hellman, I. and Norin, H., *How thirty years of focused safety development has influenced injury outcome in volvo cars.* Annu Proc Assoc Adv Automot Med, 2005. **49**: p. 63-77.

[20] NHTSA. *NHTSA Vehicle Crash Test Database*. 2014 [cited 2014; Available from: http://www-nrd.nhtsa.dot.gov/database/veh/veh.htm.

[21] R Core Team, *R: A language and environment for statistical computing*, 2012, R Foundation for Statistical Computing: Vienna, Austria.

[22] Euro NCAP, *European new car assessment programme (Euro NCAP), frontal impact testing protocol, Version 5.0*, 2009.

[23] Wågström, L., Kling, A., Norin, H., and Fagerlind, H., *A methodology for improving structural robustness in frontal car-to-car crash scenarios.* International Journal of Crashworthiness, 2013. **18**(4): p. 385-396.

[24] Davidsson, J., Carroll, J., et al., *Development of injury risk functions for use with the THORAX Demonstrator; an updated THOR*, in *International IRCOBI Conference on the Biomechanics of Impact, Sept 10-12*2014: Berlin, Germany.

[25] Foret-Bruno, J., Hartemann, F., et al., *Correlation between thoracic lesions and force values measured at the shoulder of 92 belted occupants involved in real accidents.* Stapp Car Crash J, 1978. **22**: p. 271-291.

[26] Lobdell, T.E., Kroell, C.K., Schneider, D.C., Hering, W.E., and Nahum, A.M., *Impact response of the human thorax*, in *Human impact response: Measurement and simulation*. 1973, Plenum Press: New York. p. 201-245.

[27] Ejima, S., Zama, Y., et al., *Prediction of physical motion of the human body based on muscle activity during pre-impac braking*, in *International IRCOBI Conference on the Biomechanics of Impact, Sept 17-19*2008: Bern, Switzerland.

[28] Armstrong, R., Waters, H.P., and Stapp, J.P., *Human Muscular Restraint during Sled Deceleration.* SAE Technical paper, 1968. **SAE Paper No. 680793**.

[29] Kent, R., Lee, S.H., et al., *Structural and material changes in the aging thorax and their role in crash protection for older occupants.* Stapp Car Crash J, 2005. **49**: p. 231-49.

[30] Kent, R., Henary, B., and Matsuoka, F. *On the fatal crash experience of older drivers*. in *Annual Proceedings/Association for the Advancement of Automotive Medicine*. 2005. Association for the Advancement of Automotive Medicine.

[31] Crandall, J.R., Bose, D., et al., *Human Surrogates for Injury Biomechanics Research.* Clinical Anatomy, 2011. **24**(3): p. 362-371.

[32] Yamada, H., *Strength of biological materials*. 1970, Baltimore: Williams & Wilkins.

[33] Wanek, S. and Mayberry, J.C., *Blunt thoracic trauma: flail chest, pulmonary contusion, and blast injury.* Crit Care Clin, 2004. **20**(1): p. 71-81.

[34] Wågström, L., *Structural Safety Design for Real-World Situations-Using Computer Aided Engineering for Robust Passenger Car Crashworthiness*, in *Applied Mechanics*2013, Chalmers University of Technology: Gothenburg, Sweden.

VIII. APPENDIX A - REAL-WORLD DATA INJURY RISK CONSTRUCTION

The EBS, data on drivers sustaining or not a thoracic injury AIS2+, and age of the drivers in the VCTAD sample were analysed through parametric survival analysis in R (R Core Team 2012). All injured drivers were considered as left censored and all uninjured as right censored in this analysis. The following distributions were tested in the parametric survival analysis, Weibull, log-logistic, and log-normal, as they ensure zero injury risk at an EBS equal zero. Two IRCs were constructed from each distribution, one with age adjustment and one without age adjustment. All distributions with no age adjustment are presented in Figure A1 together with the non-parametric maximum likelihood estimate (NPMLE).

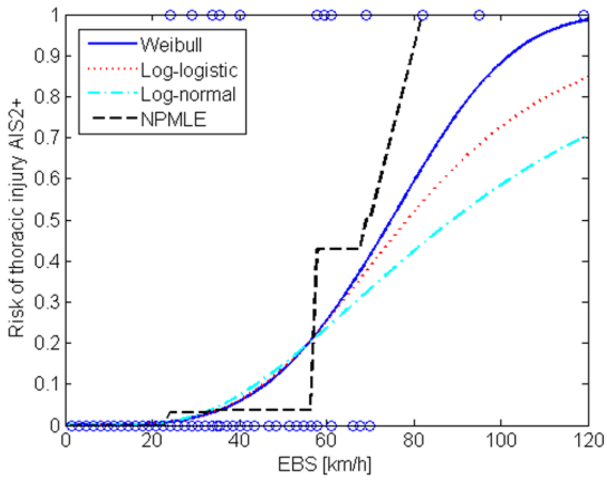


Fig. A1. IRCs assuming Weibull, log-logistic and log-normal compared to the NPMLE

Age adjustment was achieved by including age as a covariate in the parametric survival analysis. The AIC values for each IRC are presented in Table A1.

TABLE A1
AIC VALUES

Distribution	Non age adjusted	Age adjusted
Weibull	82.6	76.8
Log-normal	83.7	78.1
Log-logistic	85.0	79.2

The Weibull distribution obtained the lowest AIC values among all distributions. The AIC value decreased when age was considered as covariate. This indicates that the IRC with Weibull distribution and age adjustment is the IRC that best fits the VCTAD sample data among all six IRC in this study.

The three parameters for the IRC with Weibull distribution and age adjustement (scale, shape and age coefficient) were calculated again, but removing one observation at a time. In this way, the most influential observations were identified for each parameter and compared to the original IRC in Fig. A2.

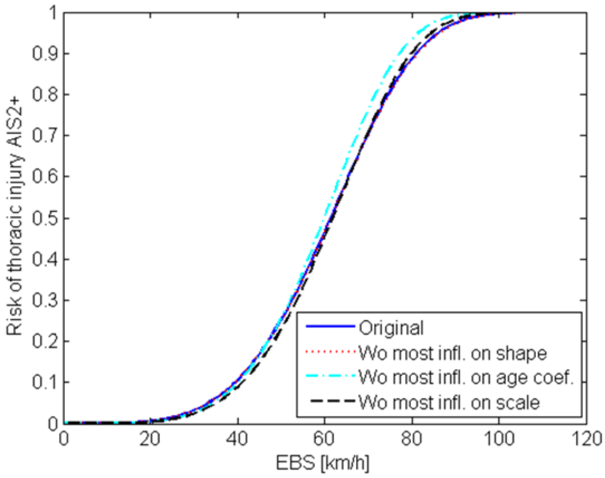


Fig. A2. IRCs without the most influential observations for each parameter

IX. APPENDIX B – TABULATED INJURY RISKS PREDICTED WITH THUMS

TABLE B1
INJURY RISKS FROM THUMS WITH 95% CONFIDENCE INTERVAL AND AGE ADJUSTED TO 61 YEARS
TYPE A ACCELERATION PULSES

EBS [km/h]	AIS2+ risk from real-world data IRC	AIS2+ risk from Dmax IRC	AIS2+ risk from DcTHOR IRC	AIS2+ risk from τ IRC	AIS2+ risk from ε_p IRC
30	0.032 (0.013 - 0.085)	0.123 (0.001 - 0.374)	0.912 (0.738 - 0.999)	0.099 (0.001 - 0.314)	0.694 (0.386 - 0.976)
50	0.251 (0.152 - 0.502)	0.571 (0.090 - 0.842)	0.998 (0.943 - 0.999)	1.000 (0.995 - 1.000)	0.924 (0.773 - 0.999)
70	0.705 (0.328 - 0.867)	0.893 (0.720 - 0.999)	1.000 (0.988 - 1.000)	1.000 (1.000 - 1.000)	0.997 (0.940 - 0.999)

TABLE B2
INJURY RISKS FROM THUMS WITH 95% CONFIDENCE INTERVAL AND AGE ADJUSTED TO 61 YEARS
TYPE B ACCELERATION PULSES

EBS [km/h]	AIS2+ risk from real-world data IRC	AIS2+ risk from Dmax IRC	AIS2+ risk from DcTHOR IRC	AIS2+ risk from τ IRC	AIS2+ risk from ε_p IRC
30	0.032 (0.013 - 0.085)	0.076 (0.001 - 0.311)	0.734 (0.001 - 0.999)	0.037 (0.001 - 0.167)	0.552 (0.062 - 0.805)
50	0.251 (0.152 - 0.502)	0.364 (0.001 - 0.614)	0.996 (0.926 - 0.999)	1.000 (0.999 - 1.000)	0.981 (0.881 - 0.999)
70	0.705 (0.328 - 0.867)	0.917 (0.756 - 0.999)	1.000 (0.986 - 0.999)	1.000 (0.999 - 1.000)	0.998 (0.951 - 0.999)