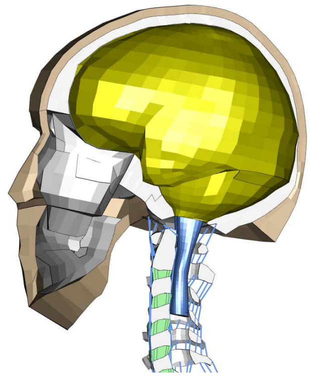
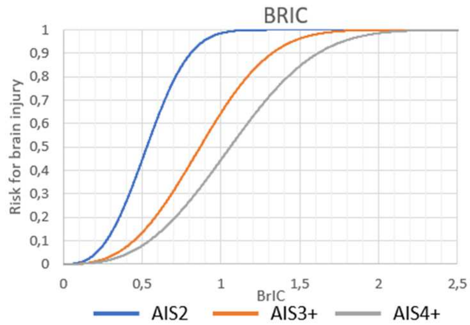
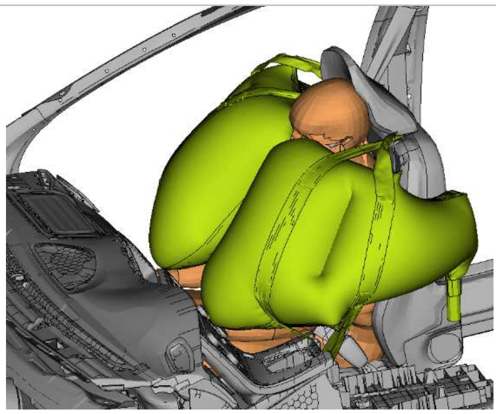
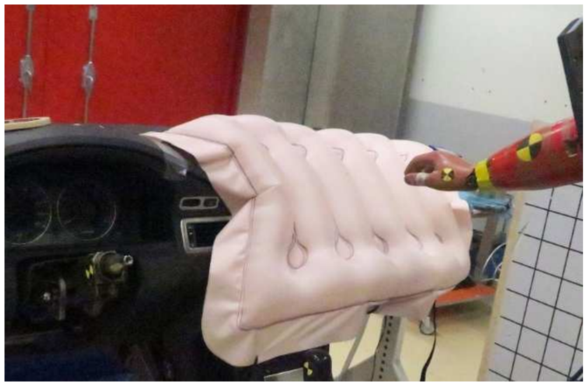
Identification and Prediction of Injuries with Long-term Consequences

Public Report









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Date: 2021-03-01

Project within 'Trafiksäkerhet och automatiserade fordon'

FFI Fordonsstrategisk Forskning och Innovation

Table of Contents

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1 Summary

Injuries to the head and upper extremity body regions more commonly result in long-term consequences than injuries to other regions of the body. This project focuses on enhancing understanding of injuries in these body regions, including development of injury assessment tools and methods. In addition, a method for predicting risk of long-term consequences, using the human body model, SAFER HBM, was developed and applied in various head and upper extremity impact studies.

To evaluate the applicability of various global head injury criteria for prediction of head injury risk in bicycle helmet evaluations and automotive occupant protection systems, a number of studies were carried out, including:

- Bicycle helmet impact tests
- Comparison of state-of-the art finite element (FE) head models for short and long duration pulses
- Free motion headform (FMH) impact tests
- Airbag evaluation in frontal oblique impact

In these studies, global head injury criteria based on linear acceleration and rotational velocity were evaluated. For the FE head models, head injury risk was evaluated by means of predicted max principal strain in the brain. The applicability of using the global head injury criteria was evaluated by comparing the injury risk predicted by the global head injury criteria with the injury risk predicted by the tissue-based injury criteria predicted by the FE head models. Generally, it was found that all injury criteria with a dominant rotational velocity component, correlate well with the computed max principal strain with the FE head model. A single global head injury criterion that seems applicable to be used for both bicycle helmet evaluation and automotive impact evaluation was not found.

Countermeasures to reduce head injury risks for vulnerable road users and vehicle occupants were evaluated. A system designed to reduce head rotational velocity in bicycle helmets was in testing found to reduce head rotational velocity and hence head injury risk. Airbag concepts designed to reduce head rotational velocity in oblique passenger vehicle impacts were also found to reduce head injury risk. A bicyclist impact airbag was also evaluated and found reduce head injury risk.

In addition, a robustness evaluation of the influence of inflatable curtain design to mitigate head injury risk in side impact was carried out. It was found that the layout of the vertical tubes of the curtain significantly influences the robustness of the inflatable curtain in mitigating head injuries.

The head model developed at KTH (Royal Institute of Technology in Stockholm, Sweden) was updated and injury risk functions for prediction of mild traumatic brain injuries and skull fracture were developed. A model of the cervical spinal cord was developed, validated, and included in the SAFER HBM model. The skull base and the C0 vertebrae were modified. The mathematical representation of the connection was improved (mesh).

With respect to upper extremities, a test method comprising a novel instrumented crash test dummy forearm was developed, physical as well as in CAE. In addition, models of human forearm and elbow were developed, and integrated into the SAFER HBM model. The human forearm model was validated by means of impact tests with PMHS lower arms. Injury risk functions for prediction of forearm injury risks were developed for the novel instrumented forearm as well as the SAFER HBM arm model. The injury risk functions were developed by replicating published impact tests with post mortem human subjects.

An airbag concepts, targeting reduction of upper extremity injuries caused by hand impacts to the instrument panel, was developed and evaluated. The concept was built and evaluated using the novel instrumented forearm, in addition to using the updated SAFER HBM together with the injury risk functions and the method to predict long-term consequences risk as a means to quantify the injury reducing benefits of the instrument panel hand airbag. A substantial injury reduction potential was obtained, overall as well as with respect to long-term consequences.

The project has developed new method, including functions for two body regions, which are disposed to relative high risks of long-term consequences. This enables devevelopment of advanced protection systems providing the industry a competitive edge, as well as contribute to protect occupant as well as vulnerable road users.

2 Sammanfattning på svenska

Skador på huvud och övre extremiteter har visat sig oftare leda till långtidskonsekvenser än skador på andra kroppsdelar. Projektet har fördjupat kunskapen kring uppkomsten av skador på dessa kroppsdelar, samt utvecklat verktyg och metoder för att möjliggöra utvärdering av skaderisk vid islag. Humanmodellen SAFER HBM har vidareutvecklats och förfinats som ett verktyg att använda för prediktering av risken för skador med långtidskonsekvenser för huvud och armar. Dessutom har ett antal studier med den vidareutvecklade SAFER HBM modellen genomförts.

Risken för huvudskada vid islag har utvärderats med globala skadekriterier baserat på både linjär- och rotationsrörelse. Dessa globala skadekriterier har jämförts med prediktering från finita element huvudmodeller i form av maxtöjning av hjärnvävnaden. Generellt visade resultaten från studierna att alla globala kriterier med en dominerande rotationshastighetskomponent korrelerar väl med den beräknade maxtöjningen i hjärnan. Inget av de utvärderade globala huvudskadekriterna visade sig vara lämpligt att använda för utvärdering av både cykelhjälm (islag med kort varaktighet) och skyddssystem i bil (islag med lång varaktighet).

Skyddssystem utvecklade för att reducera risken för huvudskador för oskyddade trafikanter och för åkande i bil har utvärderats. En luftkudde konstruerad för att minska huvudets rotation i vinklade kollisioner visade i en utvärdering med huvudmodellen att luftkudden också reducerade rotationen och skaderisken. En cyklistluftkudde visade också en reduktion av både linjär acceleration och rotation av huvudet i en kollision.

En provmetod för att utvärdera cykelhjälmar för islag i bil och mark har utvecklats. Vid utvärderingen av cykelhjälmarna monterades dessa på en huvudattrapp. Huvudatrapp med och utan hjälm sköts mot A-stolpe, huv och asfalt. För alla utvärderade konfigurationer reducerade cykelhjälmarna huvudets skaderisk.

Dessutom har en utvärdering av hur en krockgardins utformning påverkar risken för huvudskador i sidokollision genomförst i form av en robusthetsstudie. Utvärderingen visade att utformningen av gardinens vertikala kanaler påverkar robustheten och därmed risken för huvudskador vid krock. Denna kunskap kommer att användas i framtida konstruktioner av krockgardiner.

Huvudmodellen utvecklad vid Kungliga Tekniska Högskolan har uppdaterats. Modellen har kompletterats med en modell av ryggmärgen som i projektet har utvecklats, validerats och integrerats ihop med SAFER HBM modellen. Skallbasen och de översta ryggkotorna har också modifierats för att möjliggöra predikerting av skallbasfrakturer.

För utvärdering av risken för skador med långtidskonsekvenser för övre extremiteter har en mekanisk intrumenterad dockarm, en modell av den instrumenterade dockarmen och finita elementmodeller av människoarmar utvecklats och validerats. Männsikoarmsmodellerna har integrerats ihop med SAFER HBM modellen. Skaderiskfunktioner för prediktering av risken för underarmsskador har utvecklats genom rekonstruktion av publicerade islagstester med biologiskt preparat.

Ett luftkuddekoncept för att reducera risken för skador på övre extremiteter vid islag mot instrumentbrädan har utvecklats och utvärderats med SAFER HBM modellen. Konceptet har byggts och utvärderats med den mekaniska provmetoden. De i projektet utvecklade skaderiskfunktionerna har använts för att prediktera risken för långtidskonsekvenser. Konceptet predikterade minskning av skaderisk, inklusive risk för långtidskonsekvenser.

Projektet har utvecklat nya metoder inklusive funktioner för prediktering av skaderisk för huvud och armar vid islag. Detta möjliggör utveckling av nya avancerade skyddssystem som kan ge svenska fordonsindustrin och underleverantörer en konkurrensfördel och bidra till att minska skador i trafiken, både i och utanför bilen.

3 Background

Automobile safety has experienced substantial changes since the late 1980s and early 1990s, including the widespread adoption of advanced restraint designs such as multistage airbags, pretensioners, and force limiters. Substantial reductions in fatality and injury risks have been realized with the introduction of advanced restraints and improved crashworthiness. However, moderate and severe injuries (AIS2+) remain relatively common to the thorax, head, arm and hand/wrist for all ages (Forman et al., 2019). Moderate injuries (AIS 2) to the head and upper extremities are relative frequent. Head and upper extremity injuries are more likely than thorax injuries to result in permanent impairment (Kullgren et al., 2020). Impairing injuries reduces the quality of life for the victims and the societal cost for upper extremity injuries are substantial relative to injuries to other body parts (Mallory and Kender, 2019). In addition, these injuries are likely to remain in unchanged relative importance in future cars, including autonomous vehicles, if no actions are initiated.

3.1 Head Injuries

During the last decade, the discussion concerning long-term consequences of head injuries have accelerated, mainly in the context of sports, but also with respect to traffic accidents. Several studies are available today, also using different scales to quantify long-term consequences.

Selassie et al. (2008) used a population-based sample of persons with traumatic brain injuries (TBI) (South Carolina Traumatic Brain Injury and Follow-up registry) to estimate the disability rate. They estimated that around 43% had developed long-term disability. The disability rate was highest for fall injuries. For traffic accidents, pedestrian had highest rate of long-term disability followed by motorcyclists. Luchter and Walz (1995) studied police reported tow-away motor vehicle crashes in the US in 1993 and found that the injuries with greatest Functional Capacity Index (FCI) for survivors were unconsciousness, brain stem and cerebrum injuries. Injuries to the head is over-represented as leading to severe medical impairment (Malm et al., 2008). Also data from STRADA (Sjöö and Ungerbäck, 2007) concludes that head injuries among cyclists and pedestrians are the most common among the most severe medical impairment. Luchter and Walz (1995) studied the long-term consequences of head injury resulting from police reported tow-away motor vehicle crashes. The injuries with the greatest FCI for survivors were unconsciousness, brain stem, diffuse axonal injury, subdural hematoma, contusion and base fracture.

3.2 Upper Extremity Injuries

Injuries to the upper extremities have not decreased to same extent as other body regions, whereby they are among the relatively most frequent AIS2+ injuries today (Jakobsson and Lindman, 2010).

Analyzing crashes with Volvo Cars in Sweden, Jakobsson and Lindman (2010) found upper extremity fractures predominantly in frontal impacts and drivers tended to be more exposed. Fractures to the forearm, wrist and hand, in addition to the clavicula were most frequent. The main mechanisms of fractures to the wrist and the forearm were trauma to an outstretched, extended or clenched hand. Investigating upper extremity fractures in the CIREN database, 1997-2004, Conroy et al. (2007) showed that drivers and passengers had different upper extremity injury patterns and the direction of impact influenced the injury pattern. The authors identified the front vehicle interior as the most contributing injury source for forearm fractures. Harrysson & Cyren (2016) suggested that the hand and arms can impact many different areas of the instrument panel, around the steering wheel and the center stack being the most common impact area. Jernigan (2005) studied occupants exposed to full-powered airbags and concluded that they mostly sustained radius and ulna injuries. Wraighte et al. (2010) reported that clavicle fractures, as a result of seat belt loading, followed by wrist and forearm fractures were most common in frontal impacts, while in side impacts humerus, radius, ulna and clavicle fractures were most frequent. Otte (1998) studied the biomechanics of upper extremity injuries of belted car drivers and emphasized the need for car developments and crash test dummy test work.

Although usually not life-threatening, upper extremity injuries can result in long-term consequences. Upper extremities account for the second overall highest scoring of Permanent Medical Impairment (PMI) of degree 1% or more (PMI1+), given the event of injury/diagnosis (Stigson et al., 2015). Only lower extremities were found to have overall higher risk of PMI1+. Long-term consequences for upper extremity injuries were seen irrespectively of crash configuration. Kullgren et al. (2020) showed that upper extremity injuries often result in various degrees of permanent impairment with consequence in loss of quality of life and high societal costs. The Permanent Medical Impairment (PMI) is assessed on a scale from 1% to 99% (*Medicinsk invaliditet - skador, Svensk Försäkring*, 2020). As examples, amputation of a finger is graded from PMI 1% to 8%, amputation of a hand is PMI 37%, and amputation of one arm is up to PMI 51%. A functional reduction of the shoulder movement is graded up to 20%, elbow up to 37%, wrist from 3-32%.

4 Objective and Research Questions

With the overall purpose to increase the safety for vehicle occupants, pedestrians and bicyclists, in addition to those wearing helmets for other applications, by developing countermeasure for enhanced protection, the objective is to enhance understanding and evaluation of injuries leading to long-term consequences. Specifically, the project targets development of methods, tools and injury criteria for head and upper extremity injuries. The project will gain an insight on the relevance of the ability of various head injury criteria to predict head injuries leading to long-term consequences. In addition, a part of the project targeted quantification of identifying injury types leading to long-term consequences and how do these injuries relate to exposure and type of situation.

With the purpose of enabling prediction of injuries leading to long-term consequences in traffic, there is a need to identify injuries of importance and to create methods (tools, criteria etc). Specifically, this project will address the following research questions:

- Which are the injury types leading to long-term consequences and how do these injuries relate to exposure and type of situation?
- How can head injuries be predicted and prevented? Especially how can head injuries those leading to long-term consequences be predicted and prevented?
- Are the global injury criteria (BrIC, HIC and RIC) relevant measures for predicting head injuries with long-term consequences?
- How can upper extremity injuries be predicted and prevented, especially those leading to long-term consequences?
- Can human body models (HBM) be used to predict head and upper extremity injuries with long-term consequences?

5 Head and Upper Extremity Injuries; Methods and Countermeasures

5.1 A Method to Predict Long-term Consequences

A method to predict the risk for long-term consequences was developed based on the risk for permanent medical impairment (PMI). A risk for PMI of at least 1%, 5% and 10% for injuries with AIS scores of 1 through 5 for the body regions; head, face, cervical spine, upper extremities, thorax, thoracic spine, abdomen, lumbar spine, lower extremities and external (skin) was proposed by Malm et al. (2008). Figure 1 shows the table of risk for at least 1% PMI for the five AIS levels in the presented body regions.

Table 3- Risk of Permanent Medical Impairment (RPMI) on 1%+ level (i.e. 1-99%). Numbers in percent

	AIS1	AIS2	AIS3	AIS4	AIS5
Head	8.0	15	50	80	100
Cervical Spine	16.7	61	80	100	100
Face	5.8	28	80	80	n.a.
Upper Extremity	17.4	35	85	100	n.a.
Lower Extremity and Pelvis	17.6	50	60	60	100
Thorax	2.6	4.0	4	30	30
Thoracic Spine	4.9	45	90	100	100
Abdomen	0.0	2.4	10	20	20
Lumbar Spine	5.7	55	70	100	100
External (Skin) and Thermal Injuries	1.7	20	50	50	100

Figure 1. Table over risk for at least 1% PMI per body region and AIS level, taken from Malm et al (2008).

To predict the risk for LTC (RLTC), the predicted risk for AIS 1 – 5 for the body regions in Malm et al (2008) can be used directly with the predicted AIS risk from the HBM. Hence, if the HBM predicts an AIS2 injury for the upper extremities, the RLTC is the risk for injury scaled by 0.35. The overall RLTC can be calculated using the methodology in Rizzi (2016), developed to predict injury risk for motorcyclists,

$$\text{Risk for PMI} = 1 - (1 - \text{risk}_1) \times (1 - \text{risk}_2) \times \ldots \times (1 - \text{risk}_n)$$

where n is the number of body regions and risk is the highest risk for PMI in each body region. If the HBM predicts risk for several different injuries in the same body region, only the injury with the highest risk for PMI should be used or alternatively the highest AIS score.

A more detailed method would be to use the injury risk functions for AIS- injuries of specific injuries, such as arm fracture, humerus fracture etc., and scale by a factor to predict the RLTC. This would however require scaling factors, relating these injuries to RLTC. From data available in Malm et al (2008), this could be done, but requires a more extensive effort of sorting the data.

5.2 Addressing Head Injuries

Head injuries were addressed in several different studies. Different injury criteria based on kinematics (global) as well as tissue response (local) were evaluated and compared. Different test methods for helmets and vehicles were evaluated. In addition, development and evaluation of countermeasures were included in this project. A refinement of the head model developed at KTH Royal Institute of Technology was also included in the project.

5.2.1 Evaluation of Head Test Methods and Injury Criteria

5.2.1.1 Test Methods and Global Head Injury Criteria for Helmet Design

A test method mimicking a typical bicycle accident in which the bicyclist is falling to the ground with the aim to evaluate existing global kinematic head injury criteria for impact pulses related to helmet impacts was used.

The helmet was positioned on a head from the 50th percentile crash test dummy Hybrid III. The helmet/headform was dropped against a 45° angled surface in 6.2-6.3 m/s. Three impact locations were evaluated called Xrot, Yrot, and Zrot (Figure 2). These three impact locations cause main rotation around the main axis of the head, which is indicated by the name. In total 1728 helmet impacts were analyzed.

The acceleration pulses in the six degrees of freedom (x-,y-, z-. linear and angular acceleration) from these 1728 experimental tests were then applied to the KTH head model (Kleiven, 2007, 2006, 2002) to calculate the strain of the brain tissue. The skull was treated as rigid. The maximum Green-Lagrange first principal strain (MPS) in the brain tissue was evaluated. The MPS was then used as the baseline in the comparison to the different kinematical injury criteria.

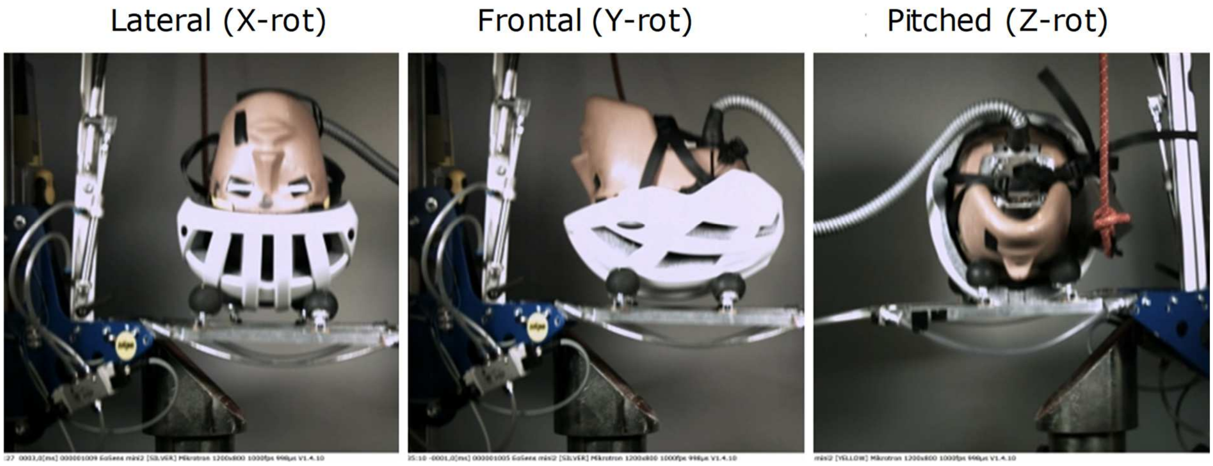


Figure 2. Three different impact areas with different impact directions.

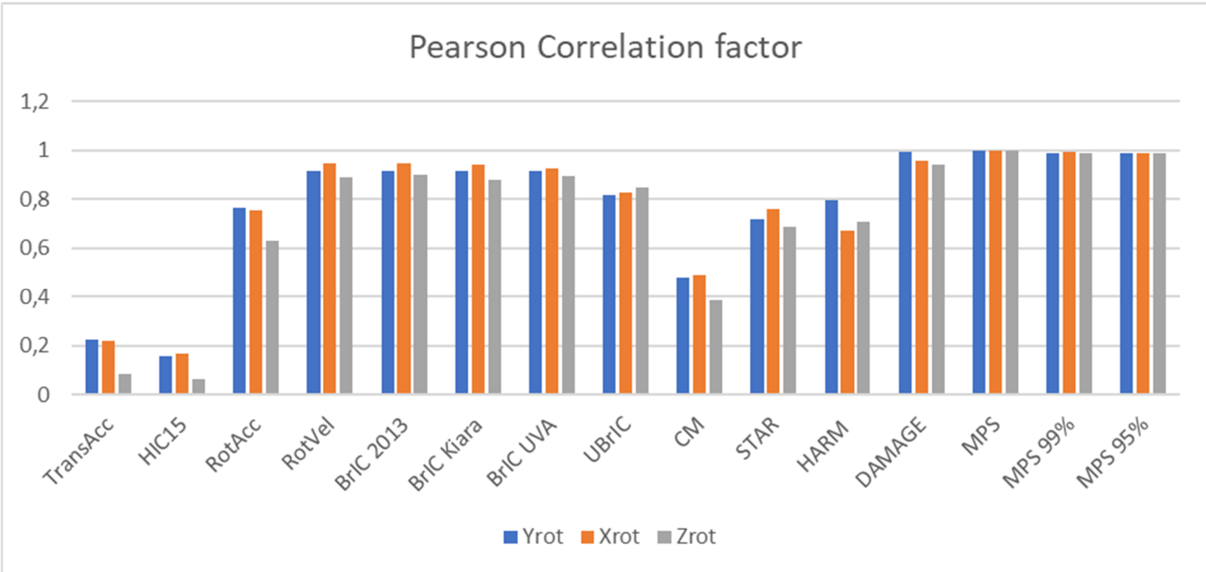


Figure 3. The Pearson's correlation (r^2) between the maximum strain of the KTH head model and different injury criteria. (BrIC – Brain Injury Criteria (Takhounts et al. 2013) and different variants of BrIC (BrIC Kiara (Giordano & Kleiven (2014) and BrIC UVA (Gabler et al., 2018a));)). UBrIC - Universal Brain Injury Criterion Gabler et al. (2018a); CM - Combined Metric (Funk et al., 2017); STAR - summation of tests for the analysis of risk (Bland et al., 2020); HARM - head acceleration response metric (Bailey et al., 2020); DAMAGE - Diffuse Axonal, Multi-Axis, General Evaluation (Gabler et al., 2018b)).

The injury criteria were chosen due to that they today are the most known used injury criteria to predict injury to the human head. The linear correlation between the computed peak magnitude of MPS from the KTH head model and the different kinematic injury criteria was then compared using the Pearson's correlation coefficient of determination (r^2) (Figure 3). It was found that all criteria with a dominant rotational velocity component, correlates well to the computed MPS with the KTH head model. The best correlation was shown for DAMAGE[1] (0.96-0.99). It was also found that BrIC[2] and UBRIC[3] show the same trends as MPS. However, criteria based on only resultant acceleration or rotational acceleration did not show the same trend as MPS.

[1] Diffuse Axonal, Multi-Axis, General Evaluation (DAMAGE) (Gabler et al., 2018b)
[2] Brain Injury Criterion (BrIC) (Takhounts et al., 2013)
[3] Universal Brain Injury Criterion (UBrIC) (Gabler et al., 2018a)

5.2.1.2 Evaluation of Brain Injury Models for Bicycle Helmet Rating and Ranking

When working with the updates of the current test standard for bicycle helmets, an important discussion is if the pass and fail criteria should be based on kinematic-based or model-based injury criteria. Therefore, the objective of this part of the project was to determine the influence of helmet ranking and rating using eight existing brain injury models and eight existing injury criteria based on global kinematics for bicycle helmets tested in oblique impact conditions against a hard surface (Fahlstedt et al., 2021). The term 'ranking' indicates the individual position of the helmet amongst the sample tested when organized based on the assessment criterion. The term 'rating' is used to describe the category to which a helmet belongs when the helmets are clustered into different groups depending on the assessment criterion.

Experimental tests presented in a previous study (Stigson et al., 2017) were used. The experimental tests included seventeen different conventional bicycle helmets from the Swedish market (2015) (Helmet A to Q). The test method was the same as presented in 5.2.1.1. The kinematics from the experiments were applied to eight different head models (Figure 4). The six components of the linear and the angular accelerations were prescribed to the rigid skull at the head center of gravity of each model.

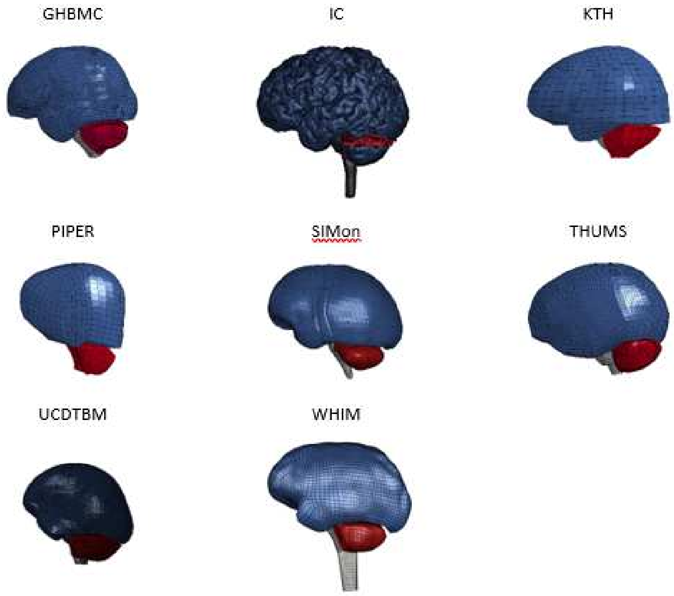


Figure 4. The geometry and mesh of the different head models

The eight different models used in the study and the outputs that were evaluated is presented in Table 1. All eight models have been developed separately and with different strategies. Since this is a study that evaluates existing head models, the outputs suggested by the developers were used, and therefore different outputs are used across the different models.

The ranking and rating of the different head models were also compared to injury criteria based on the global kinematics. The linear correlation between the peak magnitudes of MPS of the models was evaluated with Pearson's correlation coefficient of determination (r^2). For the nonparametric data, the ranking of the helmets (from 1 to 17) based on the performance of the helmets, the Kendall's tau (Kendall, 1938) was evaluated. Kendall's tau has a value between 1 and -1 where 1 means precisely the same order, and -1 means opposite order.

Table 1. The different models evaluated, and the output used.

Model	**Reference**	**Output**
Global Human Body Model Consortium (GHBMC) M50-O v4.3	(Mao et al., 2013)	95th percentile true strain and cumulative strain density measure (CSDM) (Threshold 25%)
Imperial College model (IC)	(Ghajari et al., 2017)	90th percentile G-L strain and G-L strain rate
The isotropic version of KTH model	(Kleiven, 2007)	100th percentile G-L strain
PIPER 18-year-old model	(Li and Kleiven, 2018)	95th percentile G-L strain
SIMon model	(Takhounts et al., 2008)	95th percentile true strain and CSDM (Threshold 25%
Total Human Model for Safety (THUMS) v.4.02	(Atsumi et al., 2016)	95th percentile G-L strain
UCDBTM v2.0	(Trotta et al., 2020)	100th percentile G-L strain
The isotropic version of the Worcester Head Injury Model (WHIM) v.1.0	(Ji et al., 2015; Zhao and Ji, 2020)	95th percentile G-L strain

Helmets were graded from 1-star (highest average value of injury metrics) to 4-star (lowest average value of injury criteria) based on the percentile value of the values for all seventeen helmets (Table 2).

Table 2. The threshold for the different number of stars assigned to the helmets.

4-star (best safety performance)	<25th percentile value
3-star	25th – 50th percentile value
2-star	50th – 75th percentile value
1-star (worst safety performance)	>75th percentile value

Pearson's r^2 varied between 0.53 and 0.99 between the injury criteria for the different head models. Most of the head models had a r^2 above 0.8 when compared to the other models.

None of the head models showed linear correlation to the kinematic-based criteria based on linear acceleration (peak linear acceleration and HIC[4]) or a combination of linear acceleration and angular velocity (STAR*[5]). However, most head models' injury metrics had an r^2 above 0.8 for the kinematic-based criteria that were based on angular motion.

Kendall's tau varied between 0.50 and 0.98 for the different head model outputs. The tau coefficient was relatively high between all head models (>0.8) except for the UCDTBM, IC, and SIMon models. A Kendall's tau larger than 0.8 was found for most of the head models when compared to kinematic-based criteria based on angular velocity. All head models showed a Kendall's tau below 0.62 when compared to peak linear acceleration, HIC and STAR*. The helmet rating for the models is presented in Figure 5.

[4] *Head Injury Criterion (HIC) (Versace, 1971)*
[5] *STAR* is a modification of STAR presented by Bland et al. (2020), only based on three impacts instead of 12 impacts.*

	1-star				2-star					3-star				4-star			
GHBMC	P	K	G	N	B	C	A	I	F	E	J	L	O	H	M	D	Q
GHBMC CSDM	P	G	K	B	N	A	C	I	F	J	E	L	O	H	M	D	Q
IC	G	I	K	B	P	A	E	F	C	H	N	J	L	M	Q	O	D
IC Strain Rate	G	P	K	B	N	I	C	F	A	E	J	H	L	O	M	D	Q
KTH	P	G	K	B	N	C	F	I	A	E	J	O	L	H	M	Q	D
Piper	P	G	N	K	B	C	A	F	I	E	J	L	O	H	M	D	Q
SIMon	P	K	B	G	N	C	A	F	E	I	J	L	H	O	M	D	Q
SIMon CSDM	P	K	N	G	B	C	A	F	I	E	J	L	O	H	M	D	Q
THUMS	P	K	N	B	G	A	C	I	F	J	E	L	O	H	M	D	Q
UCDBTM	G	B	I	P	D	K	E	F	C	H	N	A	M	J	Q	L	O
WHIM	P	G	K	B	N	C	A	F	I	E	J	L	O	H	M	D	Q

Figure 5. The rating of helmets based on the different head models. The different colors are indicating the different helmets. 10 of 17 helmet had the same ranking for 10 model outputs or more, but there was also some difference e.g., helmet D was ranked as 2-star by UCDBTM but ranked as 4-star for the other models.

5.2.1.3 Evaluation of Different Brain Injury Models for Short and Long Duration Pulses

The aim of this study was to evaluate different injury criteria based on global or local response for short and long duration impact pulses, where a short duration impact pulse is taken from helmet tests (duration 5-20 ms) and long duration impact pulse is taken from frontal collisions (>200ms).There has been discussion about that the different head models are better suitable for impulses with short duration (5-20 ms) compared to impulses with longer durations (>200 ms). Therefore, the objective of this part of the project was to evaluate different injury criteria based on global or local response for short and long duration pulses.

Three different existing head models (KTH, THUMS and WHIM (Table 1)) and eleven different injury criteria based on global kinematics were evaluated. The evaluation included a comparison of linear correlation between different injury criteria, correlation in the ranking (comparing the individual position of each pulse), and correlation in rating where the pulses were group into several categories based on the response. The output from the FE models presented in Table 2 together with injury risk functions were used for the rating.

The first dataset of pulses was taken from bicycle helmet oblique drop tests presented in 5.2.1.2. Hereafter the dataset is called HT. The second dataset of head pulses was recorded from the driver Hybrid III crash test dummy in 50 frontal collision crash tests against a rigid barrier with small overlap (25% overlap) and moderate overlap (40% overlap) performed by Insurance Institute for Highway Safety (IIHS). Hereafter this dataset is called IIHS. The impact velocity was 64 km/h.

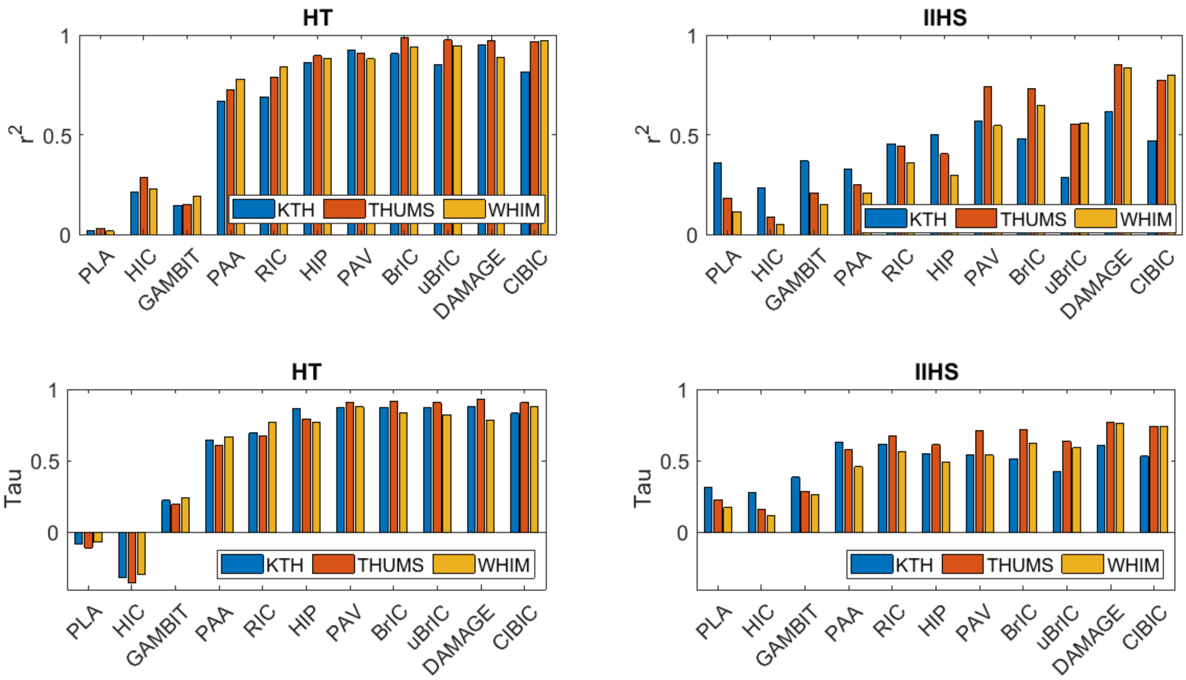


Figure 6. Top row show the linear correlation (r^2) between the different head models and the global injury criteria for HT (left) and IIHS (right) pulses. The second row is the correlation in ranking (Tau) between the different head models and the global injury criteria for HT (left) and IIHS (right) pulses

A higher Pearson's coefficient of determination r^2 was found for HT pulses compared to the IIHS pulses for all models and global injury criteria except GAMBIT[6] for the KTH and THUMS model and peak linear acceleration for all models, but in these cases, the r^2 was low (below 0.4) (Figure 6). An r^2 above 0.8 was found for HIP[7], CIBIC[8], DAMAGE, uBrIC, BrIC, and peak angular velocity for all three models for the HT pulses. The IIHS pulses showed only an r^2 above 0.8 for DAMAGE-KTH/THUMS model.

The r^2 was high between the different brain injury models for the HT with an r^2 above 0.91. Meanwhile, the correlation between models was lower for the IIHS data but still relatively high, with r^2 between 0.68 and 0.81. The highest r^2 for the HT and IIHS pulses was between THUMS and WHIM models.

KTH and THUMS models showed the same rating in 50 of 51 of the HT pulses and 36 of 50 of the IIHS pulses. KTH and WHIM models showed the same rating in 28 of 51 of the HT pulses and 39 of 50 for the IIHS pulses. Meanwhile, THUMS and WHIM models had the same rating in 29 of 51 for the HT pulses and 35 of 50 for the IIHS pulses.

The KTH model had the most same rating with global injury criterion RIC[9] (34 of 50) for the IIHS pulses and DAMAGE for the HT pulses (44 of 51). The THUMS model instead had the most same rating with CIBIC (33 of 50), closely followed by RIC (32 of 50) and peak linear acceleration (32 of 50) for the IIHS pulses, and with DAMAGE for the HT pulses (43 of 51). Meanwhile, WHIM had the most same rating with peak linear acceleration and RIC (34 of 50) for IIHS pulses and CIBIC (44 of 51) for the HT pulses.

[6] *Generalized Acceleration Model for Brain Injury Threshold (GAMBIT) (Newman, 1986)*
[7] *Head Impact Power (HIP) (Newman et al., 2000)*
[8] *Convolution of Impulse response for Brain injury Criterion (CIBIC) (Takahashi and Yanaoka, 2017)*
[9] *Rotational Injury Criterion (RIC) (Kimpara and Iwamoto, 2012)*

5.2.1.4 Comparison KTH Head Model Predictions with Free Motion Headform

The aim was to evaluate the head injury risk predicted with the KTH head model to global injury measures (HIC, BrIC; GAMBIT; KLC[10], RIC and UBrIC) predicted with the Free Motion Headform (FMH).

A of total 36 sled interior simulations using SAFER HBM were performed. The simulations were varied from six reference simulations i.e. three impact speeds 15, 17 and 19 km/h with two different barrier angles 30° and 40°. For each reference simulation a head impact surface was mounted in front of the B-pillar with two thicknesses, 1.7mm and 2.7mm. The impact surface was oriented using three different impact angles, 45°, 65° and 85°, relative the direction of travel of the vehicle.

The HBM head centre of gravity (CoG) position, i.e. x_{CoG}-, y_{CoG}- and z_{CoG}-coordinates together with the direction (measured as the angle of the head towards the vehicle horizontal and sagittal plane) were obtained 4-8ms prior to impact to position the FMH model correctly in space (Figure 7). The impact speed from each of the interior simulation was used as initial speed for the FMH simulations.

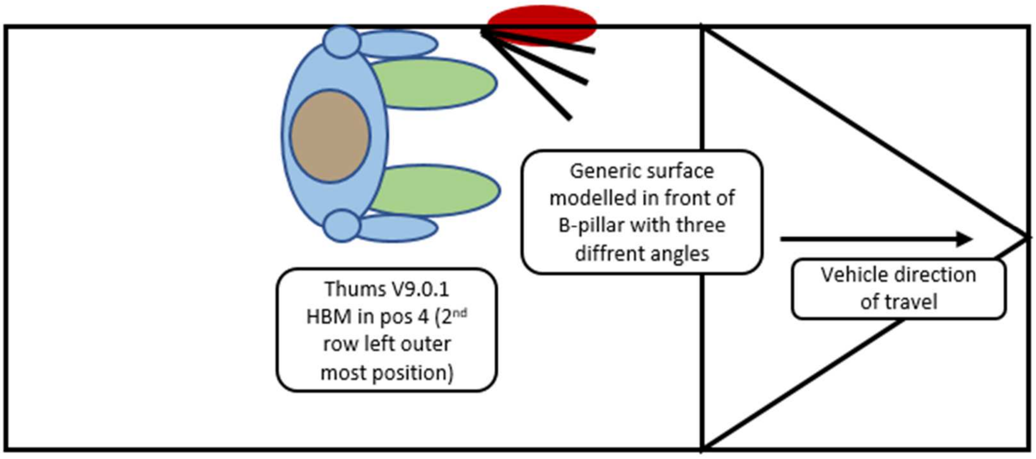


Figure 7. Schematic figure of interior sled simulation with SAFER HBM v9.0.1 and generic surface modelled in front of B-Pillar

A comparison of the following global injury criteria; HIC, BrIC, UBrIC, GAMBIT, KLC, RIC (Antona-Makoshi, 2016) and pure head angular velocity (resultant and component wise) between the SAFER HBM and FMH simulation was performed.

It was found that the head impact zones obtained from the sled simulations varied for the different barrier impact angles. The 30° impact angle resulted in a head impact on the lateral surface on the B-pillar, see the blue circles in Figure 8. The head of the SAFER HBM “slid” off the B-pillar surface. Low linear acceleration and HIC15 values were obtained. However, the risk for rotationally induced head injury increased resulting in high BrIC values.

[10] *Kleiven’s Linear Combination (KLC) (Kleiven, 2007)*

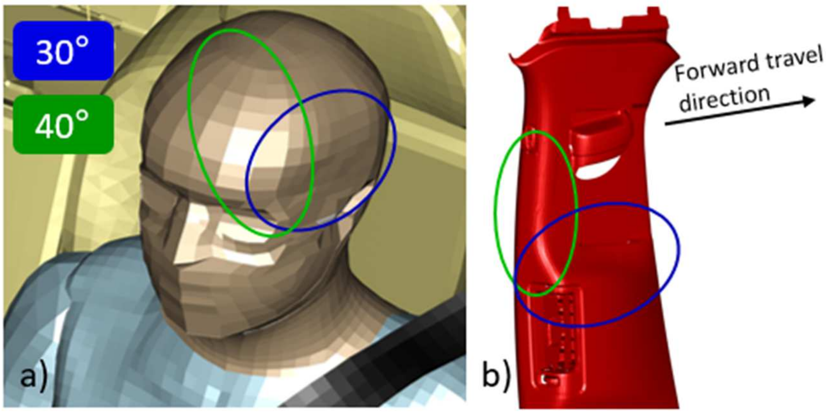


Figure 8. Impact zones on SAFER HBM and B-pillar of vehicle interior trim.

The 40° impact angle resulted in a different head impact to the B-pillar than the 30° impact angle. The impact area was more rearwards and higher up on the impact zone on the B-pillar (Figure 8). The linear acceleration and HIC15 were higher than for the 30° impact. However, the rotationally induced measure BrIC was lower than for the 30° impact.

The 36 SAFER HBM sled test results were compared to 36 FMH simulations in which the FMH impacted a generic surface placed in front of the B-Pillar. Correlation for the linear acceleration and HIC15 was obtained between the predictions from the SAFER HBM and FMH. However, there was no correlation for the rotationally induced injury measures BrIC, GAMBIT, KLC, RIC and UBrIC.

The last part of the study compared 36 SAFER HBM and 36 FMH simulations respectively impacting a generic surface placed in front of the B-pillar. This comparison showed a good correlation of the linear acceleration, SAFER HBM HIC_{15} vs. FMH HIC(d), but there was no correlation between the rotation based global injury measures (BrIC, GAMBIT, KLC, RIC and UBrIC) between SAFER HBM and the FMH method.

Both GAMBIT and KLC includes linear acceleration in the equation. That can explain the somewhat closer correlation between the SAFER HBM and FMH prediction for those measures. For the other injury measures that are based on only rotational measures RIC, BrIC & UBrIC there was poor correlation.

5.2.1.5 Head Injury Risk Evaluation for Drivers in a Frontal Oblique Impact

The aim of the study was to evaluate the head injury risk predicted with both global and local injury measures for a state-of-the-art driver protective system in an oblique frontal loadcase, according to the proposal by NHTSA. The loadcase comprises an impact at 90 km/h, by a moving deformable barrier at 15° angle with 35% overlap into the stationary vehicle. The kinematics from such a test was applied to the rigid base of a detailed interior model of a large passenger vehicle (Figure 9). The vehicle model included belt system, driver airbag and inflatable curtain. The THOR v1.5.1 model and the SAFER HBM v9.0.1. were positioned as driver in the vehicle environment.

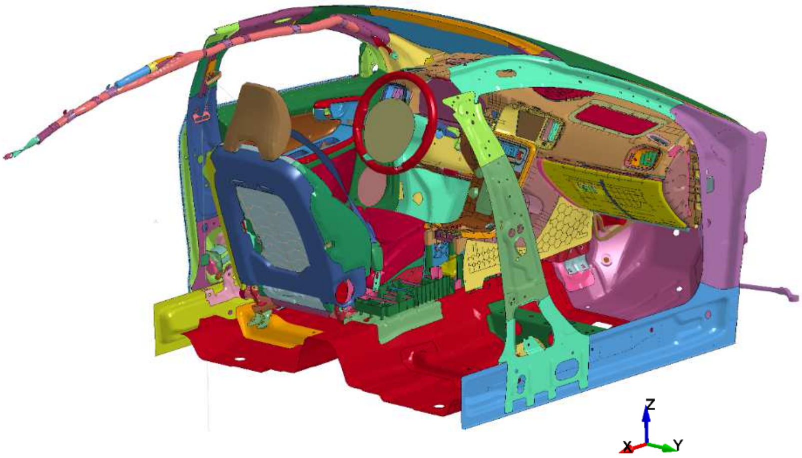


Figure 9 The vehicle environment used in the NHTSA oblique frontal impact load case

The parameters varied include the coefficient of friction between the occupant model and the interior (airbag, seat, etc.) and variations in direction of pulse application. The friction was varied between 0.3 and 0.6, with the baseline at 0.45. The pulse impact direction was achieved by rotating the x- and y-directions about the z-axis by ±10°, 20°, 30° and 40°, to study. It is a simplification due to the fact that the pulse will vary depending on impact angle. However, for robustness evaluation and relative comparison between various impact angles the simplification was considered appropriate.

Head injury risks of AIS2+ were calculated using BrIC, UBrIC, DAMAGE and 1st principal Green-Lagrange brain tissue strain in the KTH head model.

The SAFER HBM had more interaction with the driver airbag than the THOR model, giving a larger effect of either sliding past (at low friction) or sticking to the airbag (at high friction) and hence more changes in head kinematics.

In the baseline simulation, a low injury risks (AIS2<15%) was calculated using all injury criteria but BrIC, which showed a 64% and 71% AIS2+ risk for THOR and the SAFER HBM, respectively. The THOR model showed small influence of friction (Figure 10**Error! Reference source not found.**). Larger variations were seen when using the SAFER HBM and the global injury risk criteria (Figure 11**Error! Reference source not found.**), with up to 73% change in AIS2+ risk using DAMAGE. The change was however smaller using the strain-based criteria (max 25%).

The SAFER HBM had more interaction with the driver airbag than the THOR model, giving a larger effect of either sliding past (at low friction) or sticking to the airbag (at high friction) and hence more changes in head kinematics.

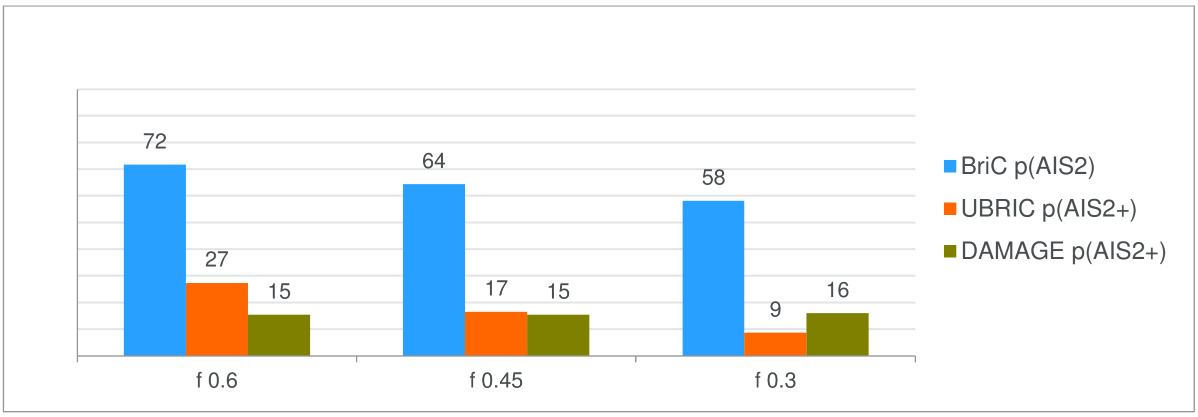


Figure 10. Calculated BrIC, UBrIC and DAMAGE head injury risks using the THOR model, varying the coefficients of friction (0.6, 0.45 and 0.3) between the model and the interior

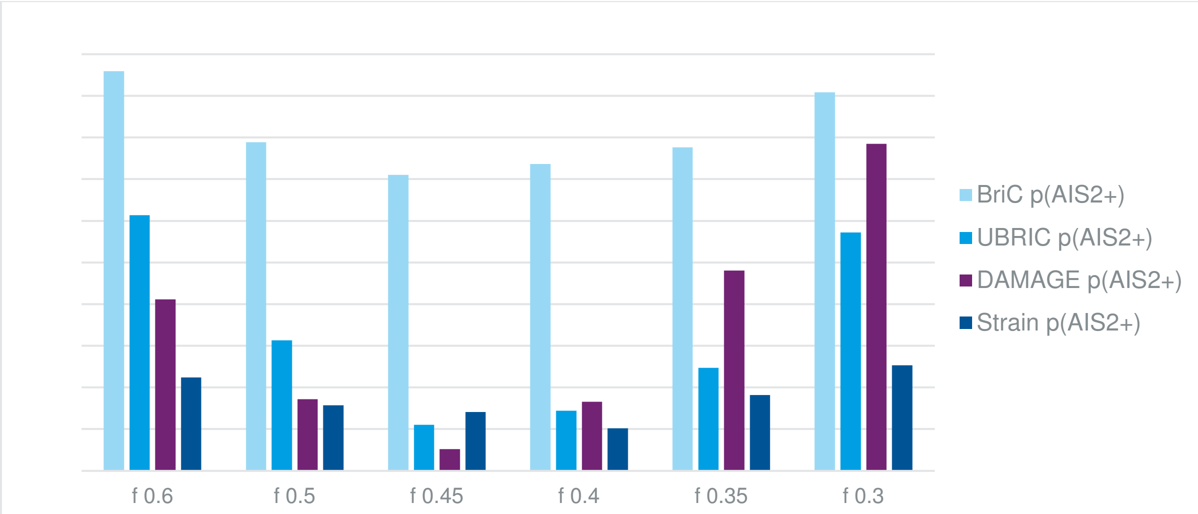


Figure 11. Calculated BrIC, UBrIC, DAMAGE and strain-based head injury risks using the SAFER HBM, varying the coefficients of friction (0.6, 0.5, 0.45, 0.4, 0.35 and 0.3) between the model and interior.

The variation of impact angle showed relatively large changes in injury risk for angles less than -20° and greater than 30° for the THOR model predictions (Figure 12). The SAFER HBM predicted significant changes in injury risk at negative angles and above 30° for the global injury criteria. However, the SAFER HBM model predicted relatively small changes in injury risk based on strain between -20° - 20°.

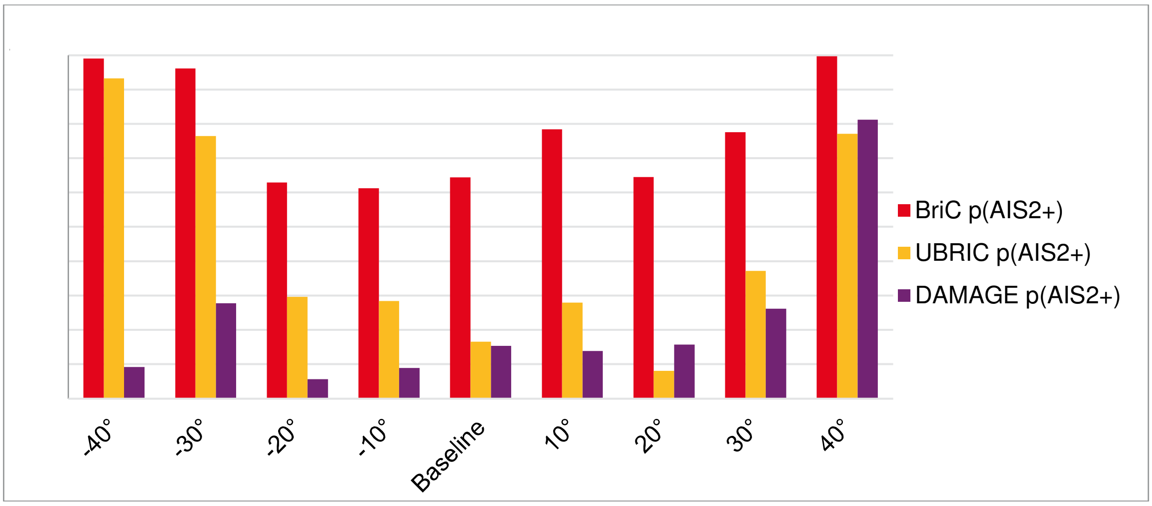


Figure 12. Calculated BrIC, UBrIC and DAMAGE head injury risks using the THOR model, impact direction.

5.2.1.6 Small Overlap vs Moderate Overlap for Head Injury Risk Evaluation

The aim was to compare the head injury risk predicted by SAFER HBM with the head injury risk predicted with HIII and THOR in small and moderate overlap crashes.

To evaluate global and local head injury criteria, a continuation of previous work based on full scale vehicle crash tests was done (Slagmaat Mv et al., 2019), where crash test dummy head kinematics was used to evaluate a number of head injury criteria. The vehicle crash tests were performed by the Insurance Institute for Highway Safety (IIHS) consisting of small and moderate overlap vehicle crash tests with an instrumented Hybrid III (HIII) in the driver seat.

a) 

b) 

Figure 13. Crash test configuration for IIHS small overlap (left image) and moderate overlap (right) tests.

From a selection of vehicle tests (Table 3), the vehicle crash pulse data (EDR) was used to evaluate the possibility to reproduce the tests with simulations.

Table 3 Selected IIHS small and moderate overlap crash tests for simulation.

Moderate Overlap		Small Overlap	
CEF1304	2014 Fiat 500L	CEN1414	2014 Fiat 500L
CEF1305	2014 Jeep Cherokee	CEN1404	2014 Jeep Grand Cherokee
CEF1306	2013 Chevrolet Spark	CEN1335	2013 Chevrolet Spark
CEF1605	2016 BMW X1	CEN1608	2016 BMW X1
CEF1608	2016 Mazda CX-3	CEN1618	2016 Mazda CX-3

The environment included a driver seat, belt system, instrument panel, steering wheel, driver airbag and inflatable curtain. The firing times for the airbags and belt pretensioners were taken from each test. All tests where simulated using FE-models of the Hybrid III v8.0.1, the THOR v1.6 and the SAFER HBM v9.0.

The head injury risk was assessed using BrIC from the three models, as well as the maximum principal strain of the brain tissue in the SAFER HBM.

Generally, there was good agreement for predicted head kinematics (accelerations and rotational velocity) between the three models. In particular there was good agreement between the HIII and THOR model predictions. No clear trend was obtained when comparing the injury risk, and no separation was possible between the small and moderate overlap tests when evaluating BrIC. When looking at the strain in the brain of the SAFER HBM, the small overlap tests resulted in lower risk of injury in all cases. It could not be determined what is the root cause for the difference seen from field data where small overlap crashes show a higher proportion of severe head injuries compared to moderate overlap (Mueller et al., 2015). There were however limitations in the EDR data, only containing linear acceleration data in the frontal and lateral direction.

5.2.2 Development of Head Injury Risk Functions

Risk functions are used to connect the outputs from the computer models to real accidents and assess the injury risk. Within this project the focus has been on risk functions for concussions and skull fractures.

5.2.2.1 Concussion

With the aim to develop risk functions for the KTH head model enabling prediction of mild traumatic brain injuries (concussion), where different datasets were used. For the KTH model, previously two datasets based on helmeted impacts from American football (Sanchez et al. 2019) or un-helmeted impacts in Australian football and rugby (Patton et al., 2013) have been used for inputs to generate risk functions for concussions. These previous injury risk functions have been a combination of AIS1 (less severe without loss of consciousness) and AIS2 (more severe with loss of consciousness up to less than 60 minutes) concussions.

Little is known about how the datasets used for the KTH model from un-helmeted impacts in Australian football and rugby, and helmeted impacts in American football, differs and how the two datasets are influencing the injury risk functions. Both datasets involve AIS1 as well as AIS2 concussions, but until now no separate injury risk functions were developed for AIS1 and AIS2 concussions. With this background the objectives of this part of the project were to:

1. Analyze the differences between two datasets of concussion for injury predictions to better understand the influence on injury risk functions.
2. Develop injury risk functions for concussions separated in AIS1 and AIS2 level.

The two datasets that were used was based on the data from McIntosh et al. (2014) (un-helmeted impacts in Australian football and rugby), and the data from Sanchez et al. (2019) (helmeted impacts in American football). The kinematics from these datasets were applied to the KTH head model to calculate the strain of the brain tissue. Both the maximum strain in the principal direction (MPS) and the projected strain along strain tensor along the fiber (MAS). The risk functions were developed by survival analysis with Weibull distribution. Also risk functions for kinematic-based injury criteria were developed and evaluated.

The kinematics of the dataset based on impacts in Australian football and rugby had been estimated previously by rigid body simulations (Fréchède and McIntosh, 2009). In summary there was 9 cases with AIS1 concussions, 18 cases AIS2 concussions, and 13 non-injured cases. More information about the dataset can be found in previous publications (Fréchède and McIntosh, 2009; McIntosh et al., 2000, 2014; Patton et al., 2013).

The kinematics of the second dataset based on American football from the National Football League (NFL) were based on previous studies that used Hybrid III crash test dummies to reconstruct the impacts. The dataset included 18 cases with AIS1 concussions, 2 AIS2 concussions, and 33 non-injured players. More information about the reconstructions and the kinematics can be found in previous publications (Newman et al., 1999, 2005; Pellman et al., 2003; Sanchez et al., 2019).

Both Australian and NFL datasets showed a significant difference between injured and non-injured group for resultant peak linear acceleration (PLA), peak angular acceleration (PAA), and peak angular velocity (PAV) ($p<0.001$) (Figure 14). Both the datasets from Australia and NFL showed a significant difference between non-injured and injured cases for all brain tissue predictors evaluated (Figure 14). For the non-injured cases no significant difference was seen between the Australian and NFL data for whole-brain MPS and white matter MAS.

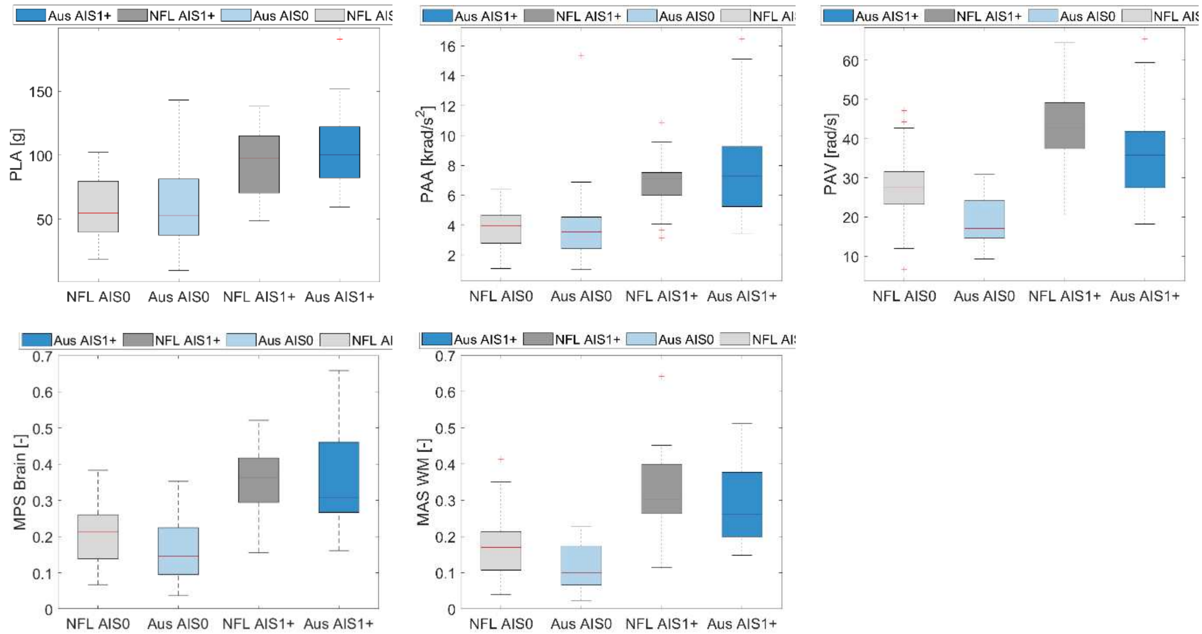


Figure 14. Boxplots of the peak resultant linear acceleration (PLA), peak resultant angular acceleration (PAA), peak resultant angular velocity (PAV), maximum first principal strain (MPS) for whole brain, and maximum axonal strain (MAS) for the white matter (NFL – gray and Aus – blue) for the non-injured (AIS0) and the injured (AIS1+) group.

For all predictors, the 50% injury risk was higher when the injury risk functions were based on the NFL data. The injury risk functions divided into AIS1 and AIS2 concussions are presented in Figure 15.

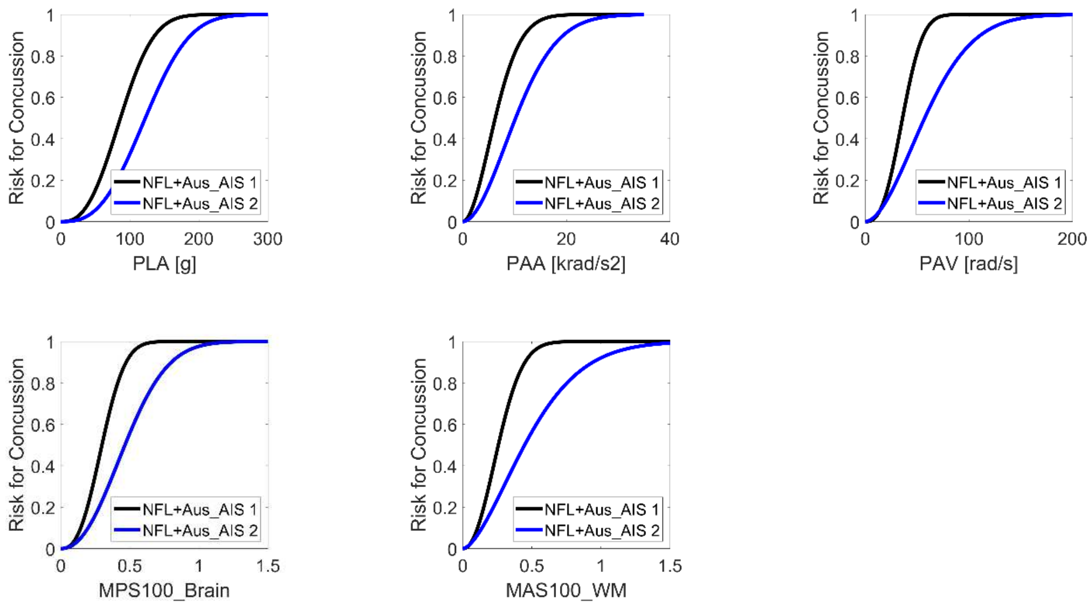


Figure 15. Injury risk functions for AIS1 and AIS2 concussions.

5.2.2.2 Skull Fracture

The aim was to enable skull fracture prediction with the KTH head model. A publication by Wood (1971) presented several hundred individual compact skull bone fracture strains. Individual fracture strain values were generated from the published plots. The data was fitted to 20 commonly used theoretical distribution functions. An Anderson-Darling test (AD-test) was chosen to evaluate the goodness of fit for the data set generated. Best fit was obtained for the Logistic, Logistic and Normal distribution functions for the various strain values. A large p-value for the AD-test indicates a close match to the data. The largest p-values were found for the Logistic, Loglogistic and Normal distribution functions for the various strain rates. Overall, the Logistic distribution showed the largest p-value indicating good fit with the data (Table 4).

Table 4. Anderson-Darling goodness of fit test p-values for the distribution functions having the highest AD test values.

	QS - 0.016 s^{-1}	**QS - 0.18 s^{-1}**	**Dyn - 2.8 s^{-1}**	**Dyn - 30.8 s^{-1}**	**Dyn - 105.6 s^{-1}**
Distribution function	*p-value AD-test*	*p-value AD-test*	*p-value AD-test*	*p-value AD-test*	*p-value AD-test*
Logistic	0.81	0.78	0.98	0.72	0.99
Loglogistic	0.92	0.55	0.84	0.91	0.97
Normal	0.60	0.85	0.96	0.65	0.99

No significant difference was found for between the three highest rates and between the two lowest rates (Figure 16). Therefore, risk functions were only attempted for data lumped into the a “quasi-static” group containing data from the two lowest strain rates and a “dynamic” group containing the three highest strain rates (Figure 16).

The cumulative distribution function (CDF) of the two-parameter Logistic distribution was used as an injury risk function and is given by:

$$\frac{1}{1+e^{-\left(\frac{x-\mu}{s}\right)}} \qquad (eq.1)$$

Where x is the fracture strain, s is the scale parameter and μ is the location parameter.

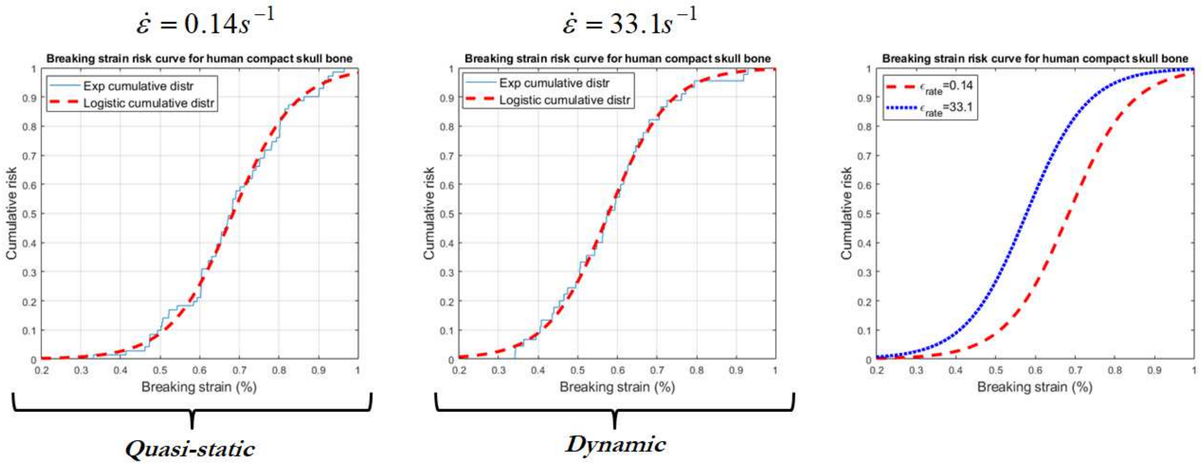


Figure 16. Fracture strain risk functions based on a two-parameter Logistic distribution (red dashed lines) generated from the experimental cumulative distribution function (light blue solid lines) for quasi-static and dynamic strain rates. Right Figure: The risk functions for quasi-static and dynamic rates together.

5.2.3 Refinement of the KTH Head Model

The aim was to refine the KTH head model and develop a spinal cord for improved injury prediction capability of the model.

A FE cervical spinal cord model, hereafter referred to CSC, has been developed to be implemented in the KTH neck model (Figure 17). The evaluation of the CSC model was performed from a previous study Yuan et al. (1998). The experiment was performed on five volunteers that were placed supine within an MRI scan. A posterior neck surface coil was used to obtain the images of the cervical spine and part of the brain.

The head of the volunteers placed in the MRI scan was attached to a close-fitting headgear with a nylon rope attached to the top of the scan. The rope prevented the head from lowering or slipping back. The end of the rope was secured by one of the stops in a slotted box. The experiment started as the volunteer comfortably flexed the head where a reference image was created. Afterwards, one stop was released from the slotted box to allow the head to drop from one position to another and create the deformed image. This procedure was performed until the head rested on the padded surface coil. The entire flexion created by the five volunteers was between 40° and 67°.

The experimental strain of the CSC was established as a linear regression of the mean of the five volunteer’s data. The posterior strain increases with increased head flexion angle and the positive values indicate an extension of the posterior surface. The tied contact surface shows a similar strain increment as the experiment while the strain for the automatic surface differs more as the strain values are smaller at higher head flexion angles. Therefore, a better correlation with the experimental data can be observed with the tied contact than the automatic contact. For the anterior strain less good correlation with experiment was seen for the anterior motion, where the no deformation was seen in the simulations which was seen in the experiments.

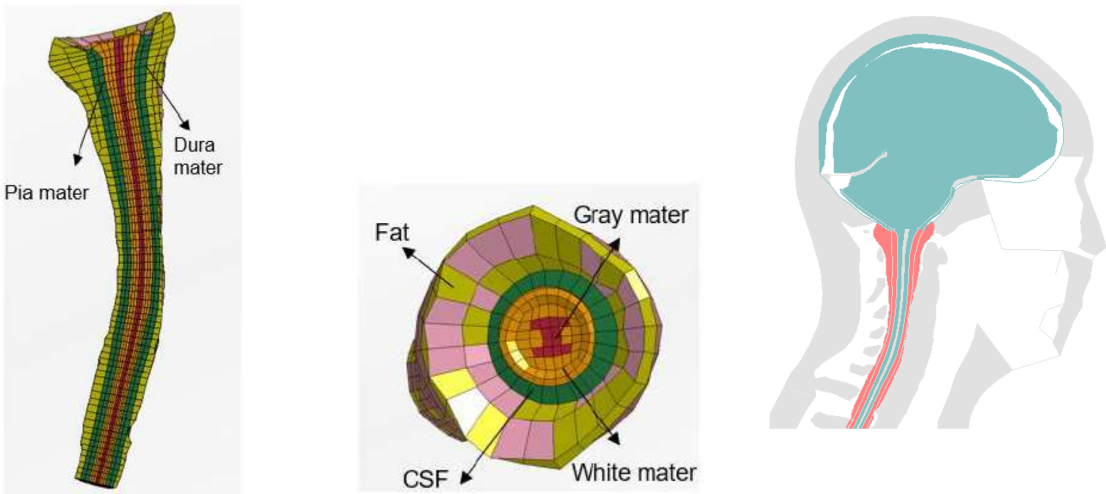


Figure 17. The different parts of the spinal cord and the spinal cord integrated to the SAFER HBM model with KTH head and neck

The C0-skull base junction of the SAFER HBM model was re-meshed. The earlier version of the SAFER-THUMS model had separately meshed C0 vertebrae of the cervical column and the skull base. To facilitate future analysis of skull base fractures, the mesh of the skull base and C0 vertebrae was modified so that a continuous mesh was obtained.

5.2.4 Countermeasure Development and Evaluation for Head

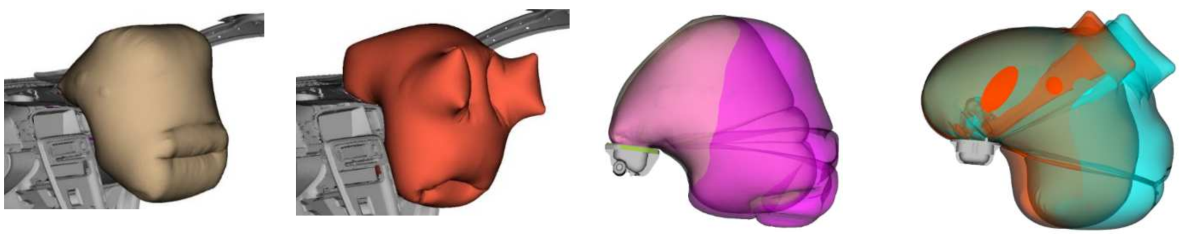
5.2.4.1 Concepts for Head Injury Risk Reduction in Frontal Oblique Impacts

The aim was to evaluate the potential injury reducing benefits for the head for various occupant protection concepts developed for oblique frontal impacts.

The SAFER HBM v9.0.1 model was positioned in the front passenger seat in a detailed vehicle interior model including passenger side airbag (PAB) and seatbelt with load limiters and trigger times corresponding to a state-of the art protective system. The potential injury reducing benefits for the head by various designs of passenger side airbags (PAB) were evaluated (Figure 18). Baseline passenger side airbags as well as passenger side airbags designed for occupant protection in oblique impacts were evaluated. Novel airbag systems such as the seat centric restraint system and far side airbag was evaluated (Figure 19). In addition, the potential benefits of seat belt modifications were also evaluated (Figure 19).

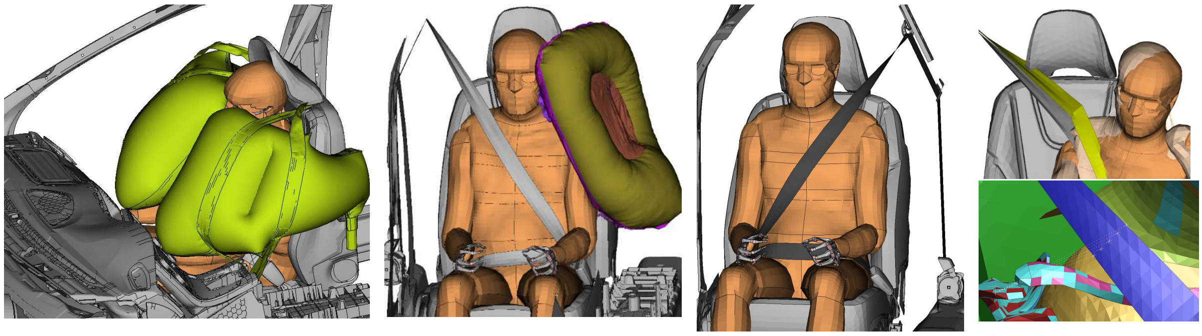
The different configurations evaluated were:

- Baseline PAB (120L)
- Oblique PAB (99L)
- The baseline PAB scaled in size 1.2 times in the direction of the passenger (Volume Scaled to 140L).
- A PAB designed for the NHTSA oblique load case (Oblique PAB)
    - Scaled in the same way as the original PAB (Volume Scaled to 110L)
    - Combined with increased load limiter level (4.4 kN)
    - Combined with increased load limiter level (6.65 kN)
    - Combined with increased load limiter level (4.4 kN) and reversed seat belt (D-Ring on the left side of the passenger)
    - Combined with a far side airbag (FSAB)
- A Seat Centric Restraint System (SCRS)
    - Combined with an increased load limiter level (4.4 kN)
- A Far Side Airbag (FAB)
- A rigid constraint added between a row of nodes on the seatbelt and the scapula of the SAFER HBM v9.0.1 (constrained belt)



Baseline PAB 120L	Oblique PAB 99L	Scaled Baseline PAB (purple) Compared to baseline (pink)	Scaled Oblique PAB (turquoise) Compared to baseline (orange)

Figure 18. Baseline PAB, Oblique PAB, Modified PAB and Modified Oblique PAB



Inflated SCRS　Far Side Airbag　Reversed seat belt　Constrained seat belt

Figure 19. Evaluated Countermeasures

For all evaluated system modifications, a reduction of more than 50% was obtained for HIC15 (left plot in Figure 20). Prediction of head injury risk reduction on BrIC was however relatively small, with all system showing an AIS2+ injury risk of over 80%,except for the SCRS showing a 40% risk of AIS2+ (middle plot in Figure 20).

The AIS2 risk based on strain (right plot in Figure 20) showed larger reduction and for more of the system variations. The scaled oblique PAB (1st purple bar) and combined with either constrained belt (green bar) and FSAB (turquoise bars) reduce the AIS2 injury risk to 75% and 64%. All new systems combined with a load limiter of 4.4 kN reduce the AIS2 injury risk below 50% (red and 2nd orange bar). However, the further increased load limiter level of 6.65 kN with the Oblique PAB gave an AIS2 injury risk of 90% (2nd blue bar).

The largest reduction was obtained when using the SCRS, with an AIS2 risk of 14% (2nd yellow bar) and when using the oblique PAB combined with reversed belt and 4.4 kN load limiter (2nd orange bar), reducing the AIS2 risk to 16%.

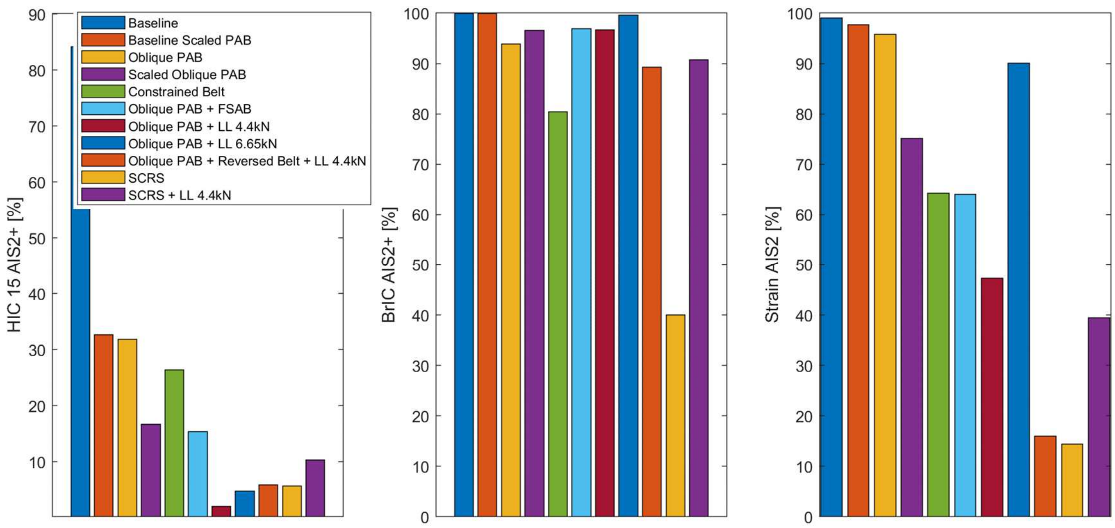


Figure 20. AIS2+ Injury risk calculated using HIC15, BrIC and strain in the brain (AIS2) comparing all evaluated saftey systems.

The risk for long-term injuries based on the 1% permanent medical impairment level (RLTC1) was also calculated for all systems for the head and showed that the highest level of RLTC1 was about 15% (Figure 21). The reason for the low levels is that the risk for permanent medical impairment (RMPI) of 1% or more for an AIS2 head injury is only 15%. Since the method only scales the AIS2 risk with 0.15, the same relationship between the levels is seen for RLTC1 and AIS2, but with a changed scale

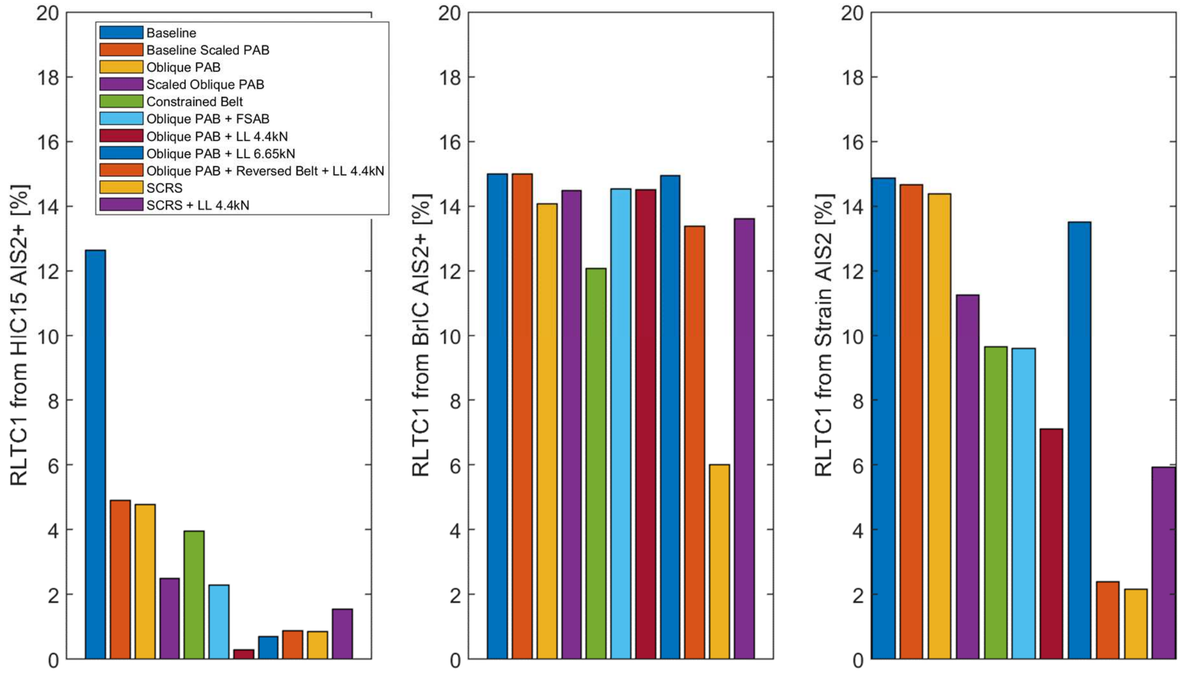


Figure 21. RLTC1 calculated using HIC15, BrIC and strain in the brain comparing all evaluated saftey systems.

Based on all the results from this study, it was found that a robust safety system for the oblique load case tested need to limit the lateral motion of the entire upper body of the passenger in order to avoid instrument panel head contact. In addition, head rotational velocity from interaction with a frontal PAB have to be limited.

A concept that could potentially reduce both head and chest injury risk is a large airbag that couples early with both head and torso and is angled in the direction of the pulse to get support from the IP. This would also reduce rib fracture risk induced by torso interaction with the central console.

5.2.4.2 Inflatable Curtain for Side Impact Head Injury Risk Evaluation

With the aim to predict long-term consequences for some head injury risk criteria, a parameter study was performed focusing on head interaction with an inflatable curtain (IC). Inspired by a real-world side impact crash, a simulation series was set up and parameters such as occupant sitting posture, crash pulse severity and impact angle to the car were varied.

The SAFER HBM v9.0.1 was positioned in a detailed vehicle interior model, similar to the real-world side impacted car, with state-of the art protection systems, including seat belt, torso side airbag (SIPSbag) and IC (Figure 22). The two driver postures; upright posture (nominal) and torso forward leaning (forward 10°) are shown in Figure 22. Three crash pulse acceleration levels were selected, all relevant for IC activation; low (14.2g), mid (18.9g) and high (20.2g). The impacts were applied in the perpendicular lateral direction (0°) as well as at ± 10° angles horizontally.

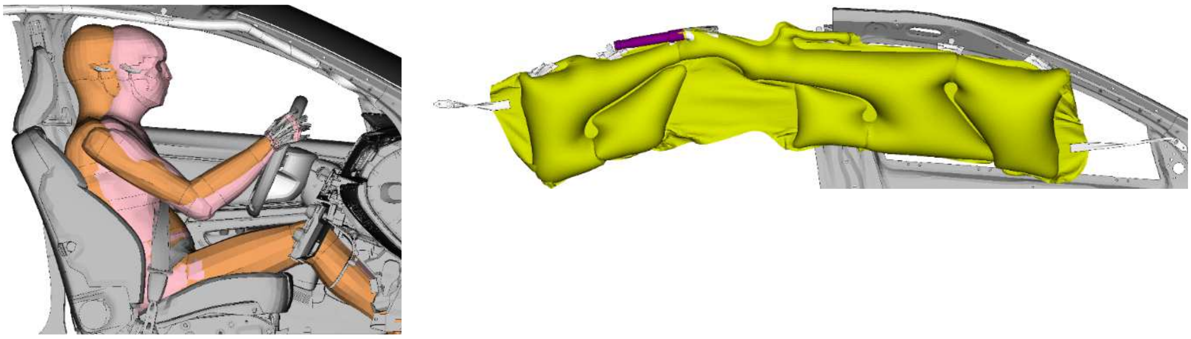


Figure 22. The simulation environment with the SAFER HBM v9.0.1 in the two sitting postures used (left) and the IC, when inflated (right).

Using strain-based criterion, the prediction of risk for concussion ranged from about 10% to about 40% in nominal posture and lateral impact, for the low to high crash pulses. The change of impact angle or the torso posture resulted increase of AIS2 injury risk (using strain) from about 10% to 20-25%. Similar trend was seen when using BrIC, however the injury risk increase was even more dramatic. Kinematics showed that the head was sensitive to the surface of the IC at point of impact, influencing the z-rotation.

The corresponding RLTC1 was calculated for HIC, BrIC, and the strain-based criterion for all the 18 combinations (Figure 23). It can be observed that RLTC1 levels are relatively low with a maximum of 13% for strain-based criterion.

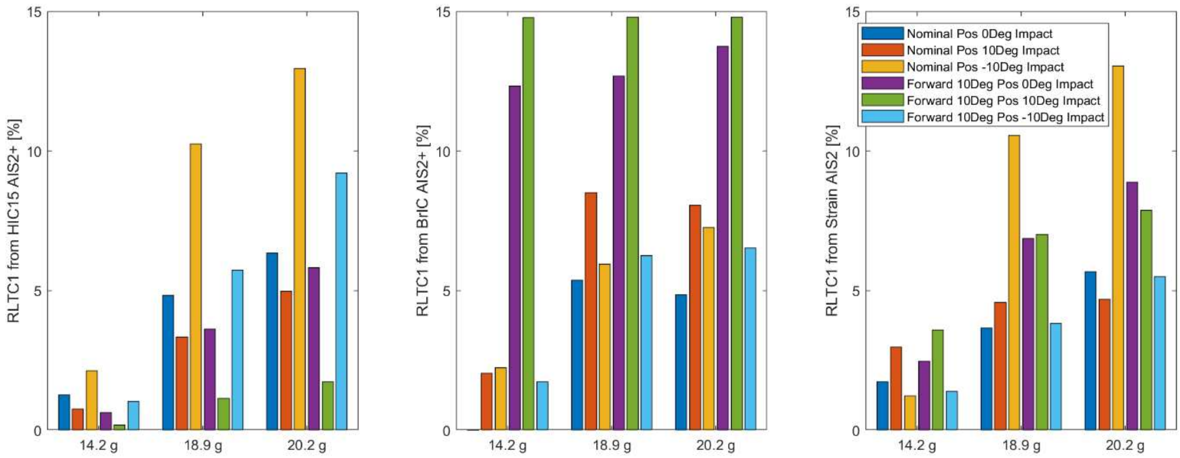


Figure 23. Head injury RLTC1 based on HIC, BrIC and strain for the 18 simulations varying sitting posture (nominal/forward), crash pulse (14.2g, 18.9g, 20.2g) and impact angle (0°, 10°, -10°).

It was found that the design of the IC, combined with the constant internal pressure, can result in a large variation in injury outcome, depending on the head impact point on the IC. It was further observed that the highest risk was due to bottoming out the IC towards the B-Pillar. Similar trends were observed in a few configurations between strain and BrIC, but there were also many differences. The strain-based criterion was assumed as the most trustworthy and used to draw the main conclusions.

5.2.4.3 Bicyclists Protection Airbag for Head Injury Risk Reduction

A bicycle protection airbag system (BPA) was evaluated by impacting a vehicle FE-model with the SAFER HBM v9.0.1 positioned on a bicycle FE-model (Figure 24) (Pipkorn et al., 2020). The head injury prediction was compared between having no head protection, a helmet (developed and validated by Fahlstedt et al., (2016)) and the helmet combined with the BPA (Figure 24). The bicycle and human body model were impacted by a Honda Accord MY 2013 (Singh et al., 2016). The windscreen of the original Honda Accord model was replaced by a previously developed and validated detailed model of the windscreen that included a failure model (Alvarez and Kleiven, 2016).

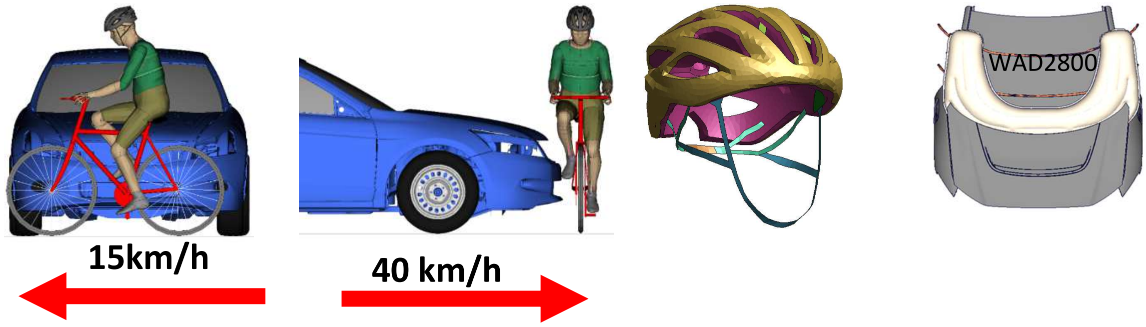


Figure 24 SAFER HBM on bicycle, helmet and bicyclist airbag

Stationary as well as moving bicyclist were evaluated. The countermeasures that were evaluated were helmet and bicyclist protection airbag (BPA).

For the stationary bicyclist, the head impacted close to the center of the windscreen while for the bicyclist with 15km/h velocity the head impacted the A-Pillar of the vehicle. The windshield was fractured in all impact configurations except for the moving bicyclist impacted by the vehicle with BPA.

For the stationary bicyclist, HIC was significantly reduced for both protective systems (Figure 25), while BrIC showed an increase using the BPA. The strain in the brain did however show a reduction in injury risk with the BPA, although relatively small (3%). The AIS2 injury risk based on strain was in general small for the stationary bicycle scenario (19-28%).

In the scenario with the bicycle travelling at 15km/h, significantly higher injury risks were seen with no protection system. Although a reduction was seen in HIC for both helmet and helmet and BPA, the AIS2 injury risk based on HIC was 100% for both no protection and helmet. The BPA showed a significant reduction in HIC. The injury risk based on both BrIC and strain was reduced for both protection system, although BrIC still showed injury risks above 80%. The AIS2 injury risk based on strain was however reduced from 97% to 52% and 31% for the helmet and helmet and BPA respectively.

For the windscreen impact, the windscreen fractured and absorbed part of the impact energy for the bicyclist with and without helmet. For the A-Pillar impact all energy was absorbed only by the helmet and helmet and BPA.

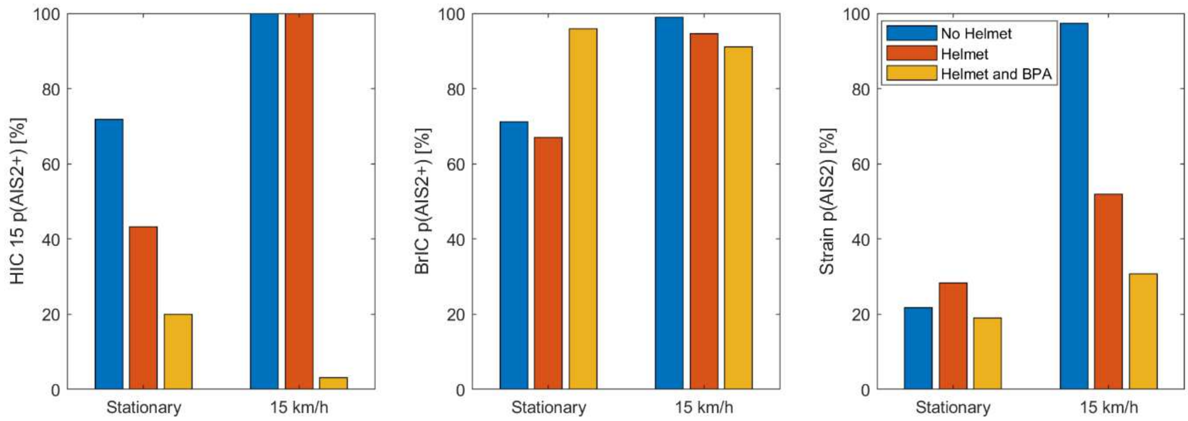


Figure 25. AIS2 risk based on; HIC15, BrIC and Strain in the brain for stationery and bicyclists travelling at 15km/h with No helmet, helmet, and helmet plus BPA.

The results indicate that the helmet show a more expected result in reducing the injury risk, with all used injury criteria, in the impacts towards the A-pillar. Although the helmet has a clear reduction of injury risk for the impact against the A-pillar, it could be considered still relatively high, with a 52% risk of AIS2, and hence a benefit of including the BPA, reducing the risk down to 31%. On the other hand, the impact towards the windscreen seems to generate a loading condition not fully considered in the development of helmets.

It was found that the BPA was an effective countermeasure in reducing head injury risk for bicyclists wearing a helmet. It can be assumed that the BPA is an even more effective countermeasure for bicyclists not wearing a helmet impacted by a vehicle. Future evaluation will be carried out evaluating the BPA benefits for bicyclists not wearing a helmet when impacted by a passenger vehicle.

5.2.4.4 Helmet Evaluation in Hood, A-pillar and Asphalt Impacts

The objectives of this study were to develop a helmet-to-vehicle impact test method and to evaluate the potential injury reducing effects of some current bicycle helmets, when impacting the vehicle as well as asphalt.

The helmets were mounted on a headform (usually used for pedestrian-to-vehicle impact tests) and launched at 36km/h (9.7m/s). The vehicle impact points included the hood and the A-pillar. The impact to the A-pillar was carried out with 2 different impact angles; 20° and 65° (Figure 26). In addition, impacts to asphalt were carried out for reference.

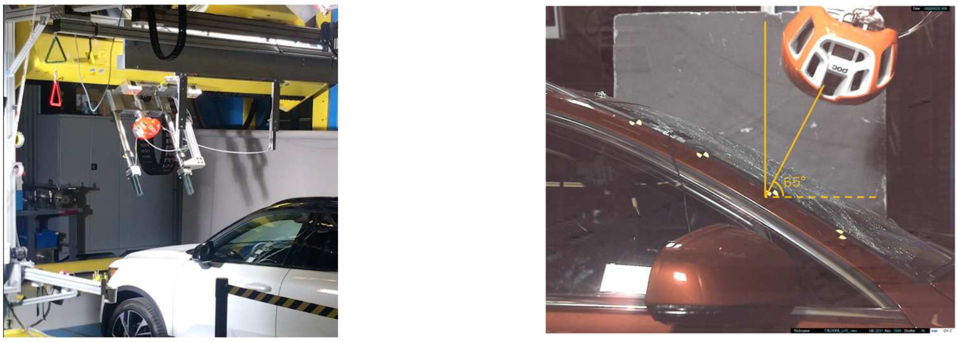


Figure 26. Helmet-to-vehicle impact test method; overview (left), and A-pillar impact angle at 65° (right)

Three different POC helmets were included; two ‘In mold’ helmets, with and without the rotational severity system, SPIN) and one ‘Hard-shell’ helmet, as well as without helmet (Figure *27*).

‘In mold’	In-mold with SPIN	In-mold without SPIN	‘Hard shell’

Figure 27 The POC helmet types tested.

The tests provided insights into performance of the different helmets, the differences of the impact configurations as well as learnings about the novel helmet-to-vehicle impact test methods. Predictions of injury reductions, measured through HIC15 and BrIC, were seen for all the helmets as compared to no helmet. The ‘In-mold’ helmets reduced the injury measures more than the ‘Hard-shell’ helmet, however, the advantages of ‘Hard shell’ helmets being more resistant to penetration and less sensitive to damage during normal use is not evaluated in the test.

For the hood impacts, the reduction in BrIC was greater than for HIC15. While for the A-Pillar and asphalt impacts the reductions in HIC15 were significantly higher than BrIC. The highest injury risks were predicted when impacting the asphalt and the A-pillar.

This novel test method was based on pedestrian test method. The learnings when applying the pedestrian test rig for helmet testing included modifications to the rig. In addition, the impact angle and assuring the targeted top-of-helmet impact point were difficult to control and needs further trimming to ensure repeatability. The use of the spherical pedestrian headform, instead of the Hybrid III head as usually used in helmet testing with oblique impacts, introduced challenges of getting a tight fit of the helmet. This caused undesired translation and rotation of the headform within the helmet. The rubber covering the headform contributed to reduce this rotation. Another issue was that the center of gravity and the mass moment of inertia are different between the spherical headform and the HIII head, whereby the results from the tests with the two different headforms cannot be compared.

Another difference to standard helmet testing was the impact speed. The impact speed of 9.8m/s is higher than used in other helmet testing, ranging from 5.4 to 6.2 m/s. However, the impact speed chosen in this test series is lower than the impact speed of 11 m/s as used by Euro NCAP in their pedestrian impact tests.

5.3 Addressing Upper Extremity Injuries

Upper extremity injuries were addressed in several different studies, encompassing investigation of impact mechanisms in frontal impact crashes, development of measurement tools and injury risk functions for assessment of injury risks, including the estimation of long-term consequences. The tools included a novel instrumented physical crash test dummy forearm, applied for impacts to instrument panel environments, and its virtual FE counterpart, as well as a novel arm model for the SAFER HBM. Using the novel method for injury risk estimations, an instrument panel airbag concept was developed and evaluated.

5.3.1 Investigating Impact Mechanisms in Frontal Impacts

A parameter study was carried out with the aim to investigate hand impact areas in frontal impacts, when varying hand on steering wheel positions. The set-up was based on two high-severity real-world crashes, involving drivers sustaining distal radius fractures. The two real-world cases are further described in Björklund et al. (2020). The SAFER HBM v9.0.1 was used and positioned in a Volvo driver environment, similar to the real-world cases. The hands were positioned on the steering wheel at five different locations, as shown in (Figure 28), in addition to positioned on the lap. This was done by rotating the joints of the shoulders and elbows, while holding the rest of the HBM fixed.

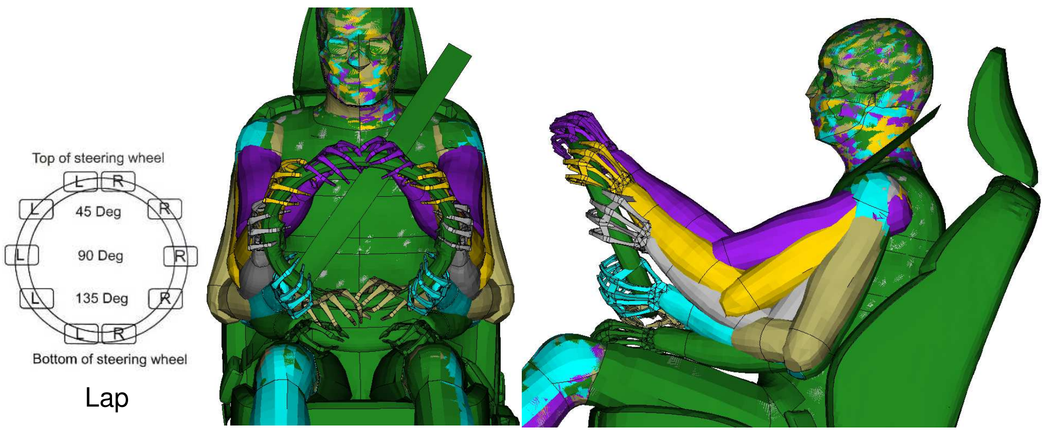


Figure 28. The SAFER HBM v9.0.1 in the variation of initial hand positions parameter study

When exposed to the two frontal impacts, as of the real-world cases, a large variation of impact points was obtained in the variation of initial hand positions. The left-hand impact points ranged from under the steering wheel, the door handle and frame on the left side, as illustrated in Figure 29. The right-hand impacts ranged from close to the steering wheel, to the top of the center console with the arm stretched out. In addition, a variation in impact mechanisms was observed; some initial arm positions led to relatively pure axial loading, whereas other resulted in more complex impact loadings, combining axial and bending.

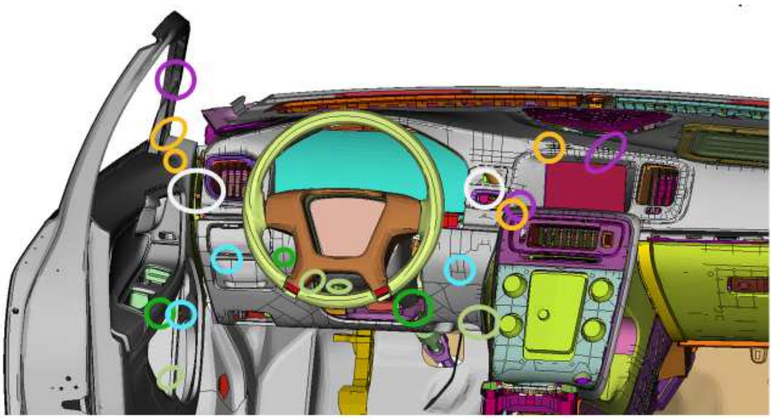


Figure 29. Upper extremity impact points in the variation of initial hand positions parameter study.

The results also showed a relatively large variation in arm impact velocity, ranging from 4 to 15m/s and 12 to 14m/s, respectively. The crash pulse with the largest lateral component resulted in the largest spread. The highest hand impact speed of 15m/s occurred when the hands were placed on the top of the steering wheel. For the other crash pulse, the highest impact speed of 14m/s occurred when the arm was positioned at 135°.

5.3.2 New Tools and Methods

5.3.2.1 A Novel Instrumented Forearm

With the aim to develop a method that can be used to assess hand and forearm injuries in vehicle safety testing, a novel instrumented forearm was developed within the project and published, see Björklund et al. (2020). A summary is provided in this report.

A Hybrid III forearm was modified to measure moment in the wrist, along with acceleration, force and moment transducers in the middle section of the forearm and was found capable of capturing possible hand and forearm injuries caused by hand impacts to vehicle interiors in frontal impacts.(Figure 30 and Figure 31). It is launched by an ejector as a free moving object.

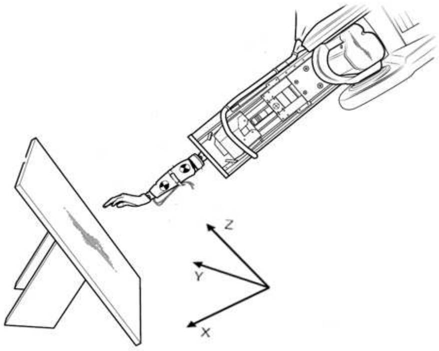


Figure 30. The hand and arm impact test method.

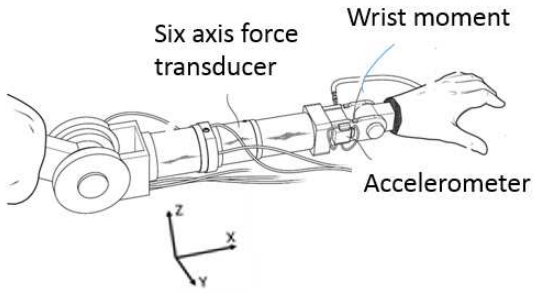


Figure 31. The novel instrumented forearm.

A parameter study towards a generic test board was performed, in addition to a test series towards a vehicle instrument panel (Björklund et al. 2020). The test method was shown repeatable, as well as sensitive to distinguish differences between the configurations. The responses were shown sensible to the irregular surface of the impacting structure, as well as to cracks and other structural damage lowering the loads through the forearm. The variations in results when slightly varying the test set-up supported the need of an easy-to-use component test method for vehicle development. The method, including the novel instrumented forearm, was found efficient to help provide input to vehicle design.

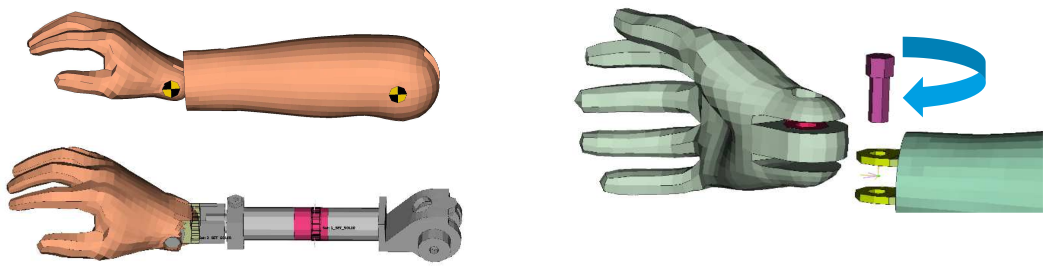


Figure 32. The CAE HIII forearm model with force and moment transducer (left lower image) and the wrist joint (right image).

In addition to the physical forearm, a CAE version was developed based on the "HUMANETICS_H3-50TH_V8.0.1_S3" model (Figure 32). The CAE HIII forearm was compared to the physical impact tests against a generic test board. The CAE HIII forearm was positioned with the wrist in 30° dorsi-flexion to correspond to the test set-up and with main axis of the forearm at an angle of 25° from the perpendicular position against a rigid plate and impacted using initial velocity of 8.4 m/s in the direction of the main axis of the forearm. The HIII forearm impact simulation was compared to three physical test repetitions and the frictional moment in the wrist and the coefficient of friction between the CAE HIII arm and plate were iterated to find a best fit as these parameters were hard to determine. It was found that the forces and acceleration levels corresponded well, but the moments in the arm and wrist were about 4-5 times lower in the simulation as compared to the tests (Figure 33).

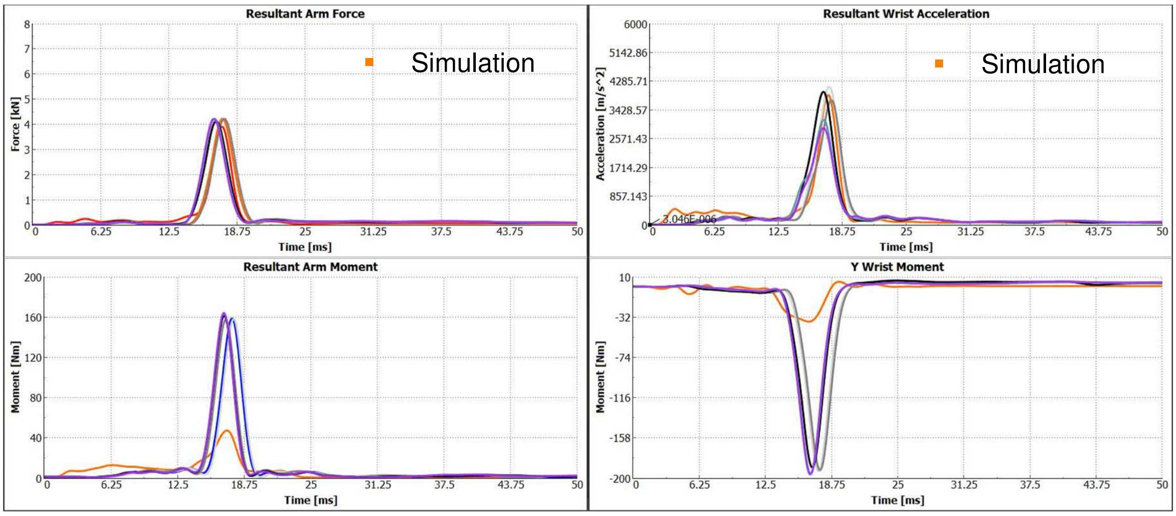


Figure 33. Corresponding physical forearm test and CAE forearm simulation. Forearm forces and moments (left plots), acceleration (top right plot) and wrist moment (lower right plot).

5.3.2.2 Development and Validation of Elbow joint and Forearm for the SAFER HBM

As part of the project, improvements of the elbow and the forearm of the SAFER HBM v9.0.1 were made through two master thesis projects. Sverrisdóttir (2019) improved the elbow joint. The main updates included the addition of the posterior aspect of the radial collateral ligament (RCL) and ulnar collateral ligament (UCL) with beam elements and stabilization of shell element ligaments. Bayat and Pongpairote (2020) refined the forearm model to improve the injury predictive capability in impact loading. Based on medical images, a CAD 3D surface model (STL) of the forearm was made to be used as source for generating a new FE-mesh with smaller elements and improved geometrical representation (Figure 34). The forearm and updated elbow joint models were also included in the SAFER HBM v10 update, together with a skin and flesh model surrounding it.

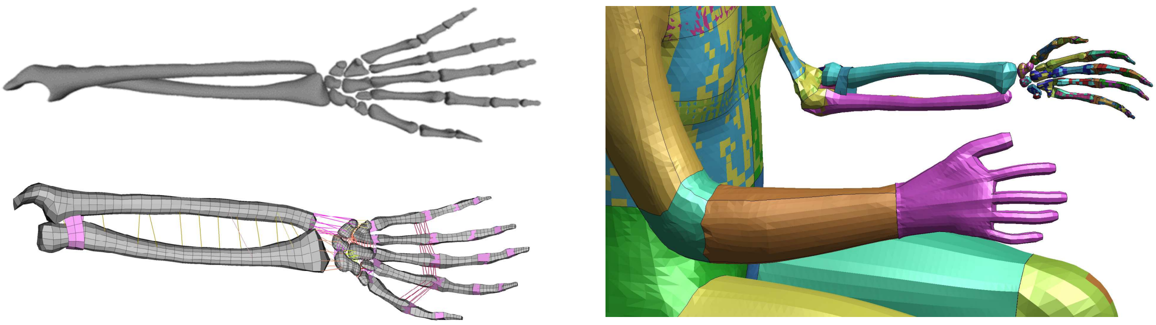


Figure 34. STL surface generated in master thesis project (Bayat and Pongpairote, 2020) to the left and as integrated model in the SAFER HBM v10.0.1 to the right.

The updated forearm model of the SAFER HBM v10.0.1 was validated by means of published PMHS experiments including 15 axial forearm impact tests with the hand in dorsi-flexion position (Forman et al. 2014). The test set-up is shown in Figure 37 and Figure 38. The forearm model was positioned with the hand in 90° dorsi-flexion and the ulna and radius bones constrained according to "fixed" or "free" elbow boundary conditions, in line with the PMHS test set-up (Figure 35**Error! Reference source not found.**). Measured displacement or accelerations of the contact plate were applied to the rigid plate in contact with the hand for the "free elbow" tests, the simulations were compared to the elbow reaction force and displacement, and for the "fixed elbow" tests to the elbow force.

The correlation was considered as “good biofidelity” according to CORA (Gehere et al., 2009). A CORA score of 0.7 was obtained when using the measured displacement as impact plate boundary condition, and CORA score of 0.76 when using the tests measured accelerations.

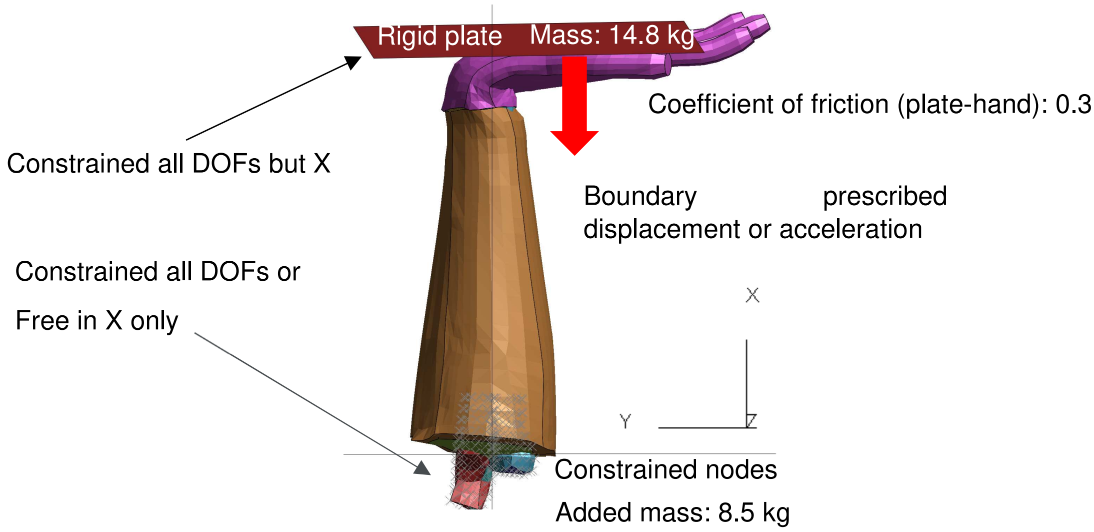


Figure 35. Simulation set-up of the SAFER HBM v10.0.1 forearm replicating the Forman et al. (2014) experiments.

5.3.3 Upper Extremity Injury Risk Functions

A strategy to develop injury risk functions for the novel instrumented forearm and the SAFER HBM forearm model was developed, see Figure 36. It includes two main paths. One path is based on recreating published human subject tests (exemplified by the PMHS tests by Duma et al 2003 and Forman et al 2014), to derive at injury risk functions. The other path is based on recreating real world crashes with detailed information on injury mechanisms and the use of whole-body HBM, as well as sub-system reconstructions using the forearm model only or the novel instrumented forearm impactor. Transfer functions are needed to adapt to the tool and method used. This can be done either as part of the PMHS test path only, or by combining the two paths. The two paths provide different possibilities and complement each other. The PMHS test recreation provided more detailed information and thereby the most direct input to injury risk functions, whereby the real-world crash recreations provide the overall context and input to relevance and feasibility.

Within this project we have adapted parts of the strategy and developed first-generation version of injury risk functions for assessing injuries to the forearm and wrist, for the novel instrumented arm and the updated HBM forearm model.

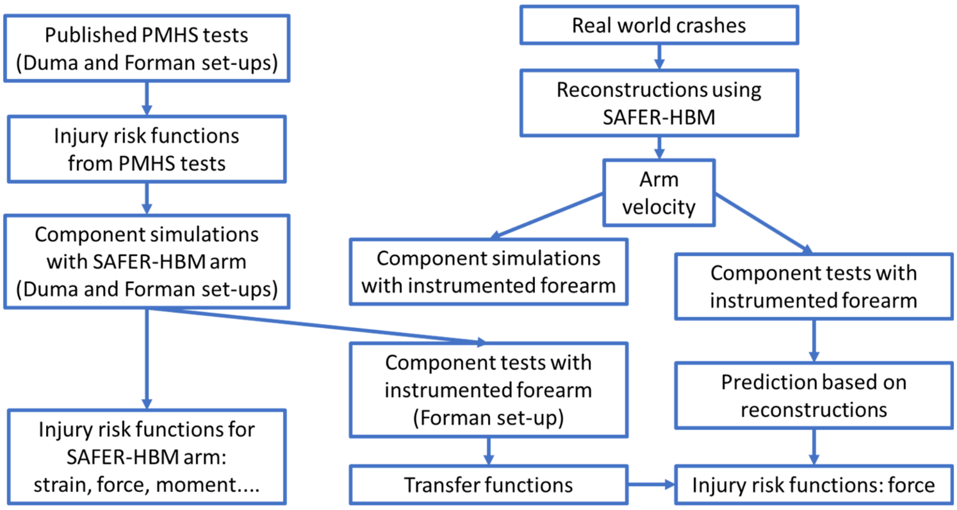
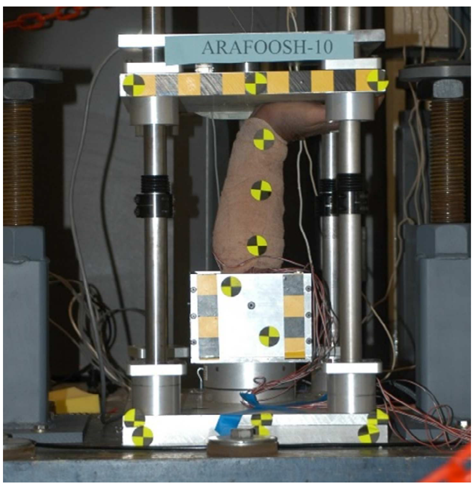


Figure 36 Strategy for injury risk function development

5.3.3.1 Injury Risk Functions for the Novel Instrumented Forearm

The 15 forearm impact tests carried out by Forman et al. (2014) were used with the overall aim to develop an injury risk function to predict forearm injuries when using the novel instrumented forearm. Work is ongoing to analyze the data and preparing a publication together with Jason Forman, University of Virginia, who assisted in executing the tests. In this report some insights are provided on the first steps; establishing a transfer function to be used when applying injury risk functions for the human subjects to be used together with the novel instrumented arm.

The novel instrumented forearm was placed in the same test rig as the human subject tests (Figure 37), whereby identical boundary conditions were achieved. The test set-up is also seen in Figure 38. Foreman et al. (2014) used bone-implanted strain gauges to detect the time of fracture. The sensors of the novel instrumented forearm recorded, in addition to the forces applied by the rig plates at the elbow and wrist. In total 11 tests with the novel instrumented arm were carried out, varying impact velocity from 1.74m/s to 5.33m/s. The forces applied at the elbow and wrist ranged from 1893N to 8766N at the wrist and 1585N to 8600N at the elbow.



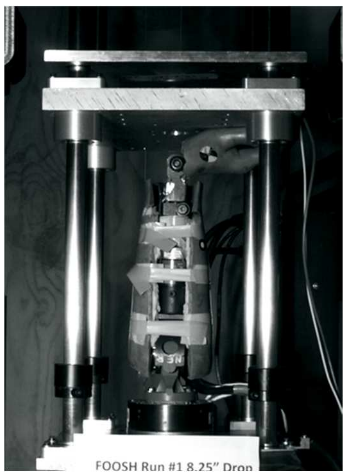


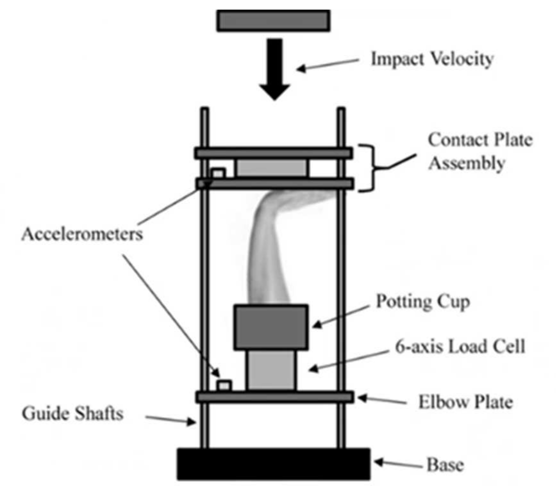
Figure 37. The test set-up of Forman et al. (2014); a human subject test in the reference (left) and the project's novel instrumented forearm (right).

Forman et al. (2014) provides injury risk functions, developed using parametric survival analysis with a cumulative Weibull distribution, for the reaction forces at the test rig plates. These can be used for the novel instrumented forearm if applying a transfer function developed under same impact conditions. The preliminary analysis shows that there is linear relationship between the force recorded at the elbow test rig plate and the axial force in the force transducer in the middle section of the forearm, with a coefficient of 0.88. Using this coefficient to shift the corresponding injury risk function in Forman et al. (2014) can serve as an injury risk function for the axial forearm loads when using the novel instrumented forearm.

5.3.3.2 Injury Risk Functions for the SAFER HBM Forearm

Two different PMHS test set-ups were used in the development of injury risk functions for the SAFER HBM forearm. The same test set-up as used for the validation of the SAFER HBM model and the development of the injury risk functions for the novel instrumented forearm (Forman et al., 2014), in addition to tests published by Duma et al. (2003). Both set-ups involved direct impacts to the hand in dorsi-flexion position, while having different degree of support at the end of the elbow (Figure 38). The set-up by Forman et al. (2014) is also shown in Figure 37.

Forman et al. (2014) suggested that an axial reaction force of 4.3kN in the elbow corresponds to a 50% fracture risk of the forearm. Duma et al. performed 17 tests with impact to the palm identifying forearm fractures, with axial reaction forces ranging from 1.7kN to 4.7kN with the arm free hanging.



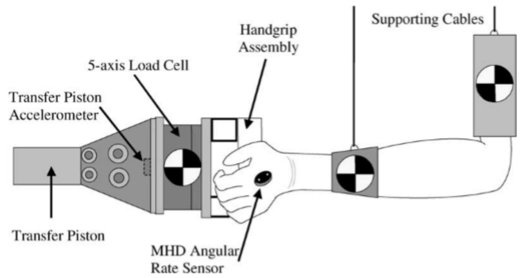


Figure 38. Experimental PMHS set-ups used for the injury risk function development for the forearm model; Forman et al. (2014) (left) and Duma et al. (2003) (right).

Simulations were performed with the HIII arm and the arm of the SAFER HBM v9.0.1 recreating the test set-up by Duma et al. (2003). A rigid plate was positioned against the palm of the models, extended to about 90°, as shown in Figure 39.

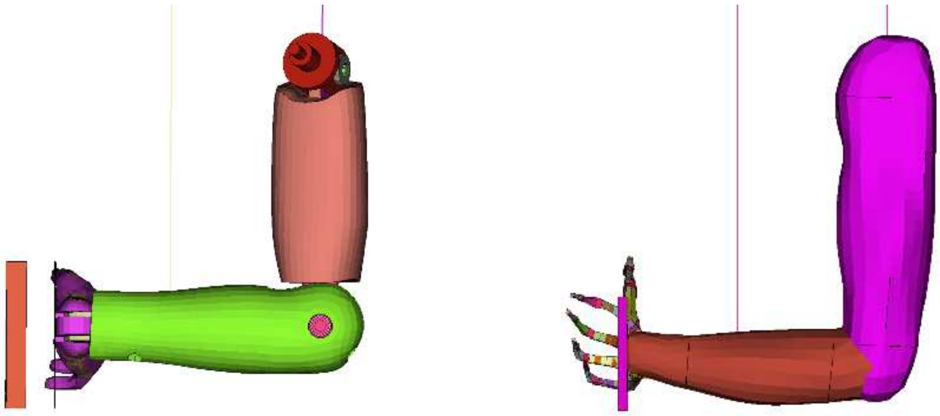


Figure 39. Simulation set-up of the experiment in Duma et al. (2003), using an impactor plate and beams representing supporting cables; HIII arm(left) and SAFER HBM v9.0.1 (right).

A preliminary injury risk function example for a strain-based criterion for the SAFER HBM v9.0.1 is shown in Figure 40. It is created using the simulation data and the published human subject test data in Duma et al. (2003), using simple logistic regression. Due to some missing data in the publication, some estimations were needed. A drawback with the logistic regression used is that the intercept at zero metric value does not necessarily give a zero risk of injury, as is the case for the risk functions generated.

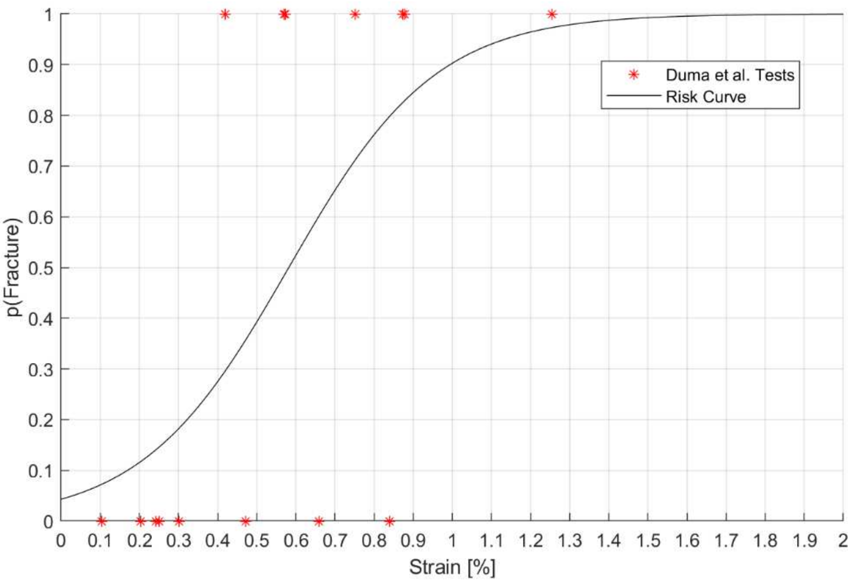


Figure 40. Injury risk function for the SAFER HBM v9.0.1, based on strain-based criterion from the simulations recreating the tests in Duma et al. (2003).

Later in the project, the SAFER HBM v10.0.1 forearm was used to recreate both test set-ups. The injury risk functions were then developed based on survival function with Weibull distribution.

Injury risk functions for the relative deformation and axial force are presented in Figure 41, comparing the results from the PMHS tests and the corresponding simulations. It can be observed that the injury risk functions based on relative displacement was similar in shape as the functions developed by Forman et al. (2014) but with slightly lower injury risk (shifted to the right). The injury risk functions based on the Forman experiment and cross-section force showed similar force level at 50% risk, but with a less steep shape.

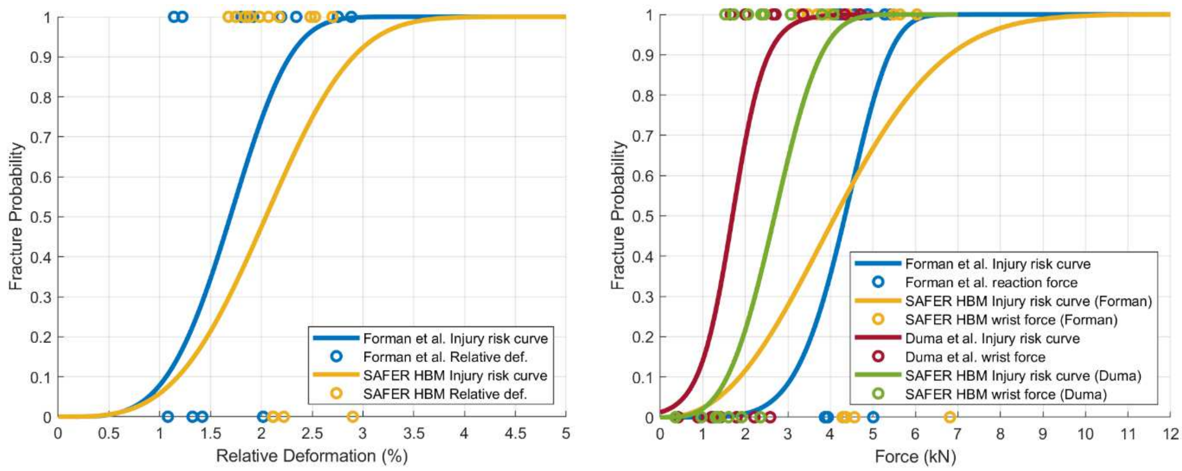


Figure 41. Injury risk functions for relative deformation (left) and axial force (right); comparing simulation results using the SAFER HBM v10.0.1 and the results from the corresponding PMHS tests by Forman et al. (2014) (left and right plots) and Duma et al. (2003) (right plot only)

5.3.4 Countermeasure Development and Evaluation

To understand potential countermeasures for enhanced protection of upper extremity injuries, specifically focusing hand and forearm in frontal impacts, several studies were performed. A parameter study towards a generic test board provided some insight into influence of a selection of factors to potential mitigate injuries to the forearm. The parameters included stiffness, damping, friction and impact angles, in addition to varying hand in dorsi-flexion and palmar-flexion at impact. This study is further described in Björklund et al. (2020).

Following this parameter study, some airbag concepts addressing hand impacts in frontal impacts were developed and evaluated using the tools and methods developed within the project.

5.3.4.1 Airbag Concepts – The Rationale and Design

Two airbag concepts were designed to reduce the forces in the forearm in an impact (Figure 42). One applies a force to the upper arms, when holding the steering wheel, in order to reduce the velocity of the hand and forearm towards the instrument panel. To evaluate a practical implementation of this, a driver airbag with a wide base design that couples the upper arms was designed, called the ‘steering wheel arm airbag concept’. The other concept was to introduce an airbag that covers the instrument panel (IP), to help reduce the impact forces at impact, called the ‘instrument panel airbag concept’.

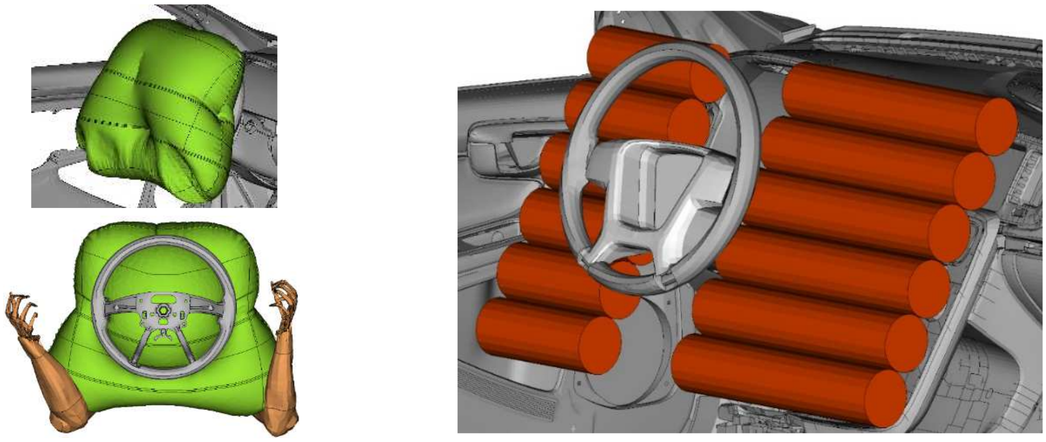


Figure 42. Two airbag concepts targeting upper extremity protection in frontal impacts; the ‘steering wheel arm airbag concept’ (left) and the ‘instrument panel airbag concept’ (right).

A simulation study using the SAFER HBM v9.0.1 (with the original upper extremities), was performed based on the set-up of the parameter study on hand position variants (Chapter 5.3.1). The ‘instrument panel airbag concept’ was tested in several volume alternatives, together with the ‘steering wheel arm airbag concept’ and the baseline system of IP only. Figure 43 displays a comparison of responses from one of the impact configurations. It shows that both concepts where able to reduce the force and moment in the forearm. However, due to the large variation of impact points and orientations observed in the study on hand position variants, it was concluded that the ‘instrument panel airbag concept’ likely was a more robust solution. Additionally, the study showed that the instrument panel airbag concept was able to reduce the force and moment to a higher extent than the driver airbag concept.

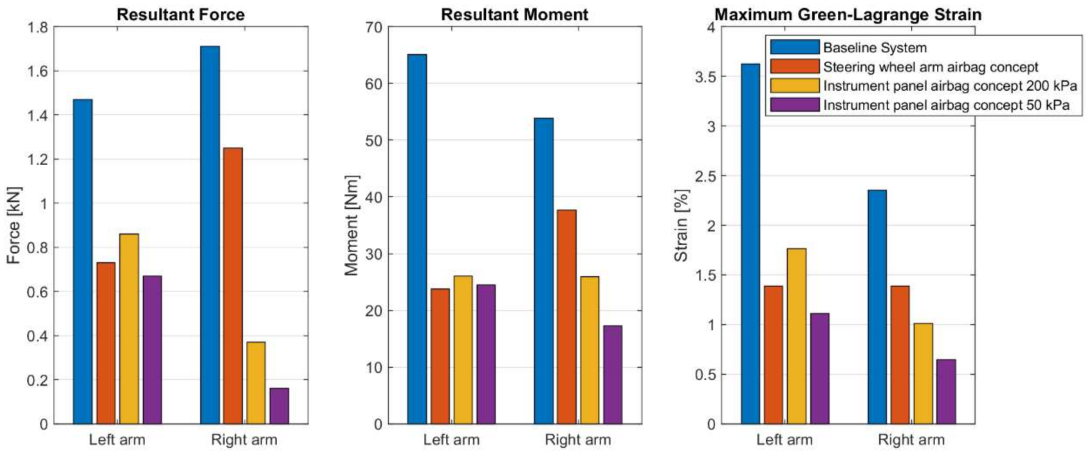


Figure 43. Force and moment measured in the distal part of the SAFER HBM v9.0.1, comparing the two airbag concepts with a baseline with IP only.

A prototype of the ‘instrument panel airbag concept’ was built and evaluated using the novel instrumented forearm and the generic test board, in addition to an IP (Figure 44). The tests showed that airbag pressure and cell distribution design influence the responses of the novel instrumented forearm. A reasonable damping effect was found for the airbag pressure of 70 kPa in the impact speed of 9.85 m/s. The moments were influenced by how the hand impacted the cells, indicating that cell distribution design is essential for robust performance.

Using the set-up of one of the real-world cases as presented in Björklund et al. (2020) and Chapter 5.3.1, and pressurizing the airbag with 70 kPa, a force reduction of 65 % (as compared to IP only impact) was achieved when impacting the top of an airbag cell. When moving the cells so the impact occurred between the cells, a lower relative force reduction (10%) was obtained, however a higher reduction of moment (42%), as compared to IP only impact.

Figure 44. Test set-ups using the novel instrumented forearm to impact the prototype 'instrument panel airbag concept'; generic test board (left) and vehicle IP (right).

5.3.4.2 Airbag Concept - "Proof of Concept"

Results from a simulation series with the 'instrument panel airbag concept' was used to assess the method to predict risk for long-term consequences, developed with the project, as well as to serve as a 'proof' of the concept, as such. Comparing simulation results with and without the concept, using the HBM, the methods on predicting risk for long-term consequences (Chapter 5.1) was applied as a means to quantify the injury reducing benefits of the 'IP hand airbag concept'.

The test series comprises two simulations with the updated SAFER HBM v10. The set-up is of one of the real-world cases (as presented in Björklund et al. (2020) and Chapter 5.3.1). The reference is the simulation targeting a reconstruction of the real-world crash. The other includes an 'instrument panel airbag concept' within the area of hand impact. Table 5 presents the maximum axial force extracted from a cross section in the distal part of the forearm, the corresponding fracture risk using the injury risk functions developed (Chapter 5.3.3.2, Forman data), in addition to the RLTC1. The RLTC1 (1% risk for long-term consequences) Risk for Long-term Consequence at 1% is the result of the method developed for prediction of injuries for long-term consequences as presented in Chapter 5.1. Although the risk of fracture and consequently the RLTC1 was low in both simulations, the 'instrument panel airbag concept' contributed with a reduction of about 10% in fracture risk and the RLTC1 reduced from 5-4% to almost zero (Table 5).

Table 5. SAFER HBM v10 maximum axial forearm force and the corresponding fracture risk and RLTC1+ calculation from the proof-of-concept simulation series for the 'instrument panel airbag concept'.

	Arm	Max Force [kN]	Fracture risk (AIS2) [%]	RLTC1+ [%]
Reference	Left	1.96	11.1	4
	Right	2.17	14.0	5
Instrument panel airbag concept	Left	0.78	1.3	0.5
	Right	0.27	0.1	0.04

Using the set-up of one of the real-world cases and the updated SAFER HBM v10 with the validated forearm, provide some real-world anchoring of the results. However, it is difficult to ensure that the injury mechanism of the distal tibia fracture in the frontal impact real-world case is captured correctly in the simulations, whereby it is difficult to put the results in relation to injury outcome. Nevertheless, the relative injury risk reduction estimations are likely relevant and can provide valuable input for countermeasure design guidance, addressing risk for long-term consequences.

6 Execution in line with Objectives

The objective of this project as it is described in the project application and in chapter 4 included studies on head and upper extremity injuries to enhance understanding and evaluation of injuries leading to long-term consequences with the purpose of developing tools and method for countermeasure development for enhance protection. Specifically, the project targeted understanding injuries leading to long-term consequences and to gain an insight on the relevance of the ability of various head injury criteria to predict those head injuries.

To address the objective how to predict and prevent head injuries numerous activities were carried out in the project. The KTH head model was further refined and validated. Risk functions were developed to predict concussion and skull fracture risk in a head impact. The potential injury reducing benefits for the head by bicycle helmets, passenger side airbags designed for oblique impact protection were evaluated.

Another objective was to evaluate if global injury criteria (BrIC, HIC and RIC) are relevant measures for head injury evaluation Injury risk predicted by global head injury criteria was compared head injury risk predicted with validated FE head models. Numerous loading scenarios were evaluated.

To enable predictions, evaluation and prevention of upper extremity injuries a novel lower arm impact mechanical and virtual (CAE) test method was developed. Risk functions for lower arm fracture risk prediction was developed. The tools and methods were used to evaluate the potential injury reducing benefits by upper extremity impact airbag concepts.

The human body models SAFER HBM was refined to improve the capability to predict concussion and further developed to enable upper extremity injury risk prediction. The modified SAFER HBM model was found to be an efficient tool for head and upper extremity impact evaluations.

The objective of addressing injuries leading to long-term consequences and how these injuries relate to exposure and type of situation was addressed in a literature study. The findings in the literature study were compiled in a report. In addition, as part of the initial objective, the project aimed to create and analyze unique data quantifying specific injury types being important to long-term consequences. The data collection could not be established in time, due to lack of resources at one of the industrial partners. Therefore, knowledge about crash configurations and other crash parameters related to long-term consequences could not be obtained through that data. The targeted report of injury types could not be made. As a result of that, the planned workshop in WP2 did not take place. This task was partly replaced by the creation of method to predict long-term consequences (chapter 5.1), which was not planned initially. In addition, as an extra contribution to the project, Helena Stigson from Folksam joined the project team, participating at the project meetings and contributing in the project discussions on long-term consequences.

7 Overall Results and Deliverables

The results of this project contribute to the reduction of injuries sustained by vehicle occupants and support the work towards the Vision Zero ambition of reducing fatalities and injuries in traffic by developing tools and methods that enable development of protection system to reduce head and upper extremity impact injury risk. Injuries to the head and upper extremities can lead to permanent impairment.

By development of world leading tools and methods such as the KTH head model that the Swedish vehicle industry can use to invent and develop new and improved protection systems the project has contributed to increasing the Swedish vehicle industry's competitiveness and to strengthen the Swedish traffic safety research edge.

With the goal to enable evaluation of head injury risk and to rank bicycle helmets various head global as well as tissue-based injury criteria were evaluated in order to define the criteria most relevant for injury prediction. Using the criteria to evaluate head injury risk bicycle helmets can be developed with the aim to reduce fatal and serious head injuries for vulnerable road users such as bicyclists. Reducing fatal and serious head injuries for vulnerable road users such as bicyclists is an important contribution to reduce the number of fatalities and injuries in traffic.

By comparing and evaluating various global as well as tissue level injury criteria using numerous tools such as mechanical HIII head in helmet testing, and eight different finite element head models an evaluation of the relevance of various global head injury criteria for prediction of head injury was carried out. The evaluation was carried out by comparing the prediction of the tissue level injury criteria from the finite element models with the global head injury criteria. Mixed agreement between the prediction from the head models and the various global head injury criteria. Best agreement was found between the tissue level injury criteria predicted by the majority of the head models and the global injury criteria DAMAGE

The KTH head model was modified by adding a spinal cord addressing the FFI goal of development of CAE and virtual tools. The modifications enable improved head injury prediction capabilities of the model. Risk function for predicting concussion (mild traumatic brain injury) was developed with the model. These risk functions were used by the industrial partners in the project to evaluate various restraint systems such as airbags and curtains. In addition, these risk functions will be used by the industrial partners to develop future products.

FE models of human forearm and elbow were developed and validated in the project. To enable assessment of forearm injuries, fracture risk functions were developed based on published human subject test data. The updated upper extremity and head models were integrated into the unique and world leading tool SAFER HBM. Injury risk functions to address the risk for forearm fractures were developed. The model was used to develop and evaluate some novel concepts as countermeasure for arm injuries for occupant protection in frontal impacts. The updated SAFER HBM model provides the Swedish industry with a global competitive edge.

In current crash test dummies there is no capability to assess the risk for upper extremity injuries. Therefore, a test method to assess upper extremity injury risk was developed in the project. The method comprises a novel instrumented forearm that is propelled against the surface to be evaluated. In addition, the arm can be mounted on a crash test dummy for evaluation of upper extremity risk in a crash or sled test.

The dominating bicyclist protection system is the helmet. In angled helmet impact rotational velocity of the head of a bicyclist are induced. Rotational velocity of the head increases the risk for the bicyclist to sustain a concussion. There are systems such as MIPS and SPIN that are designed to reduce the rotational velocity of the head of in an impact. It was found in testing and simulations with the KTH head model that those systems reduce the rotational velocity of the head in an impact and hence reduce the injury risk.

The project addresses UN agenda 2030 goal 3 – “Ensure healthy lives and promote well-being for all at all ages” by developing tools, such as the head model and the upper extremity test methods, enabling development of projection system for vulnerable road users”. The project results can be used to develop low cost bicyclist impact countermeasures for developing countries where helmet use is limited, and bicycling is very frequent.

Two workshops were organized. One about head injuries, with international participation, and one about upper extremities. In addition, the project results were used in discussion with Euro NCAP about head injury criteria for future rating methods and with Input to helmet standards for European standard committee CEN TC 158 and the corresponding Swedish committee SIS/TK 525. So far, the project has, resulted in one half time thesis (corresponding to a licentiate degree), six scientific publications and three master theses. The results have been disseminated at international conferences, such as IRCOBI and American Association for Automotive Medicine (AAAM). Additional publications are planned on studies within the project. One of these includes collaboration with Jason Forman at University of Virginia, on the establishment of injury risk functions for our novel instrumented forearm.

The project also addressed the FFI goal of fostering/promoting collaboration between industry including small and medium-sized (SME) companies and university, targeting both SME and cross-industrial cooperation. The project partners included 2 SME:s, POC and MIPS, university and 2 large companies

Volvo Cars and Autoliv. Publications from the project included papers with authors from an SME, a university and a large company.

The current project will contribute to several of the overall FFI objectives by:

- Enhancing research and innovation capacity in Sweden through innovative safety design, ensuring competitiveness and jobs.
- Fostering/promoting collaboration between industry including small and medium-sized (SME) companies (POC and MIPS) and university, targeting both SME and cross-industrial cooperation.
- Developing internationally interconnected and competitive research and innovation environments in Sweden

8 Dissemination and Publications

8.1 Knowledge and Result Dissemination

A spin off project between POC and Autoliv was initiated based on the results from this project. The scope of the project is to develop bicycle helmets to protect the head of a bicyclist.

A follow up project "In-Depth Accident Study for Improved Injury Assessment Tool and its Coupling with Driver Behaviors for Precise Injury Prevention" with the Swedish partners Autoliv, Volvo Cars, MIPS and KTH and the Chinese partners Research Institute of Highway Ministry of Transport (RIOH), Chang'an University (CAU), and industrial partner China Automotive Technology & Research Center (CATARC) financed by VINNOVA was started 200901.

The project has cooperated with University of Virginia (UVA) in the area of upper extremity injury evaluation. UVA has shared PMHS lower arm impact test data and carried out tests with the instrumented lower arm and provided the results. A joint publication based on the results from testing with the instrumented arm is planned.

How are the project results planned to be used and disseminated?		Comment
Increase knowledge in the field	x	Significant contribution, exemplified by the amount, variety and novelty of studies, and their publications and outreach by presentations.
Be passed on to other advanced technological development projects	x	Improved injury criteria implemented in the postprocessor for the SAFER HBM is used in other development projects. Autoliv and POC have started a joint helmet development project
Be passed on to product development projects	x	The improved injury criteria are used in product development projects at the project partners. The results from upper extremity and helmet testing have been used in in-house testing at the project partners
Introduced on the market		The project results are used in development of upper extremity countermeasure and helmet developments at the project partners
Used in investigations / regulatory / licensing / political decisions	x	The project results have influenced regulations, ISO-standards and rating tests (e.g. EuroNCAP)

Poster presentation at FFI's resultatkonferens" 27 September 2018, Göteborg
https://www.vinnova.se/kalenderhandelser/20182/09/resultatkonferens-for-ffi-trafiksakerhet--automatiserade-fordon/

8.2 Publications

This project has resulted in 3 master thesis reports from a total of 4 students at KTH and 1 peer review article, 4 conference publications scrutinized by referees and one “half time” presentation.

Half Time (Corresponding to Licentiate Presentation)

Pooya Sahandifar 7/5 2020, KTH – Half time (Corresponding to Licentiate) ” Aging effects on hip fracture and first insights into the safety of pavements”, Opponent/discussion leader: Professor Hanna Isaksson, Lund University

Master Thesis

Sverresdottir, K., (2019), “Improvement and Validation of THUMS Upper Extremity Model – Refinement of the Elbow Joint for Improved Biofidelity, Masters Thesis Royal Institute of Technology, Stockholm, Sweden

Valle Olivera, N., (2020), ”Modelling and Evaluation of a Finite Element Cervical Spine Cord for Injury Assessment”, Masters Thesis Royal Institute of Technology, Stockholm, Sweden

Bayat, M., Pongpairote, N., (2020), ”Arm Injury Prediction withTHUMS SAFER - Improvements of the THUMS SAFER upper extremity”, Masters Thesis Royal Institute of Technology, Stockholm, Sweden

Peer-review publications

Fahlstedt, M., Abayazid, F., Panzer, M.B., Trotta, A., Ghajari, M., Gilchrist, M.D., Ji, S., Kleiven, S., Li, X., Ní Annaidh, A., Halldin, P., (2021), Ranking and Rating Bicycle Helmet Safety Performance in Oblique Impacts Using Eight Different Brain Injury Models, Annals of Biomedical Engineering

Manuscript under Preparation for Peer-review Journals

Fahlstedt, M., Meng, S., Patton, DS, McIntosh, AS, Kleiven, S., Concussion Risk Functions from Helmeted and Unhelmeted Sports Accidents, Submitted to Annals of Biomedical Engineering

Fahlstedt, M., Alvarez, V., Halldin, P., Ji, S., Panzer, M.B., Pipkorn, B., Evaluating Correlation between Different Local and Global Brain Injury Metrics for Short and Long Duration Impact Pulses

Fahlstedt, M., Meng, S., Kleiven, S., AIS1 and AIS2 Risk Functions for Concussion – Kinematic-Based Predictors

Fahlstedt, M., Meng, S., Kleiven, S., AIS1 and AIS2 Risk Functions for Concussion – Model-Based Predictors

Pipkorn, B., Alvarez. V., Fahlstedt, M. Development of Upper Extremity Models for Radius and Ulna Fracture Prediction, Planning

Conference publications scrutinized by referees

Slagmaat, Mv., Muller, B., Pipkorn, B., Panzer, M. (2019), ”Suitability of Enhanced Head Injury Criteria for Vehicle Rating”, AAAM, Madrid, Spain

Fahlstedt, M., Halldin, P. (2019), “The Difference in Ranking of Bike Helmets When Using Different Finite Element Head Models”, IRCOBI, Florence, Italy

Björklund, M., Risberg, J., Laudon, O., Jakobsson L. (2020), “Development of a Hand and Arm Impact Test Method and a Study on Influencing Factors”, IRCOBI, München, Germany

Pipkorn, B., Alvarez, V., Fahlstedt, M., Lundin, L. (2020), ”Injuries and Countermeasures for a Bicylist Impacted by a Passenger Vehicle”, IRCOBI, München, Germany

Pressrelease

190603 “Volvo Cars and POC develop world-first car-bike helmet crash test”

9 Conclusions and Future Research

The project has taken steps towards addressing traffic related injuries that can result in various degrees of long-term consequences, specifically the project has addressed head injuries and upper extremity injuries which are among those most frequent. To assess those injuries the project has:

- Improvement of the KTH head model and developed risk functions to assess the risk for mild traumatic brain injuries and skull bone fractures in head impacts.
- Evaluated and ranked bicycle helmets
- Evaluated and compared global head injury criteria
- Evaluated head injury risk for airbags designed for reduction of head injury risk in oblique impacts
- Developed an upper extremity impact test method
- Developed and validated forearm and elbow finite element models
- Developed injury risk functions for assessment of forearm fractures
- Developed and evaluated airbag concepts for upper extremity impact injury mitigation

Injury risk functions for mild traumatic brain injuries were developed due to the fact that these injuries frequently are associated with long-term consequences. Future research will include development of head injury risk functions for other frequent head injury. Such development includes refinement and increased level of detail of the finite element head models to enable modelling of the injury mechanisms that are associated with the head injuries. To assess the injury risk associated with the head injuries mechanisms specific risk functions have to be developed. Example of such injuries are subdural hematoma, subarachnoid hematoma etc. In addition, for such developments, validation data is needed.

The project has taken the first step towards addressing upper extremity injuries which frequently result in various degrees of permanent disability. There is a need for methods and tools to address other upper extremity injuries such as injuries to the wrist, humerus and scapula that are frequently observed in vehicle crashes. Methods and tools are also needed in the industry to enable development of protection systems to mitigate injuries also to the other parts of the upper extremities. Knowledge about injury mechanisms, as well a tool to address the injuries are lacking. In addition, fundamental biomechanical data to be used for development and validation of tools to be used for evaluation of such injuries is lacking. Therefore, there is a need to continue carrying out research in the area of upper extremities.

10 Partners and Contact Persons

Autoliv:	Bengt Pipkorn (project leader) Victor Alvarez
Volvo Cars:	Lotta Jakobsson Magnus Björklund Magdalena Lindman
KTH:	Svein Kleiven Madelen Fahlstedt Pooya Sahandifar
POC:	Oscar Huss Johan Weman Philip Malmegård Magnus Gustavsson
MIPS:	Peter Halldin

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