EVALUATION OF THE HYBRID III 5TH FEMALE MODIFIED CHEST JACKET & SPINE BOX

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ABSTRACT

The SAE has coordinated development of a new chest jacket and spine box for the Hybrid III (HIII) Fifth Percentile Female Crash Test Dummy. The proposed modifications intend to correct dimensional inconsistencies in the chest jacket drawings, make the jackets in accordance with the new drawings and eliminate a potential source of mechanical noise in the data. NHTSA procured two new chest jackets, one from each supplier for evaluation. The following questions were investigated through series of inspection, certification, and out-of-position (OOP) and sled tests.

- Are the two new design chest jackets effectively the same shape, construction and performance?
- Do they both meet the drawing specification?
- Is the noise eliminated?

The study presents data collected on both Robert A. Denton (Denton) and First Technology Safety Systems (FTSS) produced dummies**.** The companies have since merged into Humanetics Innovative Solutions, Inc. The dimensional inspection data presented includes a comparison of the anthropomorphic characteristics to the design specifications. The performance of the dummy is evaluated through analysis of the three types of dynamic test data. This includes deflection, acceleration, loads and high speed video from certification tests, low risk deployment tests and sled tests. The analysis of injury values is also performed. The authors' hypothesis is that the new dummies all produce comparable dimensional data and test results. The actual variances are documented. Preliminary comparison showed dimensional compliance within 3 mm and good repeatability. Inspection reports provided dimensional data for both jackets along with laser scan results. Dynamic test data provided deflection, acceleration and load data from certification, OOP and sled testing. The data was analyzed using standard hypothesis test methods (student t-test) to accept or refute the hypothesis that the jackets are effectively the same. The test matrix was limited in sample size for both the OOP and sled tests. The use of a mandrel to assure that the jackets are dimensionally correct is a novel approach for improving quality.

INTRODUCTION

The HIII 5th percentile female is regulated by the Code of Federal Regulations (CFR), Title 49, Part 572 Subpart O. The Agency owns over 20 dummies mostly of the FTSS brand. There are noticeable differences in the location and shape of the breast of the chest flesh assembly between the FTSS and Denton brands that have been resolved by the new jackets. Since 2006 NHTSA has been providing both brands of dummies to its contracted test sites so that NHTSA's Federal Motor Vehicle Safety Standard (FMVSS) No. 208 tests, especially the low risk deployment tests can be conducted with the same dummy type (Denton or FTSS) 5th percentile female that was used by the vehicle manufacturer in its development tests (as requested by the Alliance of Automotive Manufacturers [1]). In addition, it has also been suggested that high frequency noise from metal-to-metal contact is entering the dummy chest accelerometers during low risk deployment tests may be related to the spine design [2].

The chest jackets from the suppliers, FTSS and Denton were not the same nor did they fully agree with the drawings where they were inconsistent. To resolve these issues the SAE has coordinated development of a new chest jacket, modified the spine box and created a mandrel and recommended drawing changes [3, 4] (Figures 1 to 3). At this time Humanetics manufacturers both jackets.

Figure 1. Harmonized chest jackets.

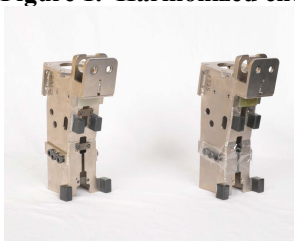

Figure 2. Harmonized spine boxes.

Figure 3. Harmonized chest jackets on harmonized mandrels.

The purpose of NHTSA's evaluation was to determine if the modified HIII 5th female dummies were acceptable for incorporation into Part 572. The dummies were evaluated through a series of inspection, certification, and low risk deployment and sled tests. Evaluations were performed of the repeatability and reproducibility (R&R), durability, measured injury assessment reference values, new parts, and drawings. Two laboratories (VRTC and TRC) were used to conduct R&R testing on the dummy certification. A 2006 Volkswagen Passat was used for the low risk deployment tests and Transportation Research Center HYGE sled was used to simulate a 2010 Ford Taurus NCAP crash pulse.

The SAE Dummy Test Equipment Committee has developed two "J" documents, recommended practices, to document the harmonized chest flesh (SAE J2921) and spine box changes (SAE J2915) [3, 4]. It should be noted that while SAE J2921 evaluates the response of the dummy to thoracic impact it does not address torso flexion response which is a Part 572 requirement. The committee has since decided to address the torso flexion response [5].

METHODS

The modified chest flesh assemblies and spine boxes were procured from both FTSS and Denton. The parts were installed on two dummies (144 and 416) from NHTSA's inventory that were prepared with parts entirely from those original equipment manufacturers. These modified dummies were compared with two 5th female dummies one from each manufacturer that were built in accordance to the CFR Part 572. Dummies 140 and 509 were prepared with parts entirely from those of the original equipment manufacturer and did not have the modified chest flesh assemblies or spine boxes. Denton dummies used in this study were serial nos. 140 and 144 and FTSS dummies were 416 and 509. In addition, chest assembly mandrels and a 3-D inspection file were also procured to be used for dimensional comparisons.

Throughout this program, head, chest and pelvis acceleration, neck, lumbar and femur load and chest rotary potentiometer displacement data were inspected for any indications of mechanical noise.

Inspection

The inspection of the FTSS and Denton chest jackets and spine boxes were conducted by measuring the dimensions and comparing them to either the Part 572 specification or the SAE proposed specification. This assured that the project was evaluating jackets that had not shrunk and were within the Part 572 requirements and that the new modified jackets complied with the SAE specification. The specifications for the jackets included three drawings: the chest flesh front, side view and the sternum pad. Five drawings defined the modified spine box with its new side plates [3, 4].

The mandrels were inspected by scanning them with a FARO Technologies 3-D Platinum Laser Arm [6] and InnovMetric Software Incorporated's PolyWorks 3-D scanning software [7]. A 3-D file of the Denton mandrel served as the reference. The best fit alignment between the mandrel from each manufacturer and the reference was established through use of the PolyWorks ImInspect program. Thirteen points were selected for comparison.

Certification Testing

The certification tests used to calibrate the dummy and to determine the repeatability and reproducibility of the HIII 5th Female dummy are defined by CFR title 49 part 572, section 130 [8]. The tests specified by Subpart O include head drop, frontal neck flexion and extension, thorax impact, knee, knee impact, and torso flexion. All tests were performed except for the knee impact test. The dummy set up for the thoracic impact and torso flexion tests is shown (Figures 4, 5).

Low Risk Deployment Testing

The new spine box attachment bolts were tightened to the 28 Nm recommended by SAE [4] whereas the original spine box bolts were tightened to 21.5 Nm [9]. Each of the four dummies was seated in driver low risk deployment position #2 in a 2006 Volkswagen Passat using the procedures specified in title 49 part 571.208 §26.3 [10] and TP-208-14 [11],

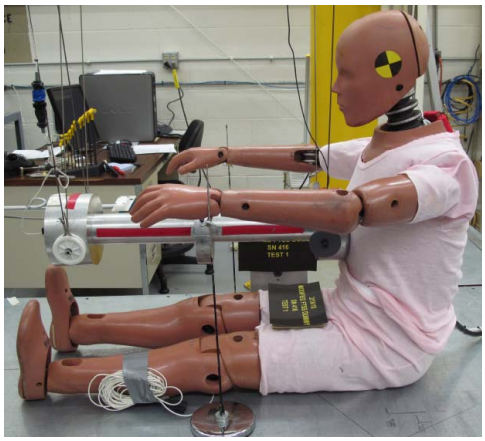

Figure 4. Thorax certification test.

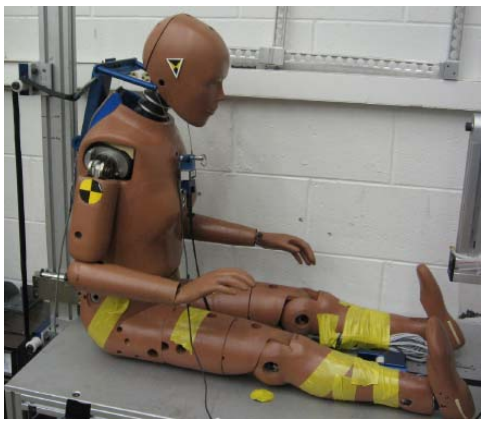

Figure 5. Torso flexion certification test.

with the chin on the rim of the steering wheel and the chest on the air bag module. The dummy is shown seated for a test in Figure 6. The tilt steering column was placed in the middle elevation and telescoped positions. Prior to the test, the dummy's position relative to the restraints and vehicle was measured. The second stage of the air bag fired 200 msec after the primary stage. After each test the steering column, steering wheel and air bag module were replaced and the dummy was inspected. The inspection evaluated rib security, spine box movement as well as verifying the torque on each bolt attaching the spine box to the lumbar load cell or adaptor. Any damage to the chest jacket and chest jacket foam was recorded. The distance between the ribs and the spine box was also measured to check for permanent rib deformation.

Sled Testing

Sled testing was performed using a 35 mph delta V

Figure 6. Low risk deployment test.

Sled Testing

sled pulse with a peak acceleration of 28g, and a duration of 92 msec based on a 2010 Taurus NCAP crash test. For each test the sled buck seats consisted of flat, rigid wood surfaces (depicted by the blue line in Figure 7), designed to position the dummies, relative to the seat belt anchorages, similar to the driver seating position in a 2010 Ford Taurus (depicted by the gray line in Figure 7). No knee bolster or attempt to restrict leg movement was used.

To achieve the seating reference locations, a 5th female dummy was seated in a 2010 Ford Taurus. Using a FARO arm, locations of the D-ring, retractor, and seat belt anchors were measured. Also dummy reference locations including H-point, head CG, shoulder, elbow, knee and tip of toes were measured. Comparisons were made between the dummy seated in a 2010 Ford Taurus and the dummy seated on the sled buck (depicted by the diamonds in Figure 7).

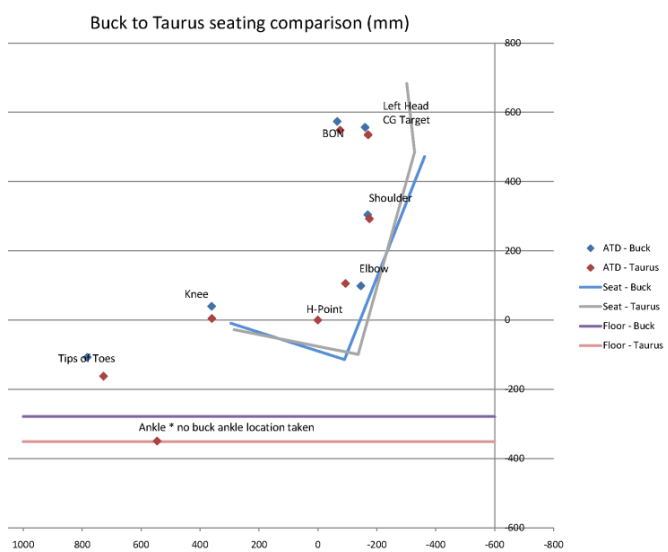

Figure 7. Seating from vehicle-to-buck.

For each test, the dummy was positioned on the hard seating surface by setting the pelvis and tibia angles to ~18.5 and ~57 deg respectively, and knee-to-knee distance to ~170 mm and recording 17 dummy and restraint system measurements. The example test setup is depicted in Figure 8.

Repeatability and Reproducibility

Three statistical measures were used to assess the dummy repeatability and reproducibility (R&R) of the dummy design. The traditional method used the coefficient of variation (CV) of peak response measures with the following assignments (Table 1). A CV less than **5**% is excellent; 5 – 8% is good; 8 – 10% is acceptable and above 10 is unacceptable [12].

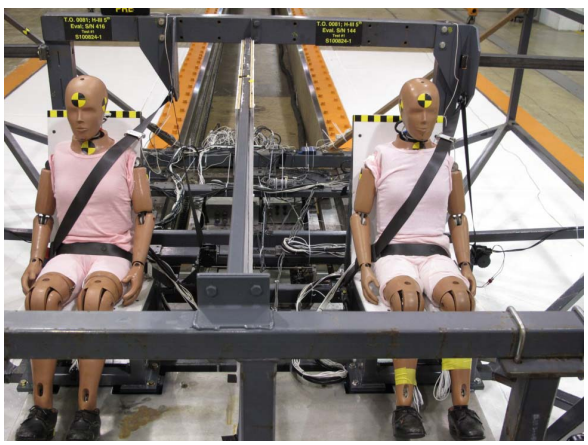

Figure 8. Sled test.

Table 1
Coefficient of Variation (CV %)

	Color Code
Excellent	Less 5
Good	5 to less than 8
Acceptable	8 to less than 10
Unacceptable	Greater than 10

This method was also applied to the time of peak response. Another method used is an average CV determined from all points in the central portion of the channel time history (excluding the lower fifty percent to avoid large variances and small means with possible division by zero). Shaw (2007) showed that with a sample size of six, an average CV below 5% is excellent [13]. In this test program, sample sizes were 3, 5, 8 and 16 so CV of sets with fewer than 6 samples would lead to larger confidence interval corridors. For example, the confidence interval for n= 3 is ± 10% and for n= 5 is ± 7%. Lastly, a comparison of the dummy means is made with the student t-test to assess if the difference between two dummy means is greater than 10% of the entire dummy population mean [13]. See Equations 1 through 3.

$$CV_{peak} = \frac{S}{\bar{X}} \qquad \text{Where:} \qquad (1).$$

S = standard deviation, $\bar{X}$= mean

$$CV_{50\%avg} = \frac{\sum_{i=1}^{n} \frac{S_i}{\bar{x}_i}}{n} \qquad \text{Where:} \qquad (2).$$

S_i = standard deviation, $\bar{x}$ = mean, n = sample size
i = ith sample

$$T = \frac{\sum_{i=1}^{d}[|(0.1)(\mu_i)| - |\mu_{1i} - \mu_{2i}|]}{S/\sqrt{d}} \qquad \text{Where:} \qquad (3).$$

μ_i is the mean of the means
μ_{1i} and μ_{2i} are sample means
d = number of pairs of data points
S = standard deviation of differences

If the two dummies are judged to be from the same population (i.e. they are reproducible) the same CV calculations performed for repeatability are made for the combined results from both dummies for reproducibility.

RESULTS

Inspection

Chest Flesh The inspection of the chest flesh resulted in several observations, the most important of which are where deviations occurred (Table 2). First, the access holes on the front of the jacket did not match the SAE drawing. Second, the diameters of the holes behind the breasts on the sternum pad are greatly different between the brands. Finally, the Part 572 thickness was not defined. The chest jacket thickness meets the new SAE specification and is 0.1 inch thicker at the bottom than the existing jacket.

Table 2.
Chest flesh assembly inspection results summary

Front View	**SAE**	**FTSS**	**Denton**
Access hole height	8.63	6.375	6.375
Hole diameter behind breast	none	0.625	0.375
Thickness	0.50	0.50	0.50

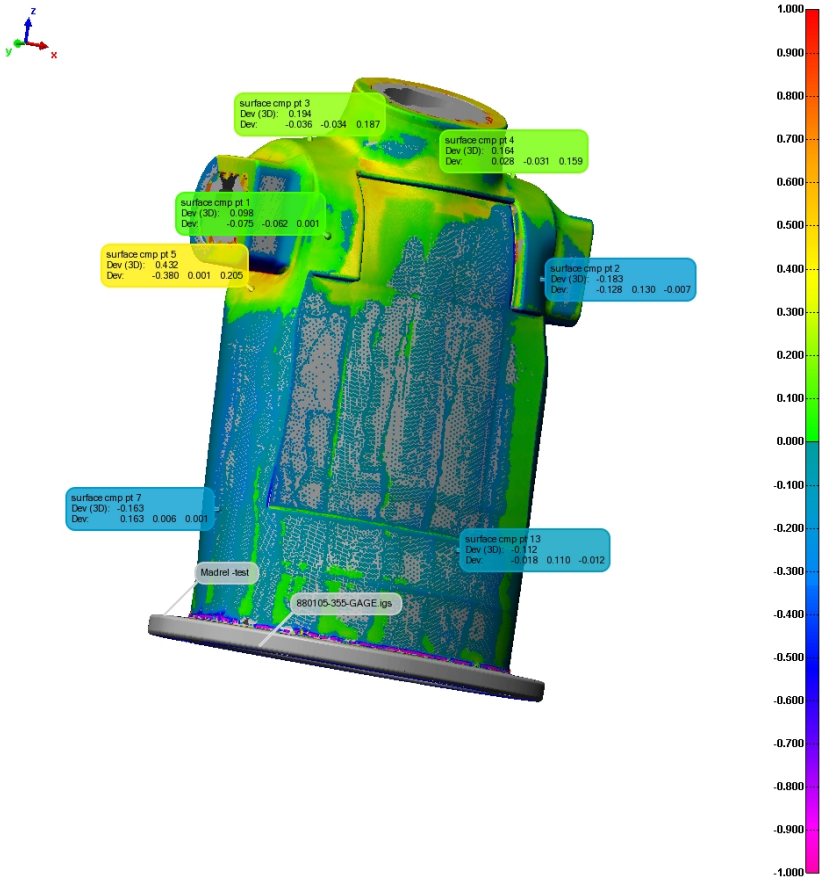

Figure 9. FTSS mandrel front view compared to Denton mandrel .Iges data set (mm).

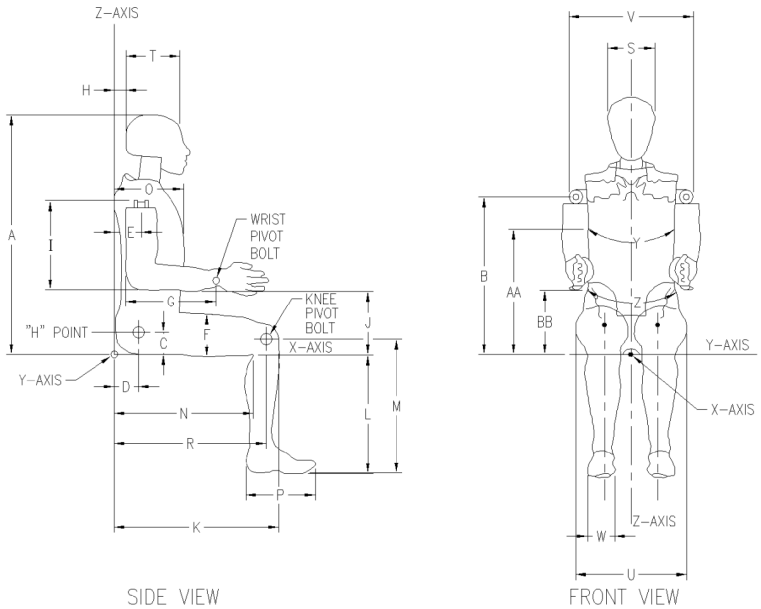

Figure 10. Reference diagram for anthropomorphic dimensions.

Table 3.
Dummy external dimensions summary

Description	Key	Specification (mm)	Results by Dummy (mm)				Pass
			D140	D144	F416	F509	
Total Sitting Height	A	774.7-800.1	784	786	778	784	Yes
Shoulder Pivot Height	B	431.8-457.2	450	445	448	450	Yes
Shoulder Pivot from Backline	E	68.6-83.8	81	82	83	81	Yes
Head Back to Backline	H	43.2-48.2	45	46	46	45	Yes
Elbow Rest Height	J	182.8-203.2	193	199	185	195	Yes
Chest Circumference with Jacket	Y	850.9-881.3	865	862	869	869	Yes
Reference Location for Chest Circumference	AA	332.7-358.1	345	345	354	350	Yes

Table 4.
Torso flexion certification results summary

Hybrid III 5th Female Torso Flexion*						
Dummy No.	**Test No.**	**Test Site^**	**Initial Angle (deg)**	**Difference between Return & Initial Angle (deg)**	**Max Force at 45 deg during 10 sec (N)**	**Rotation Rate (deg/sec)**
Specification			**≤ 20**	**+/- 8**	**320-390**	**0.5 – 1.5**
D140	**Mean**	TRC	15.5	5.0	363.8	1.0
D144 w/modified parts	**Mean**	TRC1 n = 3	14.0	3.9	371.1	0.99
	Std. Deviation		0.8	0.6	6.4	0.01
	CV (%)		5.4	15.4	1.7	0.59
	Mean (Fails)	TRC2 n = 6	13.6	3.8	**427.7**	1.0
	Std. Deviation		1.6	0.8	39.9	0.0
	CV (%)		11.9	20.4	9.3	1.2
F416 w/modified parts	**Mean**	TRC1 n = 3	19.3	5.2	345.9	1.0
	Std. Deviation		0.9	0.3	22.5	0.0
	CV (%)		4.7	5.9	6.5	0.6
	Mean (Fails)	TRC2 n = 3	17.8	4.0	**408.0**	1.0
	Std. Deviation		1.1	0.3	23.8	0.0
	CV (%)		6.1	8.7	5.8	0.0
F509	**Mean**	TRC	17.1	5.1	379.4	1.0

*See table 1 for color key. ^All torso flexion tests were conducted at the TRC test facility with n = 2 for D140 and F509.

Table 5.
Thorax certification results CV summary

CV (%)*							
Dummy No.	**Test Site****	**Impact Speed (m/s)**	**Fmax in 50 – 58 mm (kN)**	**Fmax in 18 to 50 mm (kN)**	**Maximum Deflection (mm)**	**Internal Hysteresis (%)**	**Sample Size**
Specification		**6.59-6.83**	**3.9-4.4**	**≤4.6**	**50-58**	**69-85**	**n**
D144	1	0.2	0.7	0.4	0.2	0.6	3
	2	0.0	0.6	1.8	1.2	0.0	5
	Mean	6.74	4.30	4.18	54.99	73.63	8
	1 & 2	0.4	0.8	1.4	1.8	1.2	8
F416	1	0.0	0.9	1.6	0.6	1.4	3
	2	0.2	1.3	1.8	2.5	0.6	5
	Mean	6.73	4.25	4.25	52.81	72.25	8
	1 & 2	0.2	1.5	2.0	4.2	2.2	8
Combined	1 &2	0.2	1.3	2.0	3.9	2.1	16

*See table 1 for color key. **Test site 1 = VRTC, Test site 2 = TRC

Chest Jacket Mandrel The thirteen points examined on each mandrel were within the 0.5 mm NHTSA tolerance that was applied to the 3-D digital model (see Figure 9 for the front view).

Spine Box The unmodified spine boxes were within tolerances specified by Part 572. The modified spine box uses a small (0.75" x 0.2") plate attached to each side wall of the thoracic spine that is secured with three DOT Side Impact Dummy (SID) modified 5/16-18 x 5/8" screws. No anomalies were noted.

Chest Flesh on Mandrel In contrast to the original jackets the recently manufactured modified jackets easily fit on the mandrels, and the zipper closed easily demonstrating good fit for use on dummies.

Assembled Dummy The dummy external dimensions associated with the spine box and chest jacket were within the tolerance specified on the Part 572 and SAE drawing (Figure 10 and Table 3).

Certification Testing and Repeatability and Reproducibility

The heads, necks, thoraxes, torsos and knees of all four dummies were qualified. Only the thorax and torso flexion tests of the modified dummies were evaluated for repeatability and reproducibility since the changes to the dummy involve only those regions.

Torso Flexion Certification Test The torso flexion certification results CV summary is shown in Table 4. Although the repeatability was acceptable both the Denton and FTSS brands failed the maximum force requirement at forty-five degrees flexion. For this reason their reproducibility was not calculated. Figure 11 illustrates the interaction between the thorax chest jacket and pelvis skin near the 45 degree flexion angle. Although some interaction is normal it was recommended to the DTEC to address this in the torso flexion test [5].

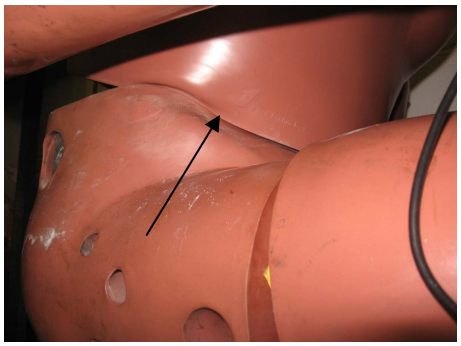

Figure 11. Torso flexion resistance by modified chest flesh interaction with pelvis flesh.

Thorax Certification The thorax certification results CV summary is shown in Table 5 and the thorax response in Figures 12 through 14. The thorax response was excellent for both modified dummies, with all dummies passing the certification corridor. The FTSS modified dummy force time history and deflection time history responses in the Vehicle Research and Test Center (VRTC) tests occur slightly before the TRC responses.

All Other Certification Test The modified dummies passed the head, neck, and knee certification tests. Since they were not influenced by the changes to the thorax, the data will not be presented here. However, during the thorax certification test, noise was evident in the chest and pelvis accelerometers. Once the femur was reinstalled and tightened with the nylon plunger in the pelvis the noise did not reoccur.

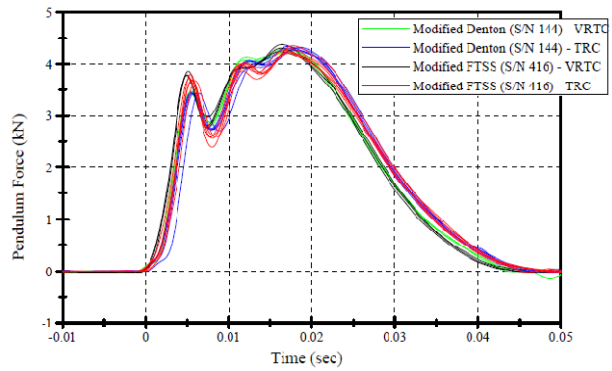

Figure 12. Thorax pendulum resistive force time history for both modified dummies.

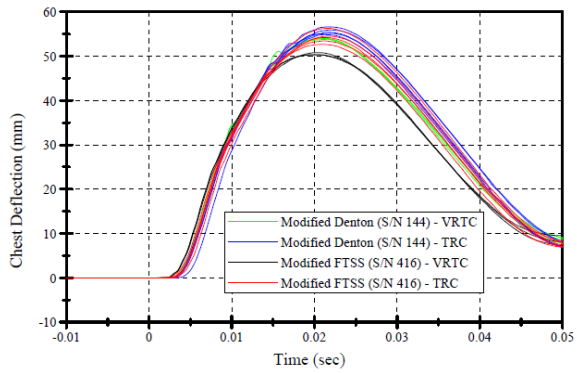

Figure 13. Chest deflection time history for both modified dummies.

Low Risk Deployment Testing

Each dummy was tested once. The chest acceleration and deflection response is shown in Figures 16 and 17. These plots show a significant test-to-test variation in both chest acceleration and chest

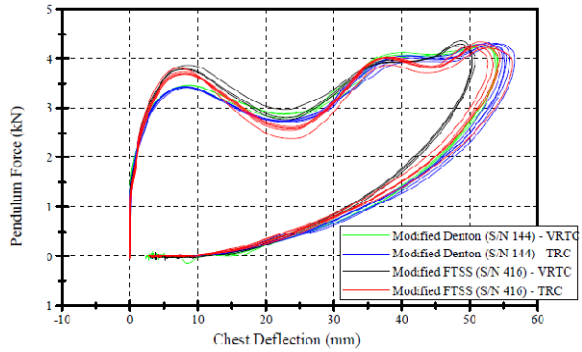

Figure 14. Thorax force deflection response for both modified dummies at two labs.

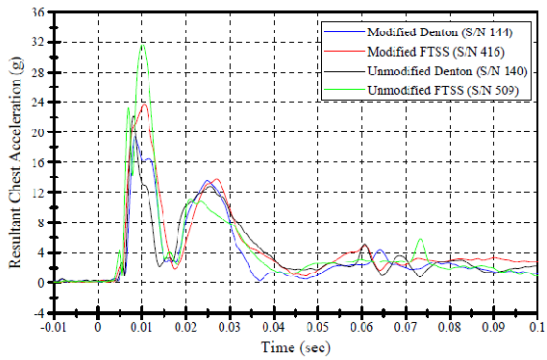

Figure 16. Chest acceleration during exposure to low risk deployment.

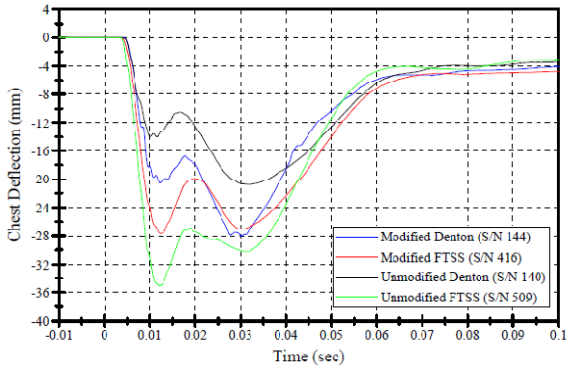

Figure 17. Chest deflection during exposure to low risk deployment.

deflection. Clearly some of the differences seen in chest deflection among the dummies are due to the initial setup, specifically, the chest-to-air bag module distance. While the average distance was 41 mm, dummy 509 was the closest to the module at 19 mm and dummy 140 was the farthest at 51 mm. Variations in air bag deployment may also be contributing factors to the variation in chest response.

Injury Criteria The chest accelerations and deflection of the dummies with the modified chest jacket and spine box fared better than their counterparts with the original jackets (Table 6). The percentage of injury values achieved by both the modified and the unmodified dummies are well within the allowable FMVSS No. 208 Injury Assessment Reference Values (IARV) for low risk deployment. The modified dummies' largest variation among the injury values is 13% in the chest acceleration (3ms Clip) and the chest deflection. Comparison of the variation in IARV range for the original dummies is similar and the range for chest acceleration in the unmodified dummies is slightly larger (15%).

Durability Spine box post-test torque checks indicate no loosening of the spine box anchor bolts. Also, there were no mechanical noise issues observed in the chest acceleration data or any other channel. The rib depths remained constant across the series of four low risk deployment tests for each dummy and were nearly identical between dummies with a rib 1 range from 160 to 162 mm and rib 5 ranges from 154 to 156 mm.

Table 6.
Low risk deployment results percentage of injury value summary

Injury Criteria	FMVSS 208 Max IARV	Percentage of Injury Value Achieved*			
		140	144 Mod	416 Mod	509
HIC[15]	700	2	2	3	3
Clip[3 ms]	60	27	22	35	42
Chest Deflection	52	54	40	53	67
Max [NIJ]	1	30	30	29	44
Max[NTE]	1	30	30	29	44
Max[NTF]	1	23	20	21	22
Max[NCE]	1	6	7	10	4
Max[NCF]	1	21	17	22	26
Neck Load Tension	2070	27	29	29	34
Neck Load Compression	2520	5	3	7	6
Left Femur	6805	6	N/A	5	6
Right Femur	6805	7	N/A	6	7

*displayed for unmodified dummies serial no. 140 and 509 and modified dummies 144 and 416

Sled Testing

Chest acceleration and deflection traces are shown in Figures 18 and 19. Although the sled acceleration and velocity were very consistently reproduced it should be noted that there is some variation in the lap belt force (CV 12%) and inner buckle firing time (CV 26%) that may introduce data variations (Table 7).

Examining the individual channels that factor into the IARVs provides further insight into the dummies' R&R for HIC, Chest Depth and Nij (Table 8). The Injury Assessment Value (IAV) results achieved by both the modified and the unmodified dummies are shown in Table 9. All measured IAVs are well within the allowable FMVSS No. 208 IARVs for a 2010 Ford Taurus frontal crash test simulated by these low risk deployment tests. The original dummies exhibit similar means to the modified dummies.

Repeatability and Reproducibility The modified dummies exhibited excellent to good repeatability in peak IAVs, excluding Nij, despite variation in lap belt forces and inner seat belt forces and inner seat belt buckle firing times. It should be noted that the chest deflection repeatability is excellent to good and is one of the primary interests of this report. The 10 % reproducibility of the chest deflection IARV was acceptable (Table 8). The modified dummies' peak chest deflection and acceleration CLIP [3ms] were nearly identical to those of the unmodified dummies.

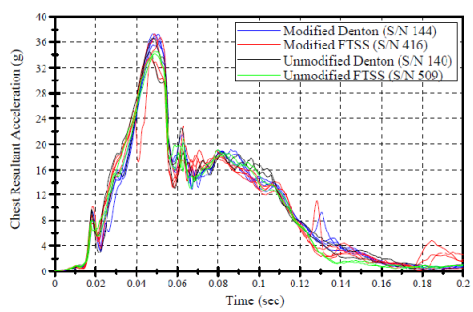

Figure 18. Chest acceleration from sled tests

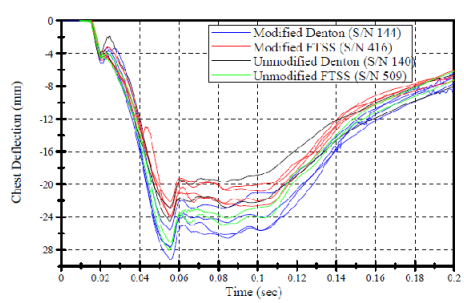

Figure 19. Chest deflection from sled tests

Table 7.
Sled and Restraint Inputs to HIII 5th Female Test Results Summary

Channel	Unit	Filter[1]	Avg	Std	%CV
Sled Accel.	g	60	28.0	0.11	0.4
Sled Vel.	mi/h	60	34.8	0.06	0.2
Lap Belt	N	60	5548.1	644.81	11.6
	msec	60	52.2	1.56	3.0
Shoulder Belt	N	60	5717.0	144.43	2.5
	msec	60	50.2	0.38	0.8
Inner Buckle	msec	none	15.5	4.04	26.0
Limiter Volt.	msec	none	57.6	1.85	3.2
Limiter Amp.	msec	none	52.8	0.20	0.4
Buckle Volt.	msec	none	12.8	0.20	1.5

*See table 1 for color key.

Table 8.
Sled peak response results CV summary

Channel	144	416	144 416
	CV %*		
Head Res. Accel	1	2	3
Up. Neck X-Axis Force	4	3	5
Up. Neck Z-Axis Force	2	8	5
Up. Neck Y-Axis Moment	3	10^	9
Low. Neck X-Axis Force	9	4	8
Low. Neck Z-Axis Force	4	9	7
Chest X-Axis Accel.	2	5	5
Chest Z-Axis Accel.	3	5	6
Chest Res. Accel.	2	6	4
Chest Defl. X-Axis	5	4	10
Pelvis X-Axis Accel.	2	6	7
Pelvis Z-Axis Accel.	4	10^	6
Pelvis Res. Accel	1	7	6
Femur L. Z-Axis Force	2	4	5
Femur R. Z-Axis Force	1	4	8

*See table 1 for color key. ^ Exceeded 10 but rounded here.

Durability Results Although there were minor screw impressions on the foam layer inside the dummy jackets after the sled tests, no problems with dummy durability was observed. The chest depth for each dummy remained constant across the series of six sled tests and was nearly identical between dummies. There were also no indications from torque measurements on the spine box anchor screws that the spine box anchor screws had loosened. Finally, a

Table 9.
Sled injury assessment value results mean and CV summary

Simulated 2010 Taurus with NCAP 2010 Taurus Sled Pulse*											
Channel	**Units**	**Filter[1]**	**IARV**	**D144**		**F416**		**Combined**		**D140**	**F509**
				Avg^	**CV %**	**Avg**	**CV%**	**Avg**	**CV%**	**Avg**	**Avg**
HIC[36]		1000	1000	196	7	183	2	191	6	191	197
T1(Begin)	ms	1000		64	1	60	1	62	3	64	63
T2(End)	ms	1000		100	1	96	1	98	2	100	99
Avg. g T1 to T2	g	1000		31	3	30	1	31	3	31	31
HIC[15]		1000	700	107	5	88	3	99	11	98	92
T1(Begin)	ms	1000		67	1	65	1	66	2	67	64
T2(End)	ms	1000		82	1	80	1	81	1	82	79
Avg. g T1 to T2	g	1000		35	2	32	1	34	4	33	33
Max[NTE]	N	600	1	0.45	9	.51	4	0.48	10	0.36	0.65
Max[NTF]**	N	600	1	0.30	3	.16	10	0.24	30	0.29	0.24
Max[NCE]**	N	600	1	0.08	59	.21	6	0.14	58	0.15	0.22
Max[NCF]**	N	600	1	0.11	107	.01	78	0.07	144	0.01	0.01
Neck Loads- Max Shear -X-Axis	g	1000	1950	546	4	508	4	530	5	525	622
Neck Loads- Max Axial Z-Axis	N	1000	2520C 2620T	1190	2	1160	2	1177	2	1156	1135
Clip[3 ms]	g	180	60	36	2	34	5	35	5	35	34
T1 (Begin)				46	2	44	2	45	2	44	47
T2 (End)				49	1	47	2	48	2	47	50
Chest Deflection	mm	600	52	28	5	23	4	26	10	24	28
Femur Loads Right Max Tension	N	600	6805	2823	1	2478	5	2675	8	2627	2522
Femur Loads - Left Max Tension	N	600	6805	3011	2	2803	5	2922	5	2777	2737

* See table 1 for color key.
^n=4 for D144 and F416. N= 2 for D140 and F509.
**experience has shown that small values in Nij have large coefficient of variations due to variations between the time of peak moment and peak axial force. Note: generally as the magnitude of the Max Nij decreases from 0.5 to 0.01 the repeatability and reproducibility decreases and is in the red zone.

review of the data channels did not indicate any signs of metal-to-metal contact within the dummy during the test and there were no signs of unexplained spikes in the chest acceleration or any other channel.

DISCUSSION

Inspection

The two brands of chest jackets are now identical in appearance and compare well to the SAE drawings, although there are slight variations in the height of the access holes and in the hole diameter in the sternum pad with respect to the drawing. The spine boxes, mandrels and assembled dummies were within tolerance of the SAE drawings. The modifications to the spine boxes are identical, each having three countersunk through holes in each side plate through which three 5/16 x 18 x 5/8" screws anchor the spine box to the lumbar load cell or load cell simulator.

Certification Testing

The dummies configured to the CFR passed the certification tests. However, while the modified dummies passed the thorax Part 572 response requirements, they did not pass the torso flexion requirement. The torso flexion force response was high; therefore, the torso is stiff in flexion. Evaluation of the new harmonized jackets for the Hybrid III small female dummy by the Japan Automobile Manufacturers Association, Inc. and Japan Automobile Research Institute also noted stiffer torso flexion responses [14].

Repeatability and Reproducibility

The thorax of the dummies proved to be repeatable and reproducible (Table 10). However, while the performance in torso flexion testing was repeatable, it did not pass the maximum force requirements at 45 degrees flexion. Therefore the reproducibility of the dummy in torso flexion was not determined.

The SAE Hybrid III Dummy Family Task Force shared similar findings [14]. Table 10 shows a comparison between the two methods used to determine the CV. The thorax reproducibility is excellent as determined by both the $CV_{50\%avg}$ and CV_{peak} methods.

Low Risk Deployment Testing

During certification testing dummies were instrumented as they would be for low risk deployment and sled testing to look for signs of noise. Noise was detected in the chest and spine accelerometers but it was attributed to the femurs and resolved by tightening the femur plungers beyond the one g requirement specified for setting up the dummy for testing.

No anomalies other than differences between the chest deflection values were noted during the low risk deployment tests and the IARVs were within the FMVSS No. 208 requirements. The variations are attributed to a combination of variation in air bag deployment and variation introduced by the setup.

Table 10.
CV comparison of peak and time series with T statistic for repeatability and reproducibility

Thorax certification response in corridor		Repeatability				Reprod.		
		$CV_{50\%}$ Avg		CV Peak				
		VRTC	TRC	VRTC	TRC	T-Statistic**	$CV_{50\%}$ Avg	CV Peak
Force	144	1.3	2.9	0.7	0.9			0.8
	416	1.6	3.1	0.9	1.3			1.5
	144 416	2.4	3.5	1.2	1.1	3.3	4.6	1.3
Deflection	144	0.7	2.3	0.2	0.5			1.8
	416	0.5	2.9	0.6	0.6			4.2
	144 416	3.7	3.0	3.7	2.3	2.3	4.6	3.7

*See table 1 for color key. ** - from VRTC Data. TRC T-statistic was even larger (4.9 for force and 111 for deflection). Note: Combining dummy serial nos. 144 & 416 indicate reproducibility.

Sled Testing

The IAVs were within established FMVSS No. 208 limits (Table 9). The modified dummies exhibited excellent to good repeatability in peak IAVs and acceptable reproducibility. The average CV method incorporates a significant portion of the time-history using a threshold for analysis from the first occurrence of 50% of the peak value to the last occurrence of 50% of the peak value. Except for the upper neck y-moment the average CV method indicates acceptable dummy repeatability with the repeatability for the chest acceleration and deflection good to excellent. However, the reproducibility of the head acceleration was poor. With the t-distribution

critical value = 2.353 the T statistic indicates that neither the head resultant acceleration, chest deflection, nor the y-moment were reproducible in the sled tests (Table 10). In contrast, the traditional method used for calculating repeatability and reproducibility, when applied to the time of the instantaneous peak responses, yields excellent results except for the neck y moment (Table 11).

Table 11.
Average CV repeatability & reproducibility for dummy serial numbers

Sled test response	$CV_{50\%avg}$			$T_{50\%}$
	144	**416**	**144 416**	**144 416**
Head Resultant Accel.	9.4	9.3	11.8	2.2
Chest Resultant Accel.	3.8	6.6	7.2	15
Chest Deflection	6.6	4.8	10.1	-52.3
Upper Neck Z-Force	6.1	6.6	8.0	9.9
Upper Neck Y-Moment	18.6	12.4	18.8	-13.7

Table 12.
Peak CV repeatability & reproducibility

Sled test time of peak response	CV_{peak}		
	144	**416**	**144 416**
Head Res. Acceleration	*1.8*	*1.6*	*3.7*
Chest Res. Acceleration	*2.5*	*6.2*	*5.0*
Chest Deflection	*0.2*	*0.2*	*0.4*
Upper Neck Z Force	*4.1*	*1.6*	*3.0*
Upper Neck Y-Moment	*4.3*	*10.2*	*10.8*

Durability

Although the modified dummies were only exposed to eight certification tests, a single low risk deployment air bag test and four sled tests there were no problems with dummy durability observed other than possible the neck on dummy serial no. 416. There were minor screw impressions on the foam layer inside the dummy that developed over the course of the four low risk deployment and six sled tests. There were no signs of noise and the tests showed that the phenomenon of spine box internal motion about mounting screws did not occur.

Drawings

The chest flesh, spine box, assembled dummy, and mandrel drawings were reviewed and prepared for federalization. The few minor dimensional issues identified on the chest flesh and sternum pad drawings can easily be resolved by changing a few drawings.

CONCLUSIONS

The FTSS and Denton modified chest jacket, spine box, chest jacket mandrel, assembled dummies and SAE drawings that were evaluated in this study provided comparable dimensional data and test results. A few areas were identified where the drawings were unclear and where the parts were dissimilar but these were minor inconsistencies.

The dummy passed the thoracic certification test with excellent repeatability and reproducibility based on peak CV. The average CV yields good reproducibility. While the repeatability of the torso flexion test maximum force at 45 degrees is acceptable, it fails the force limit. It should be noted that the existing chest jacket specified by the CFR passes the force limit and the torso flexion test. Further work on the jacket is needed to address the shortcomings evidenced by the torso flexion test results.

The modified dummies demonstrated the ability to assess the IARVs in low risk deployment and sled testing. There were no indications that noise was introduced into the dummy during low risk deployment tests, where the dummy's chest was exposed to air bags or as a result of the sled tests.

The dummy R&R was also examined for the sled tests. Based on peak CV values the resultant chest acceleration responses have excellent to good repeatability and excellent reproducibility. The values of the chest deflection also exhibited excellent to good repeatability but only acceptable reproducibility.

The dummies R&R was similar when applying the 50% CVavg method: the chest acceleration and deflection response repeatability of the dummies was excellent to good. The reproducibility of the chest acceleration was good and chest deflection was acceptable. Even though there is variability in the sled test restraints the dummies chest performed at an acceptable level.

Both the average CV and peak methods indicate unacceptable repeatability performance of the dummy serial no. 416 neck. While the neck passed the certification requirements exploration of the reasons for its unacceptable repeatability was not within the scope of this study. Further evaluation of the neck is planned and will begin by examining its repeatability and reproducibility in neck certification testing.

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EvaRID - ANTHROPOMETRIC AND BIOMECHANICAL SPECIFICATION OF A FINITE ELEMENT DUMMY MODEL OF AN AVERAGE FEMALE FOR REAR IMPACT TESTING

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ABSTRACT

Neck injury due to low severity vehicle crashes is of worldwide concern and it is well established that the risk of such injuries are higher for females than for males, even in similar crash conditions. In addition, recently developed protective systems have shown to be less protective of females than males. Hence there is a need for improved tools when developing and evaluating the performance of protective systems for occupants.

The objective of this study was to develop a finite element model of a 50th percentile female rear impact crash dummy model. The anthropometry of the 50th percentile female was specified based on data found in the scientific published literature and is called EvaRID (Eva - female/RID - Rear Impact Dummy). EvaRID is based on the same design concept as the 50th percentile male rear impact dummy, the BioRID. A first version, EvaRID V1.0, was developed in LS-Dyna. The dynamic response of EvaRID V1.0 was compared to data from rear impact tests with female volunteers. It was found that it is necessary to further adjust the stiffness of the joints in the spine in order to fully mimic the motion of the volunteers. In future, the EvaRID dummy model has the potential to be a valuable tool when evaluating and developing seats and whiplash protection systems.

INTRODUCTION

Whiplash Associated Disorders (WAD), so called whiplash injuries, sustained in vehicle crashes is a worldwide problem. In Sweden, such injuries account for ~70% of all injuries leading to disability due to vehicle crashes (Kullgren et al. 2007). The majority of those experiencing initial neck symptoms recover within a week of a car crash, however, 5–10% of individuals experience different levels of permanent disabilities (Nygren et al. 1985; Krafft 1998; the Whiplash Commission 2005). Whiplash injuries occur at relatively low velocity changes (typically <25 km/h) (Eichberger et al. 1996; Kullgren et al. 2003), and in impacts from all directions. Rear impacts, however, occur most frequently in accident statistics (Watanabe et al. 2000).

It is well established that the whiplash injury risk is higher for females than for males, even in similar crash conditions (Narragon 1965; Kihlberg 1969; O'Neill et al. 1972; Thomas et al. 1982; Otremski et al. 1989; Maag et al. 1990; Morris recover & Thomas 1996; Dolinis 1997; Temming & Zobel 1998; Richter et al. 2000; Chapline et al. 2000; Krafft et al. 2003; Jakobsson et al. 2004; Storvik et al. 2009). These studies concluded that the female injury risk was 1.5 to 3 times higher than the male injury risk.

It has been shown that existing concepts for whiplash protection seats in general are more effective for males than females (Kullgren and Krafft 2010). The risk reduction regarding permanent medical impairment was approximately 45% for females and 60% for males, Figure 1. These results suggest that the difference in effectiveness between males and females could differ for various seat concepts. It is important to further validate these differences and to understand the reason behind them in order to achieve better protection for females, but also for males.

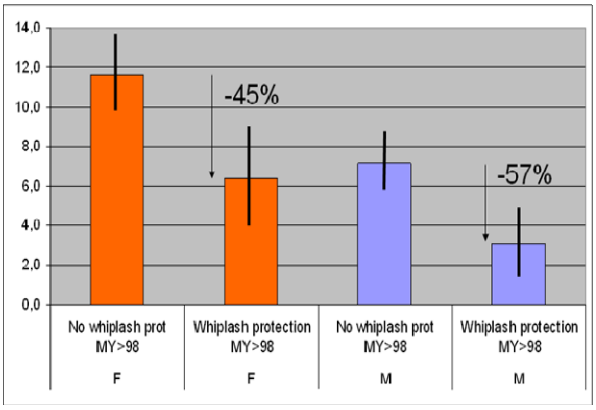

Figure 1. Whiplash injury reduction for females (F) and males (M) including 95% CI (based on the results from Kullgren and Krafft (2010)).

Females and males have different anthropometry and mass distribution, which may influence the interaction of the upper body with the seatback and head restraint and thus the injury risk. For example, the deflection of the seat frame, back rest padding and springs may depend on the mass and/or the centre of mass of the upper body with respect to the lever about the seatback hinge. The deflection of the structures of the back rest affects the plastic deformation, energy absorption and the dynamic head-to-head restraint distance as well as the rebound of the torso (Svensson et al. 1993; Croft 2002; Viano 2003). The motion of the head relative to the head restraint may be affected by seated height in relation to the head restraint geometry. It has been reported that females have a somewhat different dynamic response in rear volunteer tests, such as a higher head x-acceleration, a higher (or similar) T1 x-acceleration, a lower (or similar) Neck Injury Criterion (NIC) value and a more pronounced rebound (Szabo et al. 1994; Siegmund et al. 1997; Hell et al. 1999; Welcher & Szabo 2001; Croft et al. 2002; Mordaka & Gentle 2003; Viano 2003; Ono et al. 2006; Carlsson et al. 2008; Linder et al. 2008; Schick et al. 2008, Carlsson et al. 2010) than males.

Crash test dummies are used when developing and evaluating the occupant protection performance of a vehicle. The 50th percentile male crash test dummy correspond to a ~90th–95th percentile female with regards to stature and mass (Welsh & Lenard 2001), resulting in females not being well represented by the existing low velocity rear impact male dummies: the BioRID and the RID3D. Consequently, the current seats and whiplash protection systems are primarily adapted to the 50th percentile male without consideration for female properties, despite higher whiplash injury risk in females.

In view of the above, a European research effort was initiated under the ADSEAT (Adaptive Seat to Reduce Neck Injuries for Female and Male Occupants) project. The overall objective of ADSEAT is to provide guidance on how to evaluate the protective performance of vehicle seat designs aiming to reduce the incidence of whiplash injuries. The work concentrates on evaluating the protective performance of seats beneficial to female as well as male motor vehicle occupants. For this purpose a finite element crash dummy model of an average female is being developed. This new research tool will be used in conjunction with the BioRID II dummy model when evaluating enhanced whiplash injury protection.

This paper present the background research leading to the development of the first version of a finite element dummy model of an average female, called EvaRID, as well as the data used to develop the model. EvaRID is based on scaling an existing BioRID II dummy model in LS-Dyna. Background information on the size selection, anthropometry and the scaling method used, is described as well. The performance of the EvaRID V1.0 release is shown by comparing simulation results with volunteer data. This study was carried out within the ADSEAT project.

METHOD

EvaRID - Selection of size

Several sources were evaluated within the scope of the ADSEAT project to establish the size of female model to develop. One source used was the AGU Zurich database which records technical and medical information of persons who have suffered WAD. 2,146 data sets of females were analysed. It was found that the median height and weight of those females were 165 cm and 65 kg, respectively. The data sources contained basic measurements such as whole body height and weight. More specific

measurements such as seated height or the dimension of individual body parts were not available.

Comparing these measurements with data of the female population in different European countries showed that the weight and height found for the female most at risk correspond quite well with the average weight and height of females in the European countries; that is 165 cm and 66 kg, Table 1.

Table 1.
"Average" female anthropometries of the general population in different European countries

Country	Height [cm]	Weight [kg]	Age [years]
Austria[f, h]	167	67	43.2
Czech Republic[f, g]	167.3	-	41.9
Germany[d, f]	165	67.5	45.2
Finland[f, g, i]	164.7	69-83	43.7
France[b, f, g]	161.9	62.4	40.9
Italy[f, g]	162	-	44.8
The Netherlands[e, f]	166.8	68.1	41.2
Norway[c, f, g]	167.2	-	40.2
Spain[f, g]	161	-	42.5
Sweden[f, j]	166.8	64.7	42.6
Switzerland[a, f]	164	49-67	42
United Kingdom[f, i]	161.6	67	41.3
Average of the above given measures	**164.6**	**66.3**	**42.5**

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[g] http://www.disabled-world.com/artman/publish/height-chart.shtml
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Additionally, the documentation regarding choice of crash dummy size in the WorldSID project was studied (Moss et al. 2000) in order to evaluate if that data could give guidance when selecting the size for the 50th percentile female model. When establishing the anthropometry of the WorldSID midsize male crash dummy (Moss et al. 2000) the average weight and height of the average occupant was similar to that of Schneider et al. (1983).

Another source of data for the distribution of weight and size of the occupants sustaining whiplash injuries can be seen in the insurance company Folksam´s database that include vehicle occupants reporting injuries in car crashes. The data include 1,610 female occupants in total, aged 18 and above at an average age of 46 years. The average stature was 165.9 cm and the average weight was 65.9 kg, which correspond relatively well with the average European results presented above.

Anthropometry specifications

As an initial input to the development of the EvaRID model the following data were collected:

- Dummy (model) total weight and if possible, each assembly weight
- Seated height
- All related chest dimensions (depth and width)
- All related pelvis dimensions (depth and width)
- Lengths of limbs (distance between joints)

The data for the 50th percentile female was mainly collected from the following references:

1. Schneider et al. (1983): The goal of this study was to define the anthropometry of a crash test dummy family. Initially, this dummy family consisted of two female dummy members (5th and 50th percentiles), and two male dummy members (50th and 95th percentiles). In the first part of the project, data was collected and analysed for all four dummy members, but it was later decided that the 50th percentile female dummy member should be dropped. The statures, seated heights and weights of the dummy family members were defined based on the National Health and Nutrition Examination Survey (HANES) of 1971–1974 by Abraham et al. (1979). According to Young et al. (1983), the HANES survey provides the most current and appropriate general population model available for US adult females. The HANES data was collected on 13,645 individuals representing the 128 million persons aged 18–74 in the US population.

2. Diffrient et al. (1974): This reference "incorporates extensive amount of human engineering data compiled and organised by Henry Dreyfuss Associates over the last thirty years, including the most up-to-date research of anthropologists, psychologists, scientists, human engineers and medical experts."
3. Young et al. (1983): This research was part of a series of studies designed to obtain information about mass distribution characteristics (including moment of inertia and centre of volume) of the living human body and its segments, and to establish reliable means of estimating such properties from easily measured body dimensions. The study was based on 46 adult female subjects, selected to approximate the range of stature and weight combinations found in the general U S female population. The sampling plan for this study was to achieve a stature and weight distribution comparable to that found in the civilian female US population as reported in the HANES of 1971–1974 by Abraham et al. (1979).

Both Young et al. and Schneider et al. derived stature, weight, and seated height for the 50th percentile female from the HANES data. However, as Young only considered a limited age range (21–45 years) compared to Schneider et al. (18–74 years), the latter source was used for extracting the anthropometric data.

In addition to the above sources, anthropometric data taken from the ergonomic programmes GEBOD and RAMSIS was used to validate the collected data. Product Information from Humanetics (previously FTSS) was also used to collect information on the BioRID II hardware dummy model for direct comparison of anthropometry data. Finally, for comparative purposes, part of the 50th percentile male data was based on McConville et al. (1980).

RESULTS

Schneider et al. (1983) described in detail how the anthropometry was defined for the 5th percentile female as well as the 50th and 95th percentile male crash test dummies. The same method was used, when appropriate, in order to establish the anthropometry of EvaRID. The stature, weight and seated height of EvaRID were based on Schneider et al. (1983) since this data set has defined the sizes of the existing dummies.

Table 2 summarizes the dimensions of the EvaRID model in comparison to the existing BioRID II model hardware. The stature, total weight and seated height were derived from Schneider et al. (1983). The assembly weights for the females are based on data reported by Young et al. (1983). The statures for the male was derived from Schneider et al. (1983), whereas the weights of each body part is based on data from McConville et al. (1980) or Schneider et al. (1983).

Table 2.
Stature, weights, seated height of EvaRID and 50th percentile male by Schnieder et al. (1983) and BioRID II.

Variable	50th percentile Female EvaRID	50th percentile Male Schneider et al. (1983)	50th percentile Male BioRID (FTSS)
Total stature [cm]	161.8	175.3	177
Total weight [kg]	62.3	77.3	78.24
Head [kg]	3.53	4.14	4.54
Thorax [kg] (including neck/spine)	17.06	24.73	26.61
Abdomen [kg]	2.56	2.37	
Pelvis [kg] (including flaps)	15.72	17.98[1)]	15.80
Arm upper [kg]	1.39	1.77	2.02
Arm lower [kg] (including hand)	1.15	2.02	2.23
Leg upper [kg] (excluding flaps)	5.71	5.33[2)]	5.99
Leg lower [kg]	2.84	3.59	5.44
Foot [kg]	0.62	0.98	
Seated Height [cm]	84.4	90.1	88.4

1) McConville et al. (1983): Flap: 3440 cc => Pelvis incl. flaps: 11964+3440=18844 cc (=17,98 kg, Volume*0.954)
2) McConville et al. (1983): Flap: 3440 cc => Upper leg excl flap: 9029-3440=5589 cc (=5332 kg, Volume*0.954)

The weight of body parts, absolute and relative compared to overall weight is provided in Table 3 and Figure 2 of the EvaRID and BioRID II dummy models.

Table 3.
Comparison of mass distribution (in percent of the total weight) of the BioRID II and the EvaRID

Body Part		EvaRID Mass [kg]	EvaRID % of total	BioRID II Mass [kg]	BioRID II % of total
Head	x1	3.5	5.7	4.54	5.8
Torso[1)] (incl. neck/spine)	x1	19.6	31.5	26.61	34.0
Pelvis (incl. flaps)	x1	15.7	25.2	15.80	20.2
Arm upper	x2	1.4	2.2	2.02	2.6
Arm lower (incl. hand)	x2	1.2	1.9	2.23	2.9
Leg upper (excl. flaps)	x2	5.7	9.2	5.99	7.7
Leg lower (incl. foot)	x2	3.5	5.5	5.44	7.0
TOTAL		**62.3**	**100**	**78.24**	**100**

1) The torso consists of the thorax, the abdomen and the spine (including the neck).

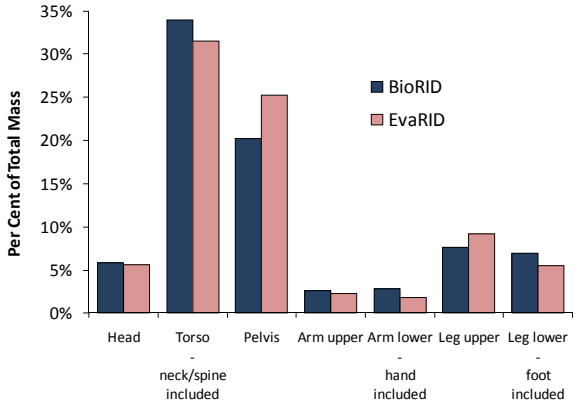


Figure 2. Comparison of mass distribution (in percent of the total weight) of the BioRID II and the EvaRID.

Table 4-6 provide remaining anthropometric data for the various body segments. The data was obtained from either Young et al. (1983) or Diffrient et al. (1974) since Schneider et al. (1983) only provided stature, weight and the seated height for the 50th percentile female.

Table 4.
Chest dimensions of EvaRID based on Diffrient et al. (1974) and Young et al. (1983)

Chest dimensions of EvaRID	Dimension [cm]
Parts of chest	
Shoulder (Diffrient et al. 1974)	
- Circumference	98.0
- Breadth	40.6
Armpit (Diffrient et al. 1974)	
- Circumference	89.2
Bust height (Young et al. 1983)	**116.4**
- Circumference	95.4
- Breadth	28.8
- Depth, mid-sagittal	17.8
- Depth, bust point (maximum protrusion of bra cup) (Diffrient et al. 1974)	23.1
- Distance, bust point – bust point	18.0
Chest below bust (Diffrient et al. 1974)	
- Circumference	79.0
10th rib height (Young et al. 1983)	**102.5**
- Circumference	75.9
- Breadth	25.7
- Breadth (Diffrient et al. 1974)	25.4
- Depth (Diffrient et al. 1974)	16.5
Waist	
- Circumference (Diffrient et al. 1974)	74.2
- Breadth (Young et al. 1983)	30.34

Table 5.
Pelvis dimensions of EvaRID based on data reported by Young et al. (1983)

Pelvis dimensions of EvaRID	Dimension [cm]
Part of pelvis	
Iliac crest height	**97.6**
Omphalion height	**95.9**
- Circumference	86.7
- Breadth	31.1
Buttocks	
- Circumference (standing)	100.1
- Breadth (standing)	37.3
- Depth (right buttock, standing)	24.1

Table 6.
Leg dimensions of EvaRID derived from Diffrient et al. (1974) and Young et al. (1983)

Variable	Dimension [cm]
"Inseam" height (Diffrient et al. 1974)	**75.2**
- Circumference	59.2
Gluteal furrow height (Young et al. 1983)	**71.7**
- Circumference	59.4
Mid-thigh height (Young et al. 1983)	**62.2**
- Circumference	51.9
- Depth	16.5
Knee height (Diffrient et al. 1974)	**45.7**
- Circumference (over the middle of the patella) (Young et al.1983)	37.0
- Breadth (across the femoral epicondyles) (Young et al. 1983)	8.8
Calf (Young et al. 1983)	
- Circumference (the maximum circumference of the calf)	35.6
- Depth (at same height as maximum circumference of calf)	10.8
Ankle height (Diffrient et al. 1974)	**8.1**
- Circumference (maximum circumference of ankle) (Young et al. 1983)	21.4
- Breadth (minimum breadth above medial & lateral malleoli) (Young et al. 1983)	5.4

EvaRID V1.0 – Model development

Since the EvaRID V1.0 model was based on the existing BioRID II model, the mass and dimensional ratios of the 50th percentile female should be used instead of the BioRID II. To meet the anthropometric requirements in terms of mass and dimension, the longitudinal dimensions and mass were initially scaled to obtain values related to the 50th percentile female. Breadth and depth dimensions were then established based on the scaling method of each body segment in detail, described in (Chang et al. 2010). The method was applied to all body parts.

Basic Scaling Methodology used to establish the Scale Factor Length (SFL), Scale Factor Breadth (SFB) and Scale Factor Depth (SFD) were calculated as follows. Firstly, the mass ratio of the EvaRID over BioRID II was calculated. Secondly, the SFL of EvaRID over BioRID II was determined based on the 50th percentile female data reported by Diffrient et al. (1974) and the dimensions of the BioRID II model. The SFB and SFD were then derived by taking the square root of mass ratio over SFL. Below are some general remarks related to the scaling of specific body parts:

The size of the head was scaled to meet all three dimensional requirements in depth, width and length. The mass requirement was met by adjusting the density of the skull.

The neck height was defined as the mastoid height less the cervical height. Considering the complexity of the neck modelling, the SFD was selected to be the same as the SFL. The SFB was assumed to be the same as the SFB for the torso, which was established by comparing the shoulder joint distance of EvaRID to the shoulder joint distance of BioRID II. Due to the lack of an accurate landmark of the mastoid and the cervical spine, the 50thpercentile male data from McConville et al. was used for the neck height. The mass ratio of 0.664 was derived from SFL*SFB*SFD and was slightly less than the mass ratios of the head (0.778) and the torso (0.737).

It should be noted that the longitudinal dimension of the limb is different from the total length of the limb. The longitudinal dimensions were measured as follows: The upper arm was measured from shoulder joint to elbow joint; the lower arm was measured from the elbow joint to the end of middle finger tip; the upper leg was measured from the hip joint to the knee joint; the lower leg was measured from the knee joint to the bottom of the heel.

The torso was divided into two sections; the upper torso and the pelvis. The upper torso in this study was defined as the torso without the pelvis and it extends from the cervical to the iliac crest. The EvaRID maintain the same back profile as the BioRID II due to the scaling factors used for the SFL and SLD being the same. The upper torso mass was derived by subtracting the pelvis mass from the torso mass. The breadth was defined as the distance between shoulder joints. The Scale Factor Depth was then derived as follows:

$$SFD_{Upper\ Torso} = \frac{Mass\ Ratio_{Upper\ Torso}}{SFL_{Upper\ Torso} \times SFB_{Upper\ Torso}} \quad (1)$$

Regarding the pelvis, no major difference was found between the dimensions of 50th percentile female and 50th percentile male pelvis. From the data published by Diffrient et al. (1974), the 50th percentile female has a distance of 180 mm between the hip joints, which matches the hip joint distance (179.6 mm) of BioRID II. Furthermore, the articles of Young's et al. and McConville's et al. indicate that there is little

difference between the 50th percentile female pelvis and the 50th percentile male pelvis. With regards to the pelvis angle the EvaRID maintains the same pelvis angle of 26.5 degrees as the BioRID II.

Finite element model of the seat

A seat model representing the test set-up was constructed for the initial evaluation of the dynamic response of the EvaRID V1.0 model. A Volvo 850 seat base dating from the early 90's was used as a seat in the tests. Differences in the seat base were considered to have negligible influence on the validation and therefore an available Taurus seat base was used during the evaluation. The seat back consisted of four stiff panels which were covered in a 20 mm thick layer of medium quality Tempur foam and lined with plush fabric (Volvo 850, year model 1993). The panels and foam were fitted according to the specified dimensions of the Volvo seat. The stiffness of the supporting springs on the seat were measured from and implemented in the finite element model. The head-restraint consisted of a stiff panel which was covered by 20 mm thick soft and 20 mm thick medium Tempur foam.

Dynamic data for model evaluation

The EvaRID V1.0 model was exposed to the same impact conditions as that of the volunteers in the test at 7 km/h in Carlsson et al. (2008). The dynamic response of the EvaRID V1.0 model was reproduced and compared to the responses in the volunteer tests. In Carlsson et al. a series of rear impact sled tests with eight female volunteers, representing the 50th percentile female, were performed at a change of velocity of 5 km/h and 7 km/h. The volunteer data is summarized in Table 7.

Table 7.
Volunteer data in Carlsson et al. (2008)

Female volunteers		Average
Age	[years]	24
Stature	[m]	1.66
Weight	[kg]	60
Seated height	[m]	0.88
Neck circumference	[m]	0.33

In the tests, the head of the volunteers were equipped with a harness with tri-axial accelerometers mounted on the left side and an angular accelerometer mounted on the right side, approximately at the centre of gravity on each side of the head. Two linear accelerometers, in x and z direction, were placed on a holder attached to the skin at four points near the spinal process of the T1. The upper body was equipped with a harness with tri-axial accelerometers mounted on the chest. Linear accelerometers were placed on the bullet sled and on the target sled. The test setup and the position of markers and instrumentation of the volunteers are shown in Figure 3 and 4. Additionally, the head-to-head rest distance and contact time and Neck Injury Criteria (NIC) were extracted from the data set. The volunteers wore a lap belt during the test.

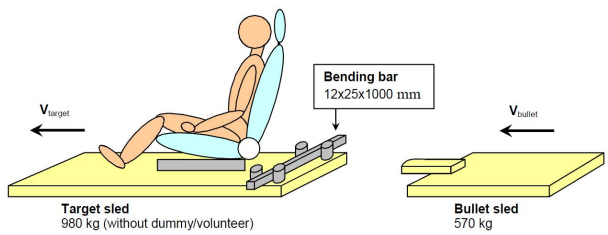


Figure 3. Sled set-up in the tests by Carlsson et al. (2008).

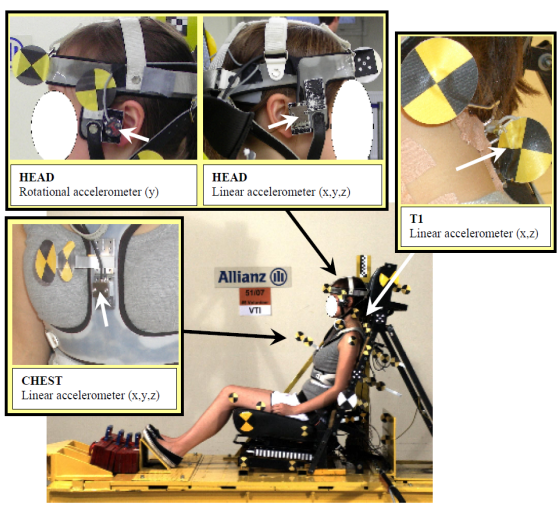


Figure 4. Position of markers and instrumentation on the volunteers in the tests by Carlsson et al. (2008).

Dynamic response corridors for the x-accelerations, the x-displacements and the angular displacements of the head, T1, and head relative to T1 were generated. Resulting corridors were created by +/-1SD from the average response of the volunteers from the tests and are depicted in Figure 5.

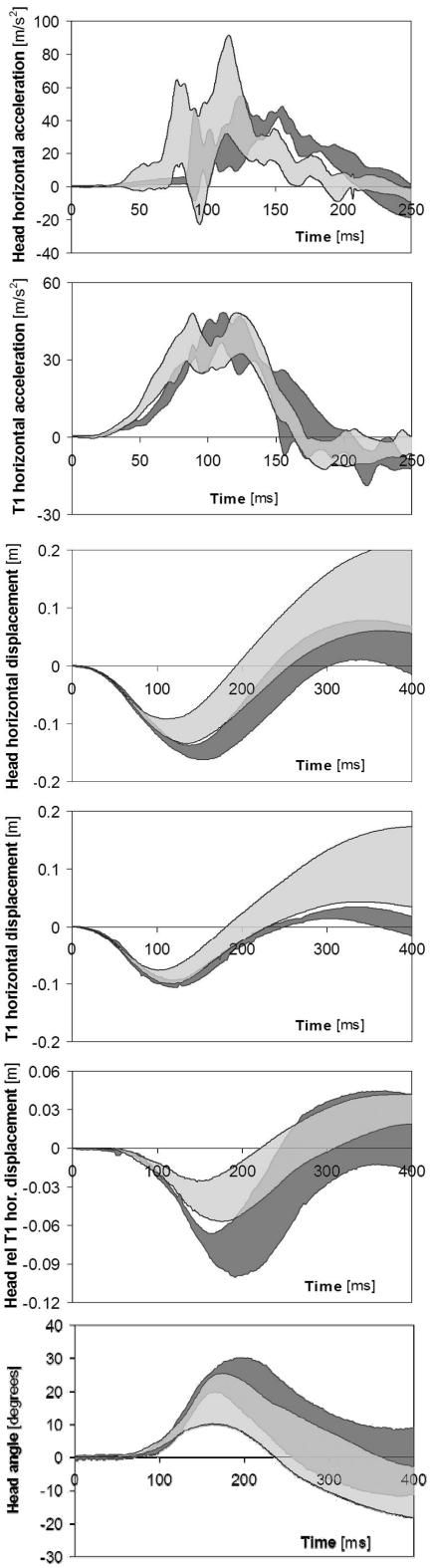


Figure 5. Examples of response corridors from Carlsson et al. (2008). The dark corridors represent tests with males and the light corridors represent tests with females.

RESULTS

Joint properties for the EvaRID V1.0 dummy model were adopted from the BioRID II dummy model. The curvature of the neck, spine etc. were the same for both models. A comparison between the EvaRID V1.0 and the BioRID II model is shown in Table 8 and Figure 6.

Table 8.
Comparison between the EvaRID VI.0 and the BioRID II

	Model	
	BioRID II (mm)	EvaRID (mm)
Head Total Height (top of head to chin)	215.9	203.0
Head Length	199.9	186.9
Head Breadth	157.6	145.8
Neck (C0-C1 joint to C7-T1 joint)	120.4	102.8
Torso (C7-T1 joint to Mid-point of hip joints)	526.5	479.4
Distance between shoulder joints	346.0	315.2
Upper Arm (shoulder joint to elbow joint)	261.4	264.0
Lower Arm (elbow joint to tip of middle finger)	248.8	234.0
Upper Leg (hip joint to knee joint)	405.5	389.6
Lower Leg (knee joint to bottom of heel along tibia)	495.5	457.0
Shoe Length	322.6	271.6

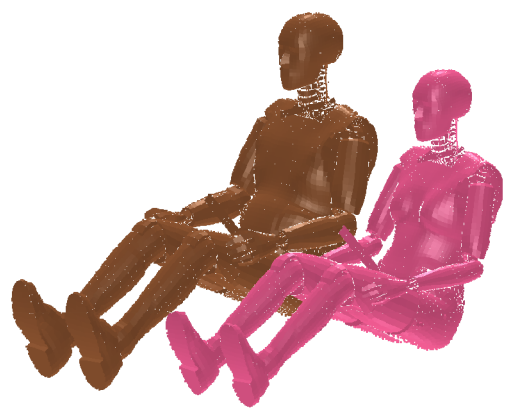


Figure 6. Comparison of EvaRID and BioRID II models.

Dynamic response

A pre-simulation positioning of the dummy model was conducted by dropping the EvaRID V1.0 model into the seat, allowing gravity to find its balanced position in the simulation. The seat was fixed to the ground with gravity as the only external force. Correlation of the initial position of the EvaRID V1.0 model with a volunteer was ensured by carefully checking each position. A representative example is shown in Figure 7. Once a balanced position was achieved by pre-simulation, the head panel was adjusted to equal the initial head-to-head rest distance, based on film analysis estimation measurements and a seatbelt was fastened before simulating the impact tests

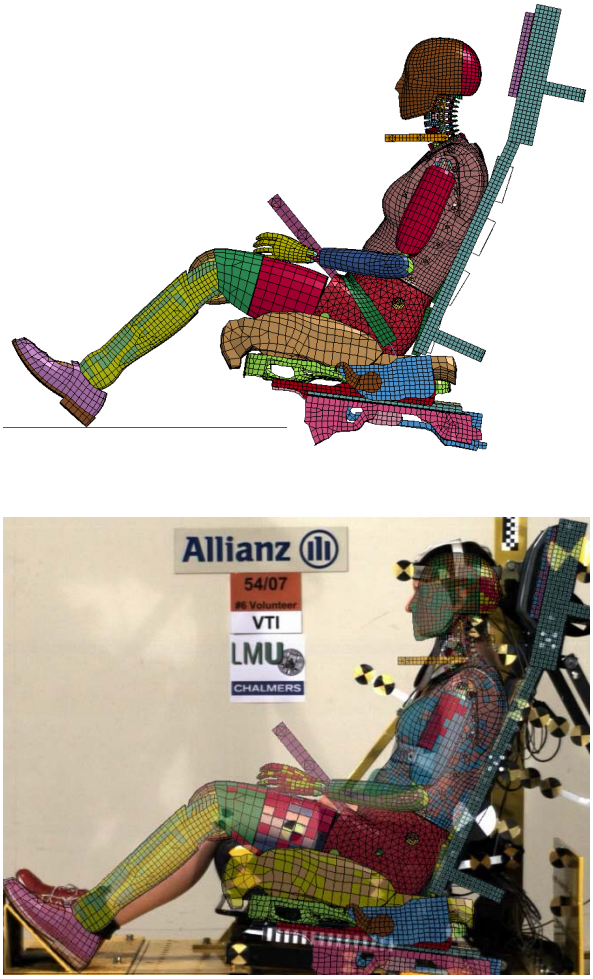


Figure 7. The EvaRID V1.0 finite element model seated in the seat and comparison of the initial posture with one of the volunteers.

The 7 km/h was used in simulations with the EvaRID V1.0. The results are shown in Figures 8-10, which compares simulated results using the EvaRID V1.0 model with response corridors and volunteer #50 who most resembled the EvaRID dummy model in terms of mass and initial position. Figures 8-10 shows the Head & T1 x-acceleration, Head & T1 x-displacement and the angular displacement of the Head & T1. T1 is the first thoracic vertebra and x-direction is in the horizontal plane.

Head and T1 x-accelerations

The head and T1 x-accelerations were mostly close to the test results and close to or within the test corridors.

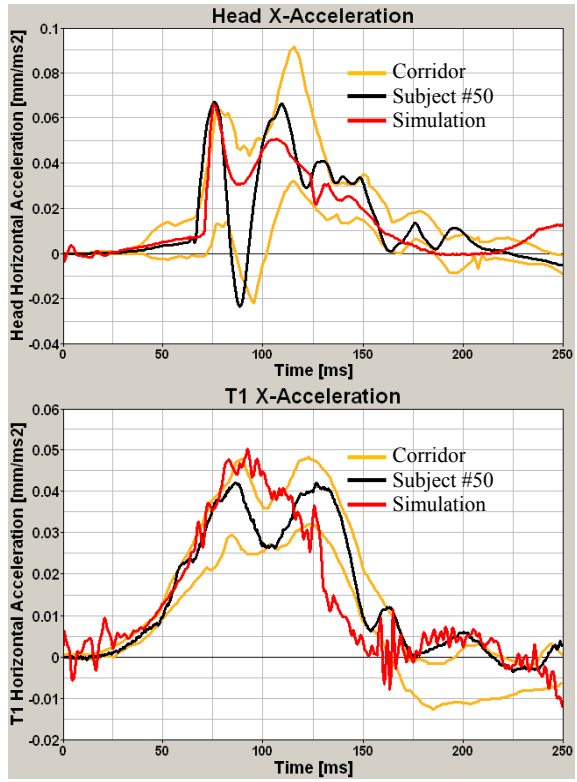
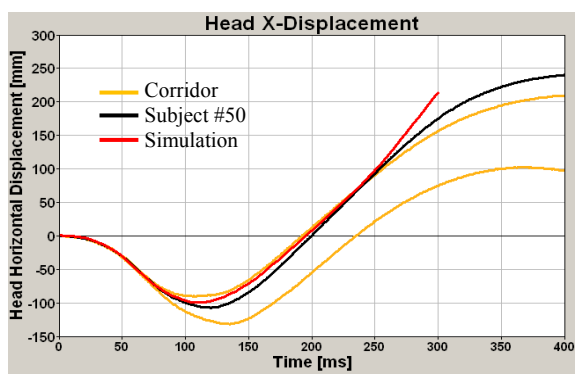


Figure 8. Comparison of head and T1 x-acceleration of the EvaRID V1.0 model response (red) with corridors constructed from all volunteer test data (orange lines) and volunteer #50 (black).

Head and T1 Displacements

The head and T1 x-displacements were close to the volunteer test results before the rebound at around 95 ms, Figure 9. From the T1 rebound and the simulation animation, it was observed that the EvaRID V1.0 model by design had a torso with much stiffer properties in extension than in flexion.



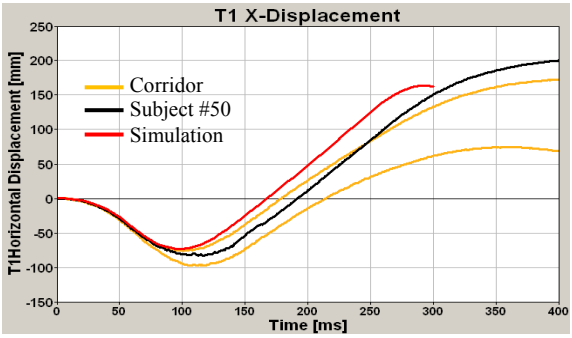


Figure 9. Comparison of head and T1 x-displacement of the EvaRID V1.0 model response (red) with corridors constructed from all volunteer test data (orange lines) and volunteer #50 (black).

Head and T1 angular displacement

The angular displacement of the head was within the corridor for the first 250 ms, however the angular displacement of T1 rotation response was not within the corridor for the first 240 ms, Figure 10. Thus the torso EvaRID V1.0 model has considerably stiffer properties in extension than flexion. The properties of the finite element model of the seat and seatbelt may also contribute to some of this discrepancy.

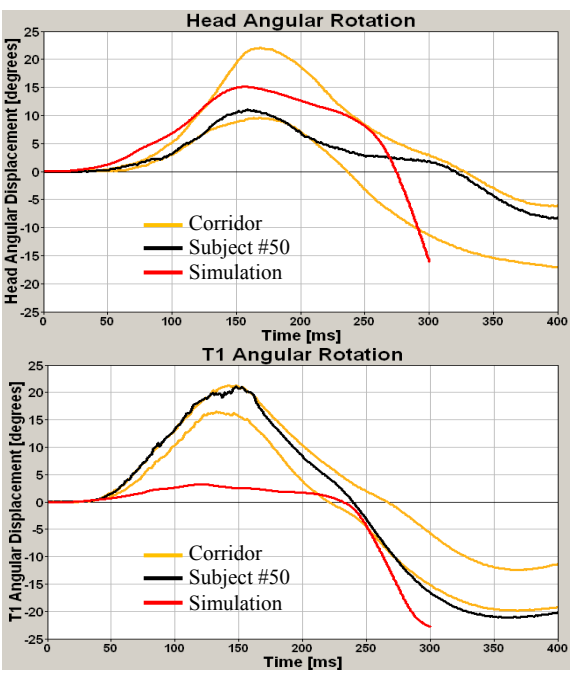


Figure 10. Comparison of head and T1 angular displacement of the EvaRID V1.0 model response (red) with corridors constructed from all volunteer test data (orange lines) and volunteer #50 (black).

Additional volunteer tests

Additional rear impact sled tests were performed in November 2010 with eight female volunteers at the change of velocity 7 km/h. The purpose was to receive input data for further improvements of the EvaRID mathematical model. The test setup was basically the same as in the previous volunteer test series (Carlsson et al. 2008), Figure 3 and 4, except for some changes to the design of the seat base and head rest to simplify the mathematical modelling of the seat. The head-to-head rest distance was increased to ~15 cm to delay headrest contact and improve the possibility of studying the retraction phase. Due to the increased head-to headrest distance the average acceleration was reduced to ~2g to ensure the safety of the volunteers. For comparison, the test series also included tests with eight male volunteers in the same test setup. The increased head-to-headrest distance allowed for more detailed comparison of the neck injury criteria such as NIC and neck loads for females and males.

The volunteers were selected based on their statures and masses being close the 50th percentile female and male, respectively, according to the University of Michigan Transportation Research Institute (UMTRI) study (Schneider et al. 1983). For the female volunteers, the stature range was 161–166 cm with an average of 163 cm, and the mass range was 54.5–66.8 kg with an average of 59 kg. In comparison to the UMTRI data, the female volunteers were on average 1% taller and 4% lighter. For the male volunteers, the stature range was 171–179.5 cm with an average of 176 cm, and the mass range was 69.8–81.0 kg with an average of 75 kg. In comparison to the UMTRI data, the male volunteers were on average 0.5% taller and 3% lighter.

The new headrest base consisted of plywood that was covered with foam blocks (polyethylene 220-E, 35 kg/m^3), and firmly attached to the stiff seat frame, Figure 11. The headrest was adjustable in height. The new seat base consisted of a rigid frame and was covered by plywood. The seatback (headrest excluded) was identical to the one used in the previous test series (Carlsson et al. 2008). In recent test series, two layers of Lycra fabric covered the seat back and seat base.

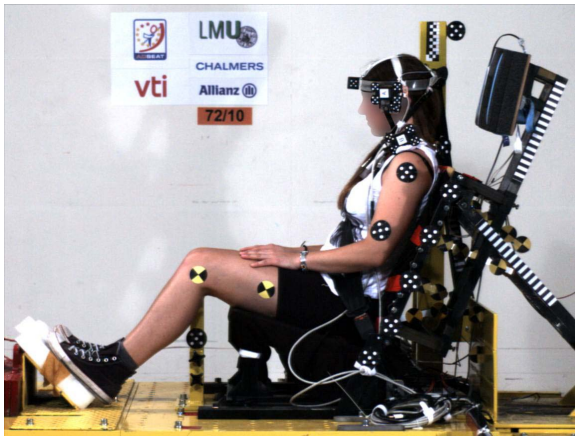


Figure 11. Test setup in voluteer tests run in November 2010.

The volunteers wore a pair of shorts and a T-shirt designed as a vest during the tests. The volunteers were seated, restrained by a three-point seatbelt and instructed to obtain a normal sitting posture, face forward, position their feet on an angled foot plate, place their hands on their lap and relax prior to the impact, Figure 11. Each female volunteer underwent two tests. By adding padding to the headrest, the headrest distance was 10 cm in the first test and 15 cm in the second. The males were only exposed to tests with a head-to-headrest distance of 15 cm.

The analysis of the data from these tests is ongoing and will be used to validate and further improve joint stiffness properties of the EvaRID mathematical model.

DISCUSSION

Based on real-world crashes it is well known that females have a higher risk of whiplash than males. A study has also shown that anti-whiplash concepts are more effective for males than females (Kullgen and Krafft 2010). Studies have shown many differences between males and females that may influence the difference in risk, for example differences in seated posture (Jonsson et al. 2007) and differences in muscle activity (Foust et al. 1973) and head to neck size (States et al. 1972) to mention just a few. In this study it is shown that females on average are 18% lighter and 7% shorter than males. In order to improve whiplash protection concepts for both males and females, but especially for females, it is important to better understand what influences the dynamic response and the risk of injury.

As a first step in the process of developing a finite element model of an average female the EvaRID V1.0 was created. The EvaRID V1.0 model was developed by scaling anthropometry, geometry and mass properties on the BioRID II dummy model. The weight distribution of EvaRID is somewhat different from the BioRID. EvaRID has a slightly lighter torso, a heavier pelvis and somewhat heavier upper legs. The joint stiffness properties of the spine, torso and neck were the same as that of the BioRID. In order to obtain a T1 angular motion as reported from volunteer tests, the stiffness of the EvaRID model will need to be tuned.

During the evaluation it was observed that the EvaRID V1.0 model showed less angular motion of the torso/spine in extension in comparison to the subjects, Figure 9. Due to having inherited the design and properties of the BioRID II dummy model, adjustments in joint characteristics were not made in the V1.0 model. Consequently, the T1 rotation is expected to be considerably less compared to the response of the real subjects during the extension motion. The rotation of the head relative to the T1 suggests that further improvement on the T1 (or spine) flexibility is important to correct the neck motion.

The next step in the process of developing the EvaRID model, tuning of the stiffness of the spine will be made. However, it may be the case that the construction of the BioRID II model does not contain the components needed to mimic the dynamic response of both males and females. In such case, new design features will be taken into consideration.

It was noted that the headrest was relatively close to the head in volunteer tests by Carlsson et al. (2008), which indicate that the head response is largely governed by the headrest properties and not entirely by the neck properties. Thus, these volunteer tests are less suitable for fine tuning the neck parameters, stressing the importance of using a greater headrest gap in the upcoming volunteer tests.

In this study, initial evaluation of the EvaRID V1.0 model was based on tests with one set of female volunteers. Additional test results from female volunteers that can be reproduced as finite element simulations are needed in order to further evaluate and develop the dynamic response of the EvaRID model.

CONCLUSIONS

A computational dummy model, called EvaRID, of a 50th percentile female to be used in rear impact tests was developed in the ADSEAT project based on anthropometry data found in the literature. Geometry and mass data was taken from sources serving as basis for the anthropometry of previous crash dummies. The model was obtained by scaling anthropometry, geometry and mass properties of an existing BioRID II model. Stiffness and damping properties of materials and discrete elements were kept in accordance with the BioRID II model in the initial EvaRID model version.

To evaluate how close the dummy model´s response was to that of a human, the EvaRID V1.0 was compared to the corridors and response curves gained in volunteer tests with females. Further work on the joint stiffness properties was found to be needed for the model to achieve a response fully within the corridors in the volunteer tests. Furthermore, additional volunteer data with a greater initial gap between the head and the headrest would be valuable to further improve the dynamic response of the dummy model. Such studies are included in the ADSEAT project. Once fully validated, the dummy model will be utilised in the design and evaluation of adaptive seat systems in order to provide enhanced neck injury protection for female occupants.

ACKNOWLEDGEMENT

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TRAFFIC ACCIDENT ANALYSIS TOWARDS THE DEVELOPMENT OF AN ADVANCED FRONTAL CRASH TEST DUMMY INDISPENSABLE FOR FURTHER IMPROVING VEHICLE OCCUPANT PROTECTION PERFORMANCE

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ABSTRACT

In this study, authors investigated and analyzed the injured body region and injury type for vehicle occupants from recent traffic accident statistics; that will be a basis to determine what human body regions should be evaluated in the frontal crash test, and what injury parameters should be measured utilizing an advanced test dummy. From the traffic accident statistics of the National Police Agency (NPA) in 2004 to 2008, the number of injured front seat occupants (i.e., drivers and passengers) by injury severity of vehicles damaged on front in the vehicle-to-vehicle accidents and the single vehicle accidents were collected. This then was analyzed by the seatbelt use, gender, and age group.

In five years from 2004 to 2008, the fatalities (the sum of drivers and passengers) due to the head and thorax injuries since 2005 tended to decrease conspicuously, whereas fatalities due to abdominal injuries were almost constant except for a slight increasing in 2008 from 2007. Reviewing the fatalities with regard to the seatbelt use, gender, and age group, more frequent injured body regions of the fatalities were the thorax, head, and abdomen. Of these, the most frequent injury types were organ injury on the thorax and abdomen, and skull fracture. Reviewing the fatalities by age group, the fatality rate was highest with injuries on the head in case of under 25 year olds, and was highest on the abdomen in cases of 26-64 year olds and over 65 year olds.

According to investigation and analysis in this study, in order to further improve the occupant protection performance during frontal crash, it was suggested that the abdominal injury that is impossible to evaluate in the Hybrid III and the injury measurement capability of the abdomen are particularly requested for a future frontal dummy.

INTRODUCTION

Understanding the real-world's vehicular accident situation is indispensable in the crash test dummy development process. The accident data analyzed statistically will be a basis to determine what human body region should be evaluated in the crash test, and what injury parameters should be measured utilizing a test dummy. Currently, though the Hybrid III frontal crash test dummy is used worldwide, a more advanced dummy with more biofidelity and higher measurement performance is requested for more sensitively evaluating the advanced restraint devices to further improve safety performance. Under these circumstances, an advanced frontal crash test dummy THOR (Test Device for Human Occupant Restraint) has been developed in the United States[1], and further development and evaluation of the THOR are in progress under the international efforts in order to further improve the biofidelity and measurement performance[2].

This paper reports the results of the investigation and analysis of the injured body region and injury type for

occupants in front-damaged vehicles based on recent traffic accident statistics. It is also pointed out that towards the future discussion regarding what human body regions should be evaluated in the frontal crash test, and what injury parameters should be measured utilizing the dummy.

ACCIDENT DATA SOURCE

The traffic accident database used in this analysis is the traffic accident statistics the NPA collected from 2004 to 2008. The NPA traffic accident statistics covers all traffic accidents with one or more injured persons who occurred in Japan, and these statistics data are managed by ITARDA (Institute for Traffic Accident Research and Data Analysis). The injury severities of vehicle occupants in statistics are classified in four groups as follows:

a) fatalities (deaths within 24 hours from the accident),
b) serious injuries (injuries requiring medical treatment over 30 days from the accident),
c) slight injuries (injuries requiring medical treatment less than 30 days from the accident), and
d) no injuries (only drivers are collected).

The damaged areas of vehicles are classified in terms of eight directions as shown in Figure 1.

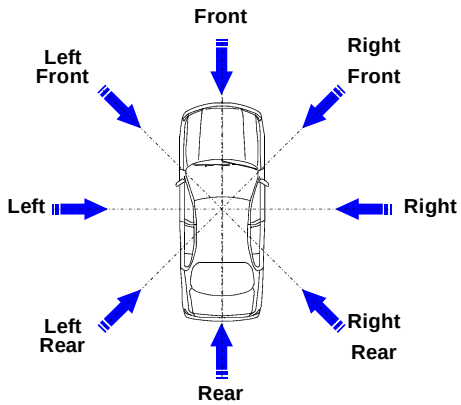


Figure 1. Classification of vehicle damage areas in NPA accident statistics

In this analysis, from the vehicle-to-vehicle accidents and the single vehicle accidents in the NPA traffic accident statistics from 2004 to 2008, vehicles damaged in the front, front right or front left of classifications shown in Figure 1 were defined as front-damaged vehicles. The number of these drivers and front passengers were collected by the classified injury severities. Both of vehicle-to-vehicle accidents and single vehicle accidents, the vehicle categories of the front-damaged vehicles were focused on the passenger cars and mini cars. As for the partner vehicles in the vehicle-to-vehicle accidents, not only the passenger cars and mini cars but also wagons and vans were subjects. As for the damaged areas of the partner vehicles, all the areas defined in Figure 1 were included. Accordingly the accident configurations of front-to-side crash and front-to-rear crash in addition to front-to-front crash were included. Regarding the single vehicle accidents, all opposing objects such as roadside structures, trees and parked vehicles were included.

GENERAL CHARACTERISTICS OF EXTRACTED ACCIDENT DATA

Table 1 shows the number of occupant casualties (i.e., the sum of fatalities, serious injuries, slight injuries) from front-damaged vehicles and the cumulative total number of occupant casualties from all vehicles covered by the NPA traffic accident statistics[3] from 2004 to 2008. The numbers of fatalities, serious injuries and slight injuries from front-damaged vehicles corresponded to 21.9 %, 19.3 % and 6.0 %, respectively, of the total number of occupant casualties from all vehicles from 2004 to 2008.

Table 1.
Casualties of vehicle occupants from 2004 to 2008

	Occupants in Frontal Crashed Vehicles	Occupants in All Crashed Vehicles
Fatalities	*2,566 (21.9%)*	*11,722*
Serious Injuries	*18,965 (19.3%)*	*98,348*
Slight Injuries	*194,851 (6.0%)*	*3,274,738*

Drivers and Front Passengers by Injury Severity

Figure 2 shows the ratios of drivers and front passengers of the total number of front-seat casualties in vehicle-to-vehicle accidents and single vehicle accidents by injury severity. Both of vehicle-to-vehicle

accidents and single vehicle accidents, irrespective of injury severity, the drivers accounted for about 80 % of the front-seat occupant casualties. Comparing to the occupant casualties by injury severity between vehicle-to-vehicle accidents and single vehicle accidents, the casualties from single vehicle accidents outnumbered the casualties from vehicle-to-vehicle accidents by 7:3 (1,879 : 687 persons) for fatalities, 6:4 (10,864 : 8,101 persons) for serious injuries, and 3:7 (53,979 : 140,872 persons) for slight injuries. Thus, the ratio of casualties of single vehicle accidents was trend on a high as the injury severity increased.

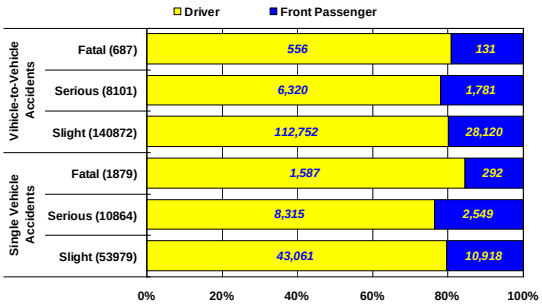


Figure 2. Ratio of drivers and front passengers by injury severity

Casualties by Seatbelt Use

Figure 3 shows the number of casualties by injury severity and the ratio by seatbelt use in the vehicle-to-vehicle accidents and the single vehicle accidents. Overall, the ratio of belted occupants in the vehicle-to-vehicle accidents was higher than that in the single vehicle accidents, for both drivers and front passengers. The ratio of belted occupants was trend on decrease as the injury severity increased, for both vehicle-to-vehicle accidents and single vehicle accidents. For fatalities, the belted drivers were about one-half of the vehicle-to-vehicle accidents and about one-third of the single vehicle accidents. However, the belted drivers accounted for over 80 % of serious injuries and slight injuries respectively, for both vehicle-to-vehicle accidents and single vehicle accidents.

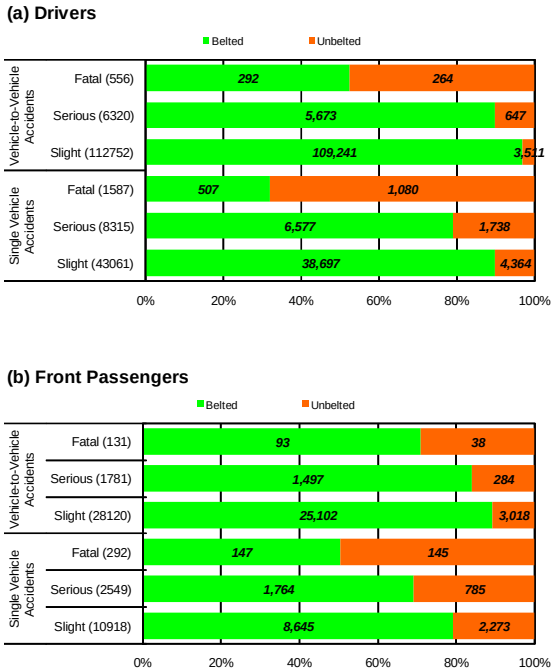


Figure 3. Ratio of casualties by seatbelt use and injury severity

Gender and Age Groups

Figures 4 and 5 show the distributions of casualties of the belted drivers and front passengers by injury severity, gender, and age groups in the vehicle-to-vehicle accidents and the single vehicle accidents. Age groups were classified in three groups of under 25 year olds, 26-64 year olds, and over 65 year olds. With regard to the belted drivers of both vehicle-to-vehicle accidents and single vehicle accidents, irrespective of injury severity, the number of 26-64 year old males was larger, however, the ratio of over 65 year old males increased as the injury severity. With regard to the belted front passengers of both vehicle-to-vehicle accidents and single vehicle accidents, the ratio of over 65 year old females increased as the injury severity.

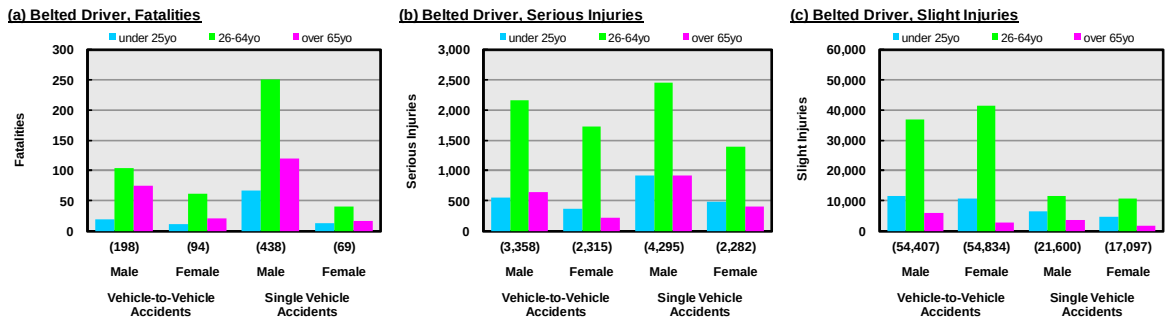


Figure 4. Belted driver casualties by gender and age groups

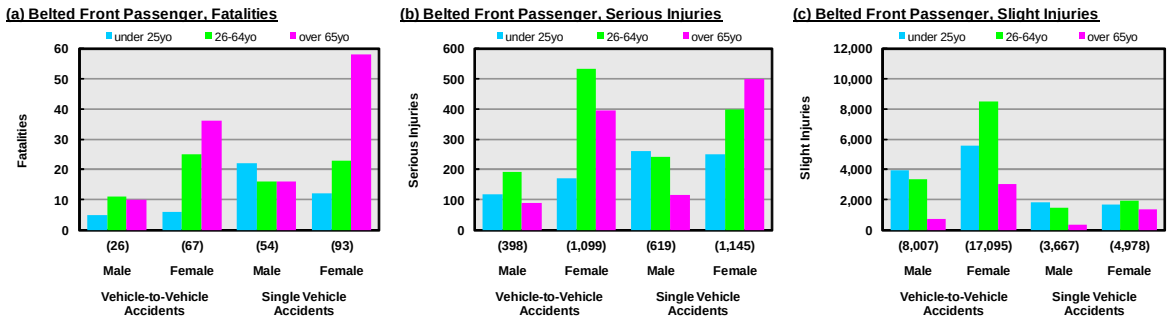


Figure 5. Belted front passenger casualties by gender and age groups

BELTED DRIVER'S INJURIES IN FRONTAL CRASH VEHICLES

In view of the above-mentioned finding that drivers account for about 80 % of front-seat casualties in front-damaged vehicles from both vehicle-to-vehicle accidents and single vehicle accidents, furthermore based on the purpose of this study, hereafter, the analysis was focused on the belted drivers. The fatalities, serious injuries, and slight injuries in the analysis were respectively the sum of vehicle-to-vehicle accidents and single vehicle accidents, and except analysis of five years trend, the cumulative total numbers in five years from 2004 to 2008 was used.

Five Years Trend by Injured Body Region

For the injured body region, the NPA traffic accident statistics generally records the most highest injured body region in traffic accidents. The statistics classify in eleven body regions: head, face, neck, thorax, abdomen, back, pelvis, arms, legs, overall body, and other. In this analysis, nine body regions excluding overall body and other were used as "injured body region" for analysis.

Figures 6 and 7 show the five years trend from 2004 to 2008 for the numbers of fatalities and serious injuries by the injured body region. For the fatalities, the head, thorax, and abdomen were frequent as the injured body region, throughout the period. The fatalities due to the head and thorax injuries since 2005 tended to decrease conspicuously, whereas the fatalities due to abdominal injuries were almost constant except for a slight increasing in 2008 from 2007. For the serious injuries, although the thorax and legs were frequent as the injured body regions most times, both trended to decline throughout the five year period, especially the legs were conspicuous.

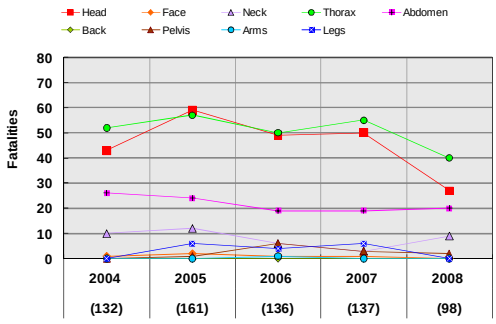


Figure 6. Five years trend of fatalities by injured body region (belted driver)

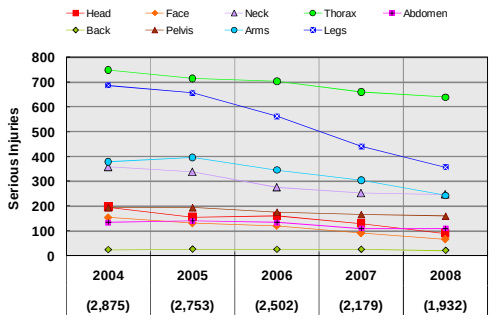


Figure 7. Five years trend of serious injuries by injured body region (belted driver)

Casualties and Injury Rates by Injured Body Region

Figures 8 to 10 show the numbers of belted driver fatalities, serious injuries, and slight injuries by injured body region, and fatality rate and serious+ injury rate by injured body region are shown in Figures 11 and 12. A fatality rate represents the number of fatalities divided by the total number of fatalities, serious injuries, and slight injuries, as in Equation (1). A serious+ injury rate represents the sum of fatalities and serious injuries divided by the total number of fatalities, serious injuries, and slight injuries, as in Equation (2).

$$Fatality\ Rate = \frac{Fatalities}{\sum(Fatalities, Serious\ Injuries, Slight\ Injuries)} \times 100(\%) \cdots (1).$$

$$Serious+Injury\ Rate = \frac{\sum(Fatalities, Serious\ Injuries)}{\sum(Fatalities, Serious\ Injuries, Slight\ Injuries)} \times 100(\%) \cdots (2).$$

As shown in Figures 8 to 10, for the fatalities, the thorax was most frequent as the injured body region, followed by the head and the abdomen. With regard to the serious injuries, the thorax was most frequent as the injured body region, however, there were fewer head injuries and more leg injuries, as compared to the fatalities. In the slight injuries, the neck injuries were most numerous among the injured body regions, and it accounted for about 60 % of slight injuries. Both fatality rates and serious+ injury rates shown in Figures 11 and 12 indicated the highest rate on the abdomen respectively; the fatality rate of abdomen was 2.2 times higher than the head, and the serious+ injury rate of abdomen was 1.9 and 1.4 times higher than the thorax and legs, respectively.

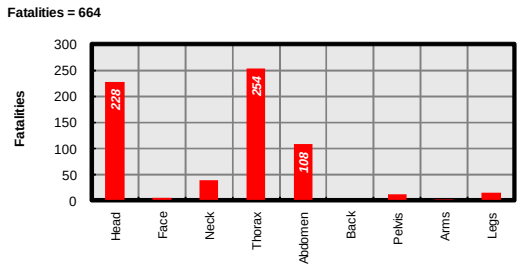


Figure 8. Fatalities by injured body region

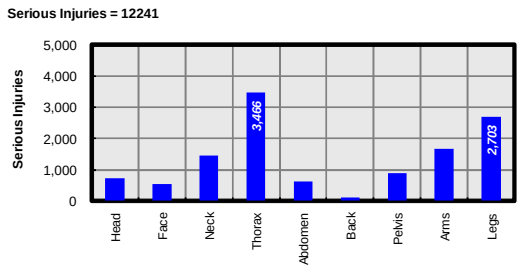


Figure 9. Serious injuries by injured body region

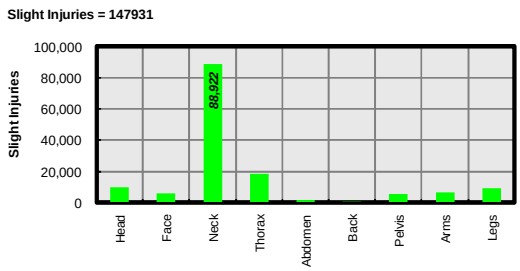


Figure 10. Slight injuries by injured body region

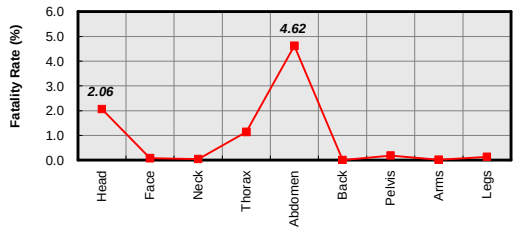


Figure 11. Fatality rate by injured body region

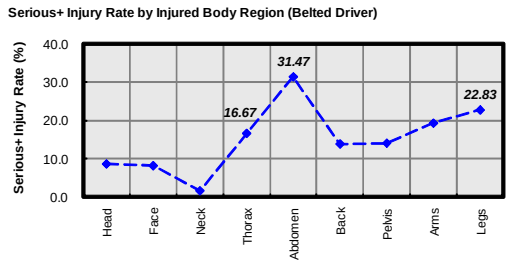


Figure 12. Serious+ injury rate by injured body region

Injury Circumstances by Estimated-delta V

The travel speed before a driver perceives a danger and takes avoidance procedure is recorded in the NPA traffic accident statistics, and this speed is named "travel speed just before the accident". ITARDA calculates the amount of vehicle speed variation in crash by using this speed and vehicle kerb weight, and defines it as "Estimated-delta V". The Estimated-delta V of a frontal crash vehicle in a vehicle-to-vehicle accident is calculated by Equation (3), (4) or (5) according to the damaged area of the partner vehicle (see Figure 1)[4]. In this study, calculated Estimated-delta V is classified in the following four groups, 0-20 km/h, 21-40 km/h, 41-60 km/h, and over 61 km/h, and the trend variation of injured drivers due to the change of Estimated-delta V was investigated. Estimated- delta V for single vehicle accidents is identical with the "travel speed just before the driver perceives a danger".

- Collision partner's damaged area ; Front, Right Front or Left Front

  $$Estimated-delta\,V = M2/(M1+M2)\times(V1+V2)\cdots\cdots(3).$$

- Collision partner's damaged area ; Rear, Right Rear or Left Rear

  $$Estimated-delta\ V = M2/(M1+M2)\times(V1-V2)\cdots\cdots(4).$$

- Collision partner's damaged area ; Right or Left

  $$Estimated-delta\ V = M2/(M1+M2)\times V1\cdots\cdots\cdots\cdots(5).$$

Where,

M1 ; Mass of frontal crash vehicle

M2 ; Mass of collision partner

V1 ; Travel speed of frontal crash vehicle just before accident

V2 ; Travel speed of collision partner just before accident

Fatalities and injuries by Estimated-delta V

Figure 13 shows the numbers of fatalities, serious injuries and slight injuries by Estimated-delta V. The fatalities were frequent in 41-60 km/h and over 61 km/h, and these speed ranges accounted for about 80 % of fatalities. On the other hand, many serious injuries were distributed in 21-40 km/h and 41-60 km/h, and these speed ranges accounted for about 70 % of serious injuries. With regard to the slight injuries, the speed ranges of 0-20 km/h and 21-40 km/h accounted for about 80 % of slight injuries.

Injured body region by Estimated-delta V

Figures 14 to 16 show the distributions of injured body regions by Estimated-delta V for fatalities, serious injuries, and slight injuries. With regard to the injured body region on fatalities, the percentages of thorax, head, and abdomen were higher in all speed ranges, and the percentage of head increased as the delta V rose. For the serious injuries, the thorax accounted for about 20 % to 30 % in all speed ranges. Though the percentage of neck was higher in the low speed range, that percentage became lower while the percentage of the legs became higher as the delta V rose. For the slight injuries, the percentage of neck was the highest and accounted for 40 % to 70 % in all speed ranges.

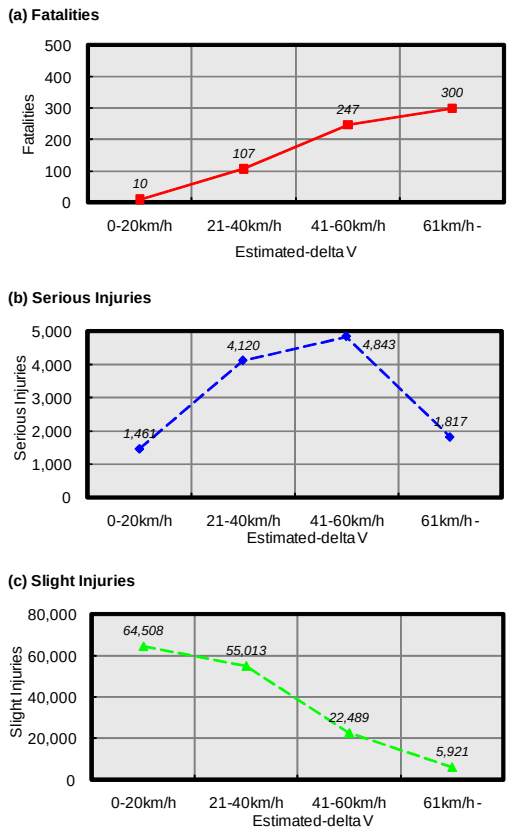


Figure 13. Fatalities and injuries by Estimated-delta V

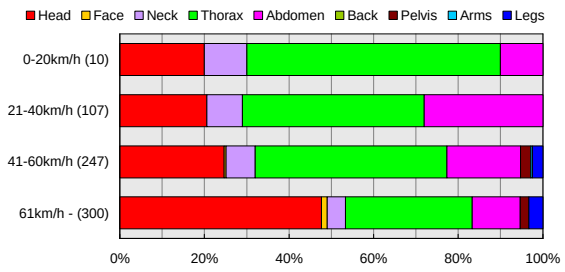


Figure 14. Distribution of injured body region of fatalities by Estimated-delta V

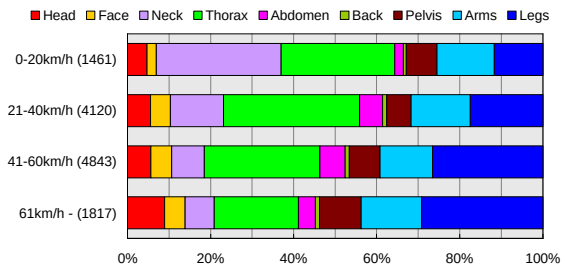


Figure 15. Distribution of injured body region of serious injuries by Estimated-delta V

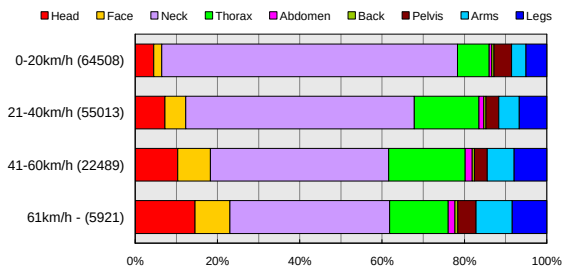


Figure 16. Distribution of injured body region of slight injuries by Estimated-delta V

Injured Body Region and Injury Type by Gender and Age Groups

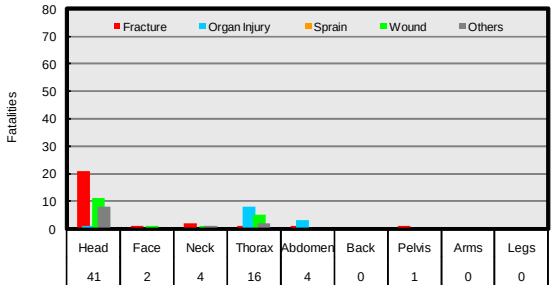
In this section, the trend with regard to the injured body regions and injury types was investigated by gender and age groups. Of the injury types defined in the statistics, five types were applied in this study: fracture, organ injury, sprain, wound (including abrasion, laceration, contusion, etc.), and others. As in previous section, three age groups were applied: under 25 year olds, 26-64 year olds, and over 65 year olds.

Injured body region and injury type of male drivers Figures 17 and 18 show the injured body regions and injury types with regard to the fatalities and serious injuries of male drivers by age groups. Though the number of fatalities greatly differed among the three age groups, a fracture or a wound of head, and an organ injury of thorax or abdomen were frequent in under 25 year olds and 26-64 year olds. In over 65 year olds, a fracture, a wound or an organ injury of thorax, and an organ injury of abdomen were outnumbered than the head injuries. In the case of serious injuries, though a fracture was the leading injury type in all the age groups, the leg fracture were most frequent in under 25 year olds, and both thorax and leg fractures were equally frequent in 26-64 year olds. Contrary to this, the thoracic fracture was most frequent in over 65 year olds.

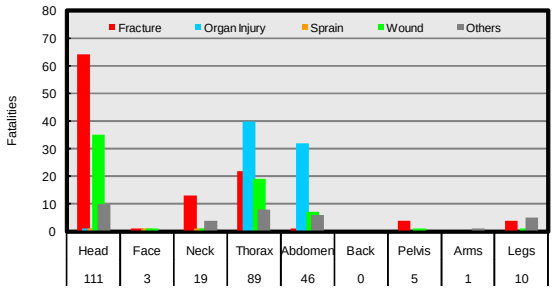
Injured body region and injury type of female drivers Figures 19 and 20 show the injured body regions and injury types with regard to the fatalities and serious injuries of female drivers by age groups. Though it is difficult to understand a trend because the fatalities of female drivers were fewer than male drivers, the fatalities of female drivers differed from male drivers in that an organ injury of thorax outnumbered a fracture and a wound of head in 26-64 year olds. Similar to male drivers, many serious injuries of female drivers were injured due to a fracture in all the three age groups. However, it differed from male drivers in that a fracture of thorax outnumbered a fracture of legs in 26-64 year olds.

Fatality rate by injured body region, gender and age groups Figure 21 shows the fatality rates of belted drivers by injured body regions, gender, and age groups. For both male and female drivers, the fatality rate of head was the highest in under 25 year olds, however, the fatality rate of abdomen was the highest in 26-64 year olds and over 65 year olds. In the case of male drivers, the fatality rates of abdomen in 26-64 year olds and over 65 year olds were respectively three times and six times higher than the fatality rate of head in under 25 year olds.

(a) Belted Male Driver, under 25yo



(b) Belted Male Driver, 26-64yo



(c) Belted Male Driver, over 65yo

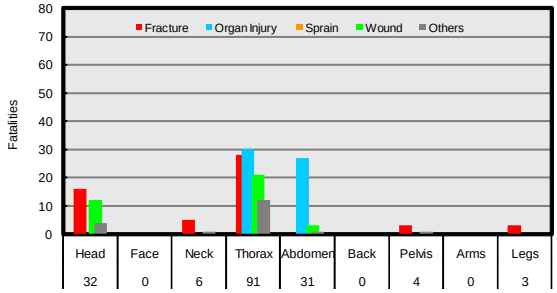
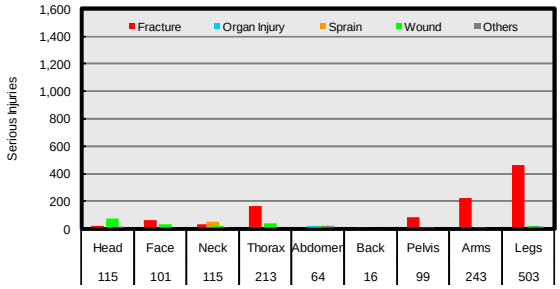
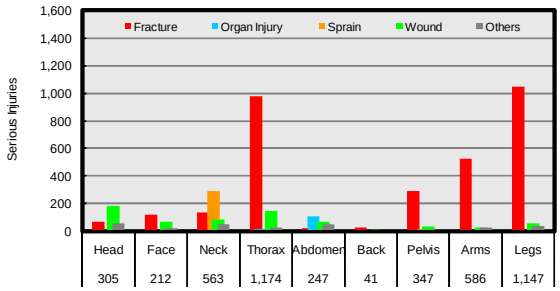


Figure 17. Injured body region and injury type by age groups for fatalities of male drivers

(a) Belted Male Driver, under 25yo



(b) Belted Male Driver, 26-64yo



(c) Belted Male Driver, over 65yo

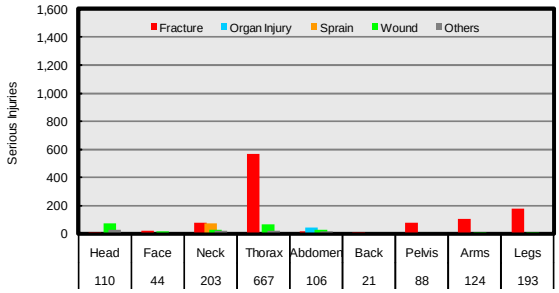


Figure 18. Injured body region and injury type by age groups for serious injuries of male drivers

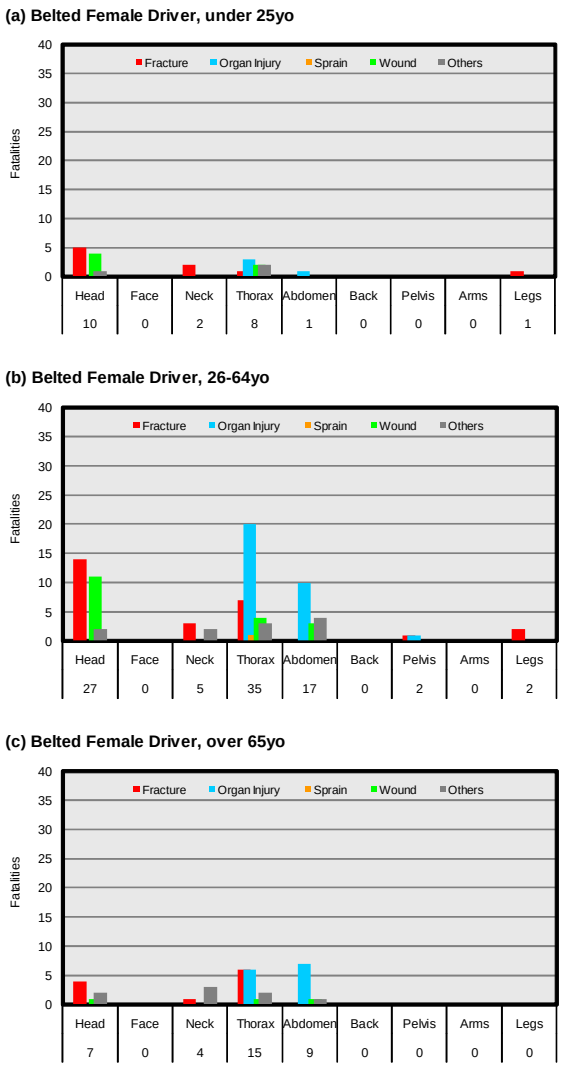


Figure 19. Injured body region and injury type by age groups for fatalities of female drivers

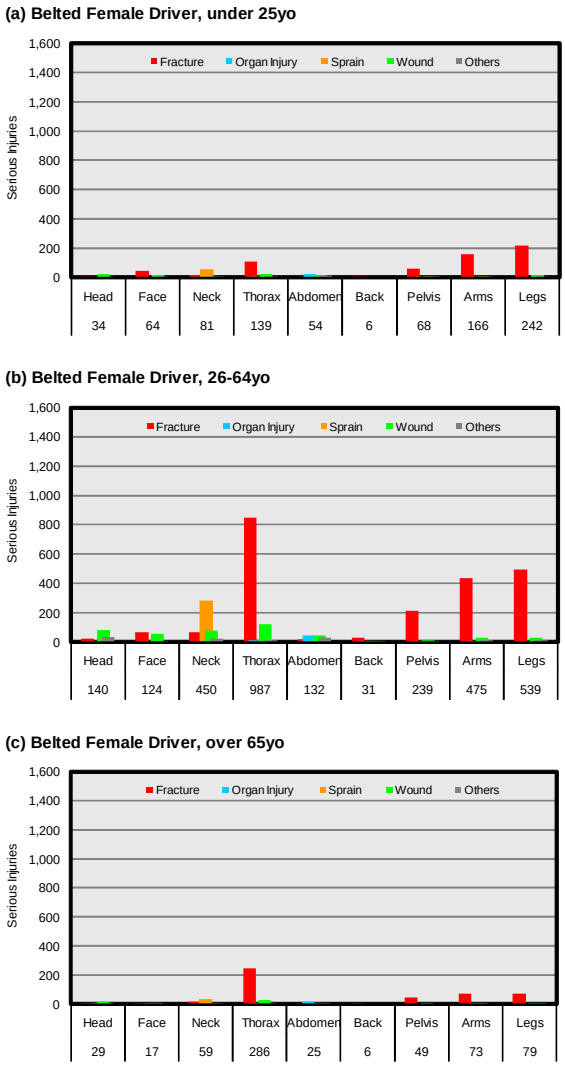


Figure 20. Injured body region and injury type by age groups for serious injuries of female drivers

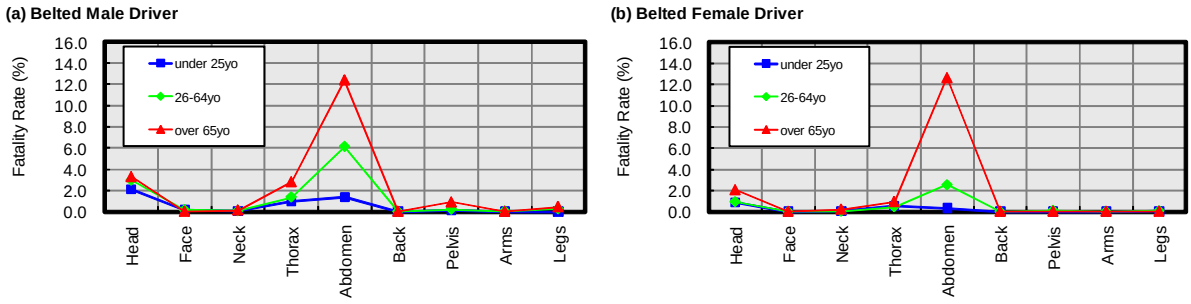


Figure 21. Fatality rate of belted drivers (male and female) by injured body region and age groups

Correlation of Injured Body Region and Injury Source

This section shows the results that analyzed the correlations between the injured body regions of belted drivers and the injury sources. It was not classified by gender and age groups.

Figures 22 and 23 show the correlations between injured body regions and injury sources for the cases of fatality caused by an organ injury and a fracture.

The fatalities caused by an organ injury had almost injury to the thorax or abdomen, and the injury source was the steering wheel in most of these cases. Many fracture fatalities had injury to the thorax and head. While the injury source to the thorax was the steering wheel in many cases, the injury source to head had steering wheel, windshield, other interior, etc.

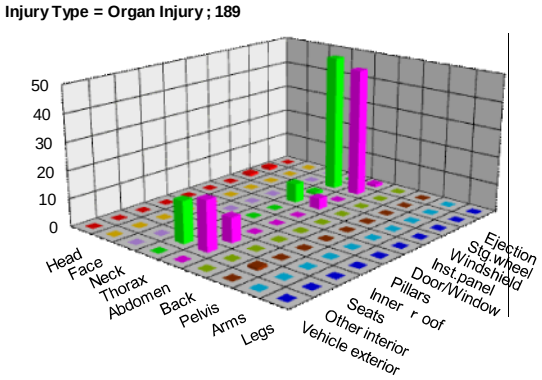


Figure 22. Correlation of injured body region and injury source of fatalities (Organ Injury)

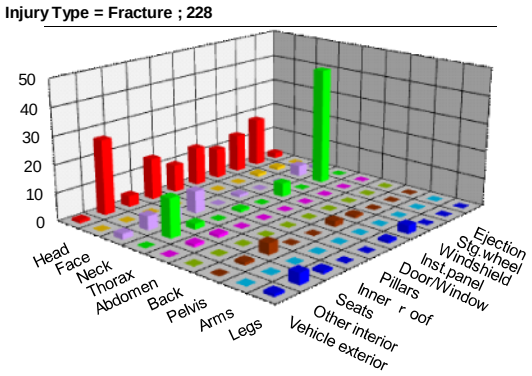


Figure 23. Correlation of injured body region and injury source of fatalities (Fracture)

Figure 24 shows the correlation of injured body regions and injury sources for serious injuries due to a fracture. The most frequent combination proved to be thorax and steering wheel, followed by the combination of legs and other interior.

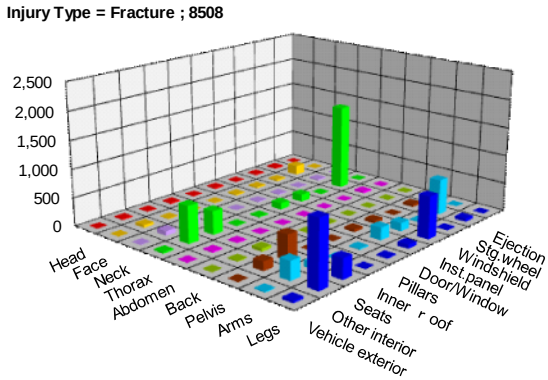


Figure 24. Correlation of injured body region and injury source of serious injuries (Fracture)

SUMMARY

Authors investigated and analyzed the injured body region and injury type for occupants in vehicles damaged on front from recent traffic accident statistics towards the future discussion regarding what human body regions should be evaluated in the frontal crash test, and what injury parameters should be measured utilizing the dummy. The findings were as follows:

General Output deduced by Accident Data Analysis

- The drivers accounted for about 80 % of front seat occupants for all of fatalities, serious injuries, and slight injuries.
- As the severity of injury rose, the ratio of seatbelt use decreased for both drivers and front passengers.
- Belted drivers accounted for one-half of fatalities in vehicle-to-vehicle accidents and about 30 % of fatalities in single vehicle accidents. However, the ratios of belted drivers exceed 80 % of both serious and slight injuries.

Belted Driver's Injuries in Frontal Crash Vehicles

- The number of fatalities caused by head or thorax injuries was on the decrease since 2005, but fatalities due to abdominal injury leveled off in the past five years. Regarding serious injuries, thorax and leg injuries were on the decrease since 2005, especially the legs were conspicuous.
- For the fatalities, the thorax, head and abdomen

were frequently injured body regions, and the thorax and leg injuries were more numerous among injured body regions in the serious injuries. For the slight injuries, neck injuries accounted for about 60 %. However, as for fatality rates and serious+ injury rates, the highest rates were respectively recorded by abdominal injuries.

- About 80 % of fatalities were distributed in Estimated-delta V of 41-60 km/h and over 61 km/h. Estimated-delta V of 21-40 km/h and 41-60 km/h accounted for about 70 % of serious injuries. 0-20 km/h and 21-40 km/h accounted for about 80 % of slight injuries.
- As Estimated-delta V rose to middle and high speed levels, the percentage of head injuries increased among fatalities, similarly, the percentage of leg injuries expanded among serious injuries.
- Male driver's fatalities were frequent for the fracture and wound of head in under 25 year olds and 26-64 year olds, furthermore, the organ injuries of thorax or abdomen were also frequent. However, in over 65 year olds, the fracture, wound and organ injury of thorax and the organ injury of abdomen outnumbered the fracture and wound of head.
- Male driver's serious injuries were most frequent for leg fracture in under 25 year olds. In 26-64 year olds, the fractures of thorax and legs were equally frequent. With regard to over 65 year olds, thoracic fracture was most frequent. Thus, the overall trend was that the percentage of thoracic fracture increased as male drivers grew older.
- Female driver's fatalities differed from male drivers in that the organ injury of thorax outnumbered the fracture and wound of head in 26-64 year olds.
- Female driver's serious injuries were similar to male drivers in that fracture was frequent in all the age groups, however, differed from male drivers in that the thoracic injury outnumbered the leg injury in 26-64 year olds and over 65 year olds.
- Regarding fatality rates by injured body region, a rate of head was the highest in under 25 year olds whereas a rate of abdomen was the highest in 26-64 year olds and over 65 year olds, for both male and female drivers.
- Fatalities due to organ injury were numerous in the thorax and abdomen as injured body region, and the injury source was almost the steering wheel.
- Fatalities caused by fracture were frequent in the thorax and head as injured body region. While the steering wheel proved to be the main injury source to thoracic fracture, the injury source to skull fracture had the steering wheel, windshield, and other interior, etc.
- With regard to serious injuries of thorax fracture caused by the steering wheel was most frequent, followed by the serious injuries of legs caused by other interior.

CONCLUSIONS

With respect to fatalities and serious injuries by injured body region, the casualties casued by the head, thorax, and leg injuries outnumbered the casualties due to abdominal injury. However, the abdominal injury was the highest for the fatality rates and the serious+ injury rates. In five years from 2004 to 2008, fatalities casued by head or thoracic injuries were trend on decrease, while fatalities due to abdominal injury leveled off. These results suggest that in order to further improve the occupant protection performance during frontal crash, the abdominal injury that is impossible to evaluate in the Hybrid III and the injury measurement capability of the abdomen are particularly requested for a future frontal dummy (see APPENDIX).

FUTURE WORK

This paper has reported the results of general analysis from traffic accident statistics regarding the injury situations that focused on the front seat occupants, especially the drivers, of frontal crash vehicles. It will be necessary to consider the influence of airbag (its activation or non-activation) on the injury severity towards the future discussion regarding what human body regions should be evaluated in the frontal crash

test, and what injury parameters should be measured utilizing the dummy. Consequently, more studies will be needed to compare injury status between airbag deplyment and non-deplyment cases. Furthermore, because the detailed information of the human injury pattern was not included in the NPA accident statistics, it was impossible to investigate which specific organ was injured well as the mechanism involved. Therefore, it will be necessary to investigate and analyze the human injury mechanism in detail, based on the new establishment of the integration with the Medical Trauma Registry Database. These studies provide a basis for global discussion regarding the development (including harmonization of specifications) of the advanced frontal dummy by clarifying the vehicle accident situations in Japan which includes many relatively smaller cars than that in the United States and in Europe.

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[4] Shimamura, M. et al.: Method to evaluate the effect of safety belt use by rear seat passengers on the injury severity of front seat occupants, Accident Analysis and Prevention Vol.37, issue1, p5-17, 2005

APPENDIX

Table A1. Comparison of the instrumentations of the THOR-NT and the Hybrid-III

	THOR-NT(Current Ver.)	Hybrid-III(Typical Configuration)
Head	9 Uniaxial Accelerometers	**Yes** (1 Triaxial Accelerometer at Head C.G.)
	1 Biaxial Tilt Sensor	No
Face	Five Uniaxial Load Cells	No
Neck	Upper Neck Load Cell (6 channels)	**Yes**
	Lower Neck Load Cell (6 channels)	**Yes**
	Front Neck Cable Load Cell	No
	Rear Neck Cable Load Cell	No
	Head Rotation Potentiometer	No
Thorax	CRUX Deflection Units - 3 Dimensional Displacement at each of Four Locations (UL, UR, LL, LR)- 4 CRUX units @ 3 channels each;	**Yes** (One-directional Displacement String Potentiometer)
	1 Triaxial Accelerometer at the C.G.	**Yes**
Mid Sternum	1 Uniaxial Accelerometer	No
Upper Abdomen	Uni-directional Displacement String Potentiometer	No
	Uniaxial Accelerometer	No
Lower Abdomen	DGSP Deflection Units - 3 Dimensional Displacement at L & R Locations (2 DGSP units @ 3 channels each)	No
Spine	1 Triaxial Accelerometer at T1 location	No
	1 Triaxial Accelerometer at T12 location	No
	T12 Load Cell (5 channels)	**Yes**
	4 Biaxial Tilt Sensors	No
Pelvis	Acetabulum Load Cell (left and right, 3 channels each)	No
	Iliac Crest Load Cells (left and right, 1 channel each)	No
	1 Triaxial Accelerometer at Pelvis C.G.	**Yes**
Femur	Femur Load Cell (left and right, 6 channels each)	**Yes**
Knee	Knee Shear Displacement, L&R	**Yes**
Lower Extremity	Upper Tibia Load Cell (left and right, 4 channels each)	**Yes**
	Lower Tibia Load Cell (left and right, 5 channels each)	**Yes**
	Tibia Acceleration (left and right, 2channels each)	No
	Achilles Tendon Load Cell (left and right, 1 channels each)	No
	Ankle Joint Rotation Potentiometers (left and right, 3channels each)	No
	Foot Acceleration (left and right, 3 channels each)	No