RESTRAINED HYBRID III DUMMY-BASED CRITERIA FOR THORACIC HARD-TISSUE INJURY PREDICTION

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ABSTRACT

Ninety-three sled tests (60 cadaver, 33 dummy) is used to assess the Hybrid III dummy and associated thoracic injury criteria to determine their ability to predict cadaver injury in matched impacts over a range of impact and restraint conditions. A statistical analysis is performed to evaluate the injury-predictive efficacy of the dummy-based maximum chest deflection, maximum chest acceleration, and CTI. A comparison of the selectivity and specificity of injury prediction models with each injury criterion reveals that the maximum dummy chest deflection is the best predictor of injury (C statistic = 0.8514) when used in an injury risk model including cadaver age, gender, and mass. Consideration of the maximum chest acceleration, either as an independent covariate or as part of CTI, is found to weaken the injury model. The restraint condition (belt-dominated loading, airbag-dominated loading, or combined loading), however, is found to be a more important injury predictor than any injury criterion.

Key Words: Injury criteria, cadavers, thorax, frontal impacts, restraint systems

THE DESIGN OF OCCUPANT RESTRAINT SYSTEMS is dictated to a large extent by the requirement to reduce thoracic injury risk. Restraint systems are assessed using a variety of tools. Human volunteer, cadaver, or animal tests may be performed to evaluate safety systems. These types of tests are expensive, laborious, and have limited repeatability, however; so anthropomorphic test devices (ATDs), or dummies, are often used as surrogates for humans. ATDs are instrumented to measure various mechanical parameters, including sternal displacement relative to the spine (chest deflection), acceleration measured at the chest center of gravity (cg), and others. Mechanical parameters, or combinations thereof, are correlated with the presence of injury and used as predictors of injury risk. Used in this manner, these mechanical parameters are referred to as injury criteria. Injury criteria are then often used in injury risk functions, which can incorporate other parameters affecting injury risk, such as occupant age (e.g., Funk et al. 2001). In order to be useful for injury prediction, an injury criterion or risk function must satisfy at least two requirements: it must differentiate injurious loading conditions from non-injurious loading conditions, and it must do this for the set of loading conditions that span the range of interest (including future advances in restraint system technology).

BACKGROUND – FACTORS AFFECTING THORACIC INJURY PREDICTION

The likelihood that a restrained (belt, airbag, or both) occupant will sustain a thoracic injury in a frontal impact depends on the characteristics of the collision (e.g., speed, occupant seating position), the characteristics of the restraint system (e.g., presence of a belt, presence of an airbag), and the characteristics of the occupant (e.g., size, age, physical condition). Characterization of injury risk in a

research or development test generally relies upon dummy-based mechanical parameters to represent the effect of the collision and restraint system characteristics. The purpose of this paper is to determine how well the Hybrid III 50th percentile male with a standard instrumentation package predicts cadaver hard tissue injury in a matched collision when the collision, restraint, and occupant characteristics are varied. A background discussion of these characteristics is therefore necessary.

EFFECT OF RESTRAINT CONDITION AND INJURY CRITERION: Thoracic injury criteria have been evaluated using a variety of loading conditions. Due to the low rates of seatbelt use in the 1960s and 1970s, early studies of thoracic injury and impact response focused on loading experienced by an unbelted subject (see Kroell 1994 for a review of thoracic impact studies performed prior to 1980). Characteristics of the thorax were developed using blunt objects representing a steering wheel hub or instrument panel, and loading rates were representative of those experienced by unrestrained occupants in severe collisions (e.g., Nahum et al. 1970, Kroell et al. 1971, Mertz and Gadd 1971, Kroell et al. 1974). As seatbelt use rates increased, it became necessary to evaluate thoracic response and injury criteria under localized belt impingement and at lower loading rates (e.g., Cromack and Ziperman 1975, Fayon et al. 1975, Mertz et al. 1991, Horsch et al. 1991, Crandall et al. 1994).

As airbag restraints became more common in the vehicle fleet, researchers recognized the need for a thoracic injury criterion appropriate for airbag loading and for combined belt-and-airbag loading. Recently, the National Highway Traffic Safety Administration (NHTSA) proposed the combined thoracic criterion CTI for the evaluation of diverse restraint conditions (Eppinger et al. 1999). CTI results from the finding that a two-parameter logistic regression model using the cadaver-based T1 peak acceleration and the maximum chest deflection as covariates correlated well with the probability of an AIS 3 or higher injury (defined by the number and distribution of rib fractures sustained by the cadaver). CTI is a sum of normalized maximum chest cg acceleration and normalized maximum chest deflection:

$$CTI = A_{max}/A_{int} + C_{max}/C_{int} \qquad [1]$$

where A_{int} and C_{int} are normalization constants defined for various dummy sizes.

EFFECT OF OCCUPANT CHARACTERISTICS (SIZE, GENDER, AGE): In addition to restraint condition, several characteristics of the occupant can affect the specific injury tolerance and the efficacy of an injury criterion for that occupant. Hard tissue injury tolerance is generally found to decrease with age and (for some types of loading) to increase with occupant size. Independently of size, females are often assumed to have lower hard tissue injury tolerance.

Effect of Occupant Age: Stein and Granik (1976) performed bending tests on three ribs from each of 79 human donors having an age range from 27 years to 83 years. They found a strong inverse correlation between breaking force and donor age at death ($p<0.001$). Those authors concluded that, like long bones, ribs apparently undergo progressive circumendosteal resorption with advancing age; but, unlike long bones, ribs show no evidence of continued subperiosteal apposition. This results in a general decrease in the percent of the rib cross-section that is cortical bone.

From a search of the available literature, it appears that the most comprehensive study of the effect of occupant age on thoracic hard tissue impact injury tolerance was presented by Zhou et al. (1996). They found that the reduction in thoracic injury tolerance for a senescent group (age 66-85) compared to a baseline younger group (age 16-35) was strongly dependent on the restraint condition: for belt loading, tolerance decreased by 72% while blunt load tolerance decreased by only 21% (frontal) or 27% (lateral).

Effect of Occupant Gender and Occupant Size: Studies often find females to have intrinsically weaker bones than males (e.g., Wertheim 1847, Rauber 1876, Burghele and Schuller 1970), but this finding is not universal. For example, Lindahl and Lindgren (1967) found no significant gender difference in ultimate tensile strength, mean deformation at failure, limit of proportionality, or the modulus of elasticity of human cortical bone. Yamada (1970) also found no significant gender differences in tensile strength or percent elongation in cortical bone. The data for compressive, bending, and torsional loading of cortical bone have been similarly mixed with respect to the effect of gender, as have the data for human cancellous bone (e.g., Messerer 1880, Evans and King 1961, Sonoda 1962, Chalmers and Weaver 1966).

Studies of gender influences on bone mineral density (BMD) for males and pre-menopausal females have likewise been mixed (e.g., Parsons et al. 1996, Slosman et al. 1994, Kelly et al. 1990). Henry and Eastell (2000) compared peak bone mineral density in male and female adults ages 20-37. Uncorrected dual-energy X-ray absorptiometry (DEXA) measures of femoral neck bone indicated higher BMD values for male subjects. After correcting for differences in skeletal size, however, these authors found no significant difference in femoral neck bone mineral apparent density (BMAD, an estimated volumetric bone density which attempts to normalize bone mineral density measurements for bone size) between male and female subjects. Lumbar spinal vertebra BMAD was actually significantly ($p < 0.0001$) greater for the female subjects in this pre-menopausal population. Studies have also evaluated the effect of bone apposition on the observed gender differences in fracture rates. Through the use of biochemical markers of bone formation, researchers have found greater bone formation in men (Resch et al. 1994, Vanderschueren et al. 1990, Duda et al. 1988) and in women (Epstein et al. 1994).

Based on these inconsistent findings, hard tissue injury tolerance differences observed between male and female specimens may be due to factors other than intrinsic differences in material strength; non-senescent males and females may not have material differences in hard tissue injury tolerance. As females age, however, the release of menopausal hormones does result in decreased bone mineral density compared to males, so age-dependent decreases in hard tissue injury tolerance are more pronounced for women than for men.

The effect of occupant size on thoracic injury tolerance to frontal impact restraint loading is not clear. Larger occupants have larger bony structures, and thus are capable of bearing greater forces than smaller occupants. In the case of non-inertial loading, the increased structural strength of larger occupants is manifest as an occupant size-dependent injury tolerance. In the case of non-force-limiting restraint loading in a frontal impact, however, the magnitude of force applied to the occupant is a monotonically increasing function of the occupant's thoracic mass. The compelling question is whether bony structural strength increases with occupant size at a greater or lesser rate than the inertial forces the structures must bear. At present, the authors know of no study that evaluates this relationship for the case of concentrated and distributed loading of the thorax during a frontal impact.

The estimation of injury risk requires an understanding of how all of the factors discussed above affect injury potential and also how they affect the predictive ability of an injury criterion. A statistical analysis of cadaver test data is a potential way to reach this understanding since regression techniques allow for the analysis of factors individually as well as factor interactions. Given the large number of factors that must be considered, however, the dataset must have certain characteristics. First, it must include cadavers that span the age and mass ranges of interest and both genders must be represented. Second, a range of restraint conditions must be considered so that the effect of load distribution can be evaluated. In addition, because thoracic injury data are censored, each age subset, mass subset, gender subset, and restraint subset must contain both injury and non-injury tests.

EXISTING CADAVER DATA

The existing data on cadaver frontal impact loading can be divided into three main groups: blunt impactor tests (e.g., Kroell 1994), seatbelt table tests (e.g., Cesari and Bouquet 1994), and sled tests (see Kuppa and Eppinger 1998 for a summary and for additional references). The blunt impactor tests were not designed to evaluate the effect of belt, airbag, and combined belt-and-airbag loading. In addition, the blunt impactor test setup appears to differ from a vehicle environment sufficiently to preclude the evaluation of combined thoracic injury criteria like CTI. The seatbelt table tests, while useful for evaluating the case of a concentrated load, does not contain any distributed-load tests for comparison. Further, the seatbelt table setup, which includes a reaction surface at the spine, precludes evaluation of any chest acceleration-dependent criterion.

The existing database of sled tests, therefore, has the greatest potential to elucidate the effects discussed in the Background section. The database contains numerous subjects over wide age and mass ranges. In addition, belt-dominated loading, airbag-dominated loading, and combined restraint loading test conditions are included. The sled-test database is, however, also subject to limitations. As discussed by Prasad (1999), Hassan and Nusholtz (2000), and Kent et al. (2000), these limitations include artifactual kinematic measurements that occurred when the cadaver's head struck the

windshield or windshield header, resulting in large acceleration peaks transmitted through the spine to the cadaver's thorax. This phenomenon was observed for tests performed with the airbag-only restraint conditions, but was not observed for belted occupants. As a result, for the purpose of a statistical analysis of restraint condition effects, these tests introduce a systematic error into the database.

In addition, the instrumentation used in the cadaver sled tests introduced systematic errors into the database. Accelerometers were used to measured acceleration at several locations on the cadavers, including the first thoracic vertebra (T1). In most of the cadaver tests, the T1 acceleration was considered to be analogous to, or at least representative of, the chest cg acceleration. Subsequent cadaver tests, which included accelerometers at both T1 and at the eighth thoracic vertebra (T8), revealed that, for some conditions, the T1 acceleration does not necessarily represent the chest cg acceleration (Shaw et al. 2001). Because T1 acceleration magnitude is a reasonable representation of chest cg acceleration magnitude for some conditions but not for others, the use of T1 acceleration is a systematic error in the development of dummy-based injury criteria.

The use of chestbands is another source of uncertainty. As discussed by Bass et al. (2000), Shaw et al. (1999) and Hagedorn and Burton (1993), chestbands are subject to errors under the localized loading caused by seatbelts. The chestbands are, however, considered to be accurate indicators of thoracic deformations under the distributed loading caused by an airbag. This restraint-dependent measurement accuracy is another potential source of systematic error in the database.

REFINEMENT AND EXPANSION OF EXISTING SLED TEST DATABASE

Despite these limitations, the existing cadaver sled test database does contain the most diverse and detailed experiments available for the evaluation of injury criteria efficacy, so means should be sought that will allow the use of these data. The largest source of systematic error, the use of cadaver-mounted instrumentation, can be reduced through the use of matched Hybrid III dummy tests (performed under identical sets of circumstances – sled buck, restraint system, sled acceleration pulse). The use of a Hybrid III also contains inherent limitations. For example, while the dummy has biofidelity for mid-sternal blunt impacts, it has been shown to be too stiff statically (Lau and Viano 1988, Horsch et al. 1990) and at loading rates observed in belt tests (Backaitis and St-Laurent 1986, Cesari and Bouquest 1990). From a pragmatic standpoint, however, regardless of dummy biofidelity, a correlation between dummy measures and injury outcome is desired since the dummy is used to evaluate vehicle restraint design in research tests as well as in compliance tests and new car assessment tests. If a criterion can be identified that allows the existing dummy to predict injury risk for restraint and impact conditions that span the range of conditions in which the dummy is to be used, then the dummy's lack of biofidelity decreases in importance.

In addition to the improved repeatability and reliability of dummy measures compared to cadaver-based measures, this dummy-based approach has an additional advantage: several of the cadaver tests that had been identified by Prasad (1999) as having erroneous cadaver-based chest deflection measurements can be reincorporated into the dataset, since the only cadaver-based data of interest are the anthropometry, gender, age, and injury outcome.

EXISTING CADAVER SLED TEST DATABASE AND NEW CORRESPONDING DUMMY TESTS: The database of cadaver sled tests presented by Kuppa and Eppinger (1998), was used as a starting point. For each set of conditions tested using a cadaver at the University of Virginia (UVA), one or more corresponding 50th percentile male Hybrid III tests were performed with a nominally identical sled buck, sled acceleration pulse, and restraint condition. Standard Hybrid III instrumentation was used, including chest cg acceleration and mid-sternal chest deflection.

Analysis of high-speed film and instrumentation signals was used to identify several tests that, for a variety of reasons, were unacceptable for this study. The tests UVA103, UVA250, and UVA259, recommended for removal by Prasad, were included back into the dataset since the allegedly erroneous cadaver-based measurements were not used. Two other tests (UVA96 and UVA97), also recommended for removal by Prasad, were not reincorporated into the dataset because both the dummy and cadaver exhibited significant head strikes on the windshield header, which generated artifactual acceleration spikes and also resulted in a significant portion of the subject's momentum being arrested through the head/neck complex rather than through the thorax and restraint system.

Unacceptable tests also included those that had a loss of dummy instrumentation or those that exhibited significant differences between dummy kinematics and cadaver kinematics. In some tests, differences in spinal and thoracic compliance resulted in the cadaver loading the steering wheel heavily through the airbag while the dummy did not. In these cases, the tests were excluded from analysis because the kinematics were sufficiently different to cast doubt on the dummy's ability to represent cadaver injury potential. In three cadaver tests (UVA356-UVA358, airbag, no belt), the unbelted cadaver sustained rib fractures from steering wheel contact through the airbag. These tests were included in the dataset, however, since the dummy exhibited a similar interaction with the steering wheel. Except for the cases identified by Prasad and the cases of obviously artifactual chest acceleration peaks, film analyses of both the existing sled tests and the new sled tests were blinded. In other words, the restraint system and measured dummy responses were not used to target tests for film analysis. The comparison of dummy kinematics and cadaver kinematics was the only factor used to determine inclusion.

After the performance of dummy tests to match all UVA cadaver tests and the removal of several sets of tests for the reasons discussed above, the dataset of UVA tests was pared to 33 cadaver tests, each having a corresponding dummy test. The literature was searched to find additional cadaver tests having 50th percentile male Hybrid III dummies tested in identical conditions. Five such cadaver tests, presented by Yoganandan et al. (1991) of the Medical College of Wisconsin (MCW), were added to the UVA database. The resulting dataset of existing cadaver tests contains 38 cadaver tests, all of which have corresponding Hybrid III tests, and all of which exhibit similar kinematics between cadaver and dummy. In many cases, more than one cadaver was tested at a given set of conditions. In these cases, often only a single dummy was tested at that condition. Details of the existing cadaver tests and the corresponding dummy tests are presented in Table 1.

NEW CADAVER TESTS AND CORRESPONDING DUMMY TESTS: To expand the scope of the existing database, an additional 18 cadaver tests and 11 corresponding Hybrid III tests were performed at UVA and 4 cadaver tests with 3 corresponding Hybrid III tests were performed at the University of Heidelberg (UH) (Table 2). Seven of the dummy tests and eleven of the cadaver tests performed at UVA involved subjects seated in the RFP (designated by a "P" in the test number) (Kent et al. 2001). All tests were performed in vehicle bucks representing contemporary mid-sized sedans. Sled deceleration pulses were based on full-vehicle barrier tests.

FINAL DATABASE: The database of new cadaver tests and dummy tests was combined with the database of existing tests. This final database contains 60 cadaver tests, all of which have corresponding Hybrid III dummy tests, for a total of 93 sled tests. Restraint conditions include two-point shoulder belt systems, three-point lap-and-shoulder (L/S) belt systems, airbag-only systems, standard L/S-and-airbag systems, and force-limiting (F/L) belt-and-airbag systems. All airbags are of the head-and-torso type as opposed to the smaller bags common in Europe. For the purpose of evaluation, these restraints were divided into three groups, based on the thoracic loading (Table 3):

1. **Restraint A** – belt-dominated loading (all cases with a torso belt and without an airbag – this includes both two-point shoulder belt systems and three-point systems),
2. **Restraint B** – airbag-dominated loading (this includes cases with an airbag and without a torso belt), and
3. **Restraint C** – combined loading (this includes all cases with an airbag and a shoulder belt, regardless of whether the belt was force-limiting or standard).

Both male (n = 43) and female (n = 17) cadavers are included in the final database, the age-at-death range is 24 years to 72 years (mean 55, standard deviation 12.1), and the mass range is 47 kg to 117 kg (mean 71, standard deviation 15.8).

The test conditions for the new tests were chosen in an attempt to fill voids in the existing database (Table 3 – bold boxes indicate deficient areas in the dataset). Specifically, RFP tests, combined loading tests, and non-injurious tests were the focus. There are, however, still deficiencies. First, more non-injurious tests are needed, especially with female subjects and the belt-only restraint condition. Even with the new tests aimed at non-injurious loading, there are only 16 non-injury tests in the database. Only two of these 16 tests involved female subjects, and only two involved a belt-dominated restraint condition.

Table 1 –Database of Acceptable Existing Cadaver Tests with Corresponding Dummy Tests

Test No.	ID	Age	Gender	Weight (kg)	Height (cm)	Sled ΔV (km/h)	Restraint	No. of Rib Fx	Thoracic MAIS	Dummy Peak Chest Accel. (g)	Dummy Peak Slider Defl. (mm)	Dummy CTI
Tests Performed at UVA												
52	**Hybrid III**					34.8	2-pt			33	50	0.85
53	91-W-03	61	F	61	153	34.9	2-pt	19	4			
54	**Hybrid III**					38.1	2-pt			35	74	1.11
55	91-EF-10	62	F	91	176	36.9	2-pt	11	3			
101	**Hybrid III**					33.8	2-pt			30	57	0.88
102	92-UM-23	60	M	95	176	33.1	2-pt	19	4			
103	92-EM-21	57	M	103	179	32.5	2-pt	13	4			
104	92-EF-25	66	F	105	179	32.3	2-pt	11	4			
112	**Hybrid III**					46.8	2-pt			36	75	1.13
113	92-EF-22	24	F	57	159	47.3	2-pt	12	4			
114	92-FF-24	60	F	65	164	47.0	2-pt	27	4			
172	**Hybrid III**					24.6	L/S			25	56	0.82
173	92-EM-33	61	M	72	167	24.6	L/S	9	3			
174	93-EF-35	57	F	62	168	25.9	L/S	12	3			
175	93-EM-27	58	M	117	185	25.7	L/S	3	2			
222	**Hybrid III**					53.4	2-pt			45	64	1.12
223	93-EM-32	51	M	61	169	55.0	2-pt	13	4			
224	93-EM-26	58	M	65	175	54.4	2-pt	16	4			
225	93-EM-37	36	M	68	177	53.9	2-pt	16	4			
226	**Hybrid III**					53.3	2-pt			41	64	1.08
227	93-FM-29	53	M	70	165	53.5	2-pt	12	3			
228	92-FM-18	47	M	85	177	54.7	2-pt	16	4			
229	93-EM-36	37	M	60	183	54.0	2-pt	17	4			
249	**Hybrid III**					54.1	2-pt			45	57	1.05
250	94-EM-39	39	M	50	177	54.9	2-pt	12	4			
251	**Hybrid III**					58.9	2-pt			52	59	1.15
252	94-EM-38	37	M	67	177	58.2	2-pt	17	4			
256	**Hybrid III**					56.5	2-pt			46	52	1.02
257	94-EM-40	33	M	112	179	56.7	2-pt	10	4			
258	94-EM-43	69	M	64	178	55.3	2-pt	14	4			
259	94-EF-41	64	F	77	163	56.4	2-pt	15	4			
293	**Hybrid III**					56.8	L/S			47	37	0.88
294	94-EF-44	68	F	55	148	56.8	L/S	10	4			
295	94-EM-42	57	M	104	187	56.8	L/S	15	4			
296	94-EM-45	59	M	74	181	59.8	L/S	26	4			
302	**Hybrid III**					57.5	L/S and AB			51	31	0.86
303	93-EM-34	64	M	50	154	57.5	L/S and AB	4	2			
304	94-EM-47	65	M	57	168	59.4	L/S and AB	15	4			
305	94-EF-48	66	F	58	161	59.4	L/S and AB	12	4			
332	**Hybrid III**					58.2	F/L and AB			37	30	0.70
333	95-EM-51	51	M			58.2	F/L and AB	6	3			
334	94-EM-49	49	M			58.2	F/L and AB	5	3			
335	95-EM-50	50	M			58.6	F/L and AB	2	2			
355	**Hybrid III**					55.7	AB only			46	58	1.08
356	95-EM-52	64	M	73.9	176	57.2	AB only	30	4			
357	96-EM-55	48	M			57.2	AB only	19	4			
358	96-EM-56	40	M			59.0	AB only	17	4			
Tests Published by Yoganandan et al. (1991)												
1	**Hybrid III**					50.8	L/S			51	34	0.90
M1	Cadaver	58	M	82	180	49.0	L/S	10	4			
2	**Hybrid III**					49.7	L/S			44	39	0.87
D2	Cadaver	57	M	73	178	49.0	L/S	10	4			
3	**Hybrid III**					50.0	L/S			43	40	0.87
V3	Cadaver	66	M	77	178	50.0	L/S	7	3			
4	**Hybrid III**					51.8	L/S			40	29	0.73
F4	Cadaver	58	M	70	178	50.0	L/S	14	4			
5	**Hybrid III**					49.0	L/S			39	27	0.70
A5	Cadaver	67	M	73	175	50.0	L/S	12	3			

Restraint Codes: 2-pt – shoulder belt only, L/S – lap and shoulder belt, F/L – force-limiting L/S, AB – airbag.

Table 2 – Database of New Cadaver Sled Tests with Corresponding Dummy Tests

Test No.	ID	Age	Gender	Weight (kg)	Height (cm)	Sled ΔV (km/h)	Restraint	No. of Rib Fx	Thoracic MAIS	Dummy Peak Chest Accel. (g)	Dummy Peak Slider Defl. (mm)	Dummy CTI
Tests Performed at UVA												
410	**Hybrid III**					57.0	F/L and AB			32	25	0.60
411	96-EM-57	60	M	Unk.	Unk.	57.5	F/L and AB	19	4			
412	96-EM-60	70	M	Unk.	Unk.	56.8	F/L and AB	14	4			
532	**Hybrid III**					48.6	F/L and AB			27	27	0.56
537	**Hybrid III**					48.9	F/L and AB			27	29	0.58
538	**Hybrid III**					48.1	F/L and AB			25	26	0.53
533	99-EF-104	67	F	64	163	48.6	F/L and AB	1	1			
534	97-EM-76	47	M	51	170	48.4	F/L and AB	4	3			
535	98-EF-95	57	F	53	163	48.1	F/L and AB	16	4			
544	97-EF-83	59	F	56	168	49.2	F/L and AB	9	4			
545	99-EM-103	67	M	74	176	48.1	F/L and AB	3	2			
571P	**Hybrid III**					47.6	F/L and AB			26	29	0.57
572P	**Hybrid III**					48.1	F/L and AB			28	28	0.58
576P	**Hybrid III**					48.1	F/L and AB			28	30	0.60
577P	99-FM-111	57	M	70	174	47.4	F/L and AB	0	0			
578P	99-FF-107	69	F	52.5	155	47.6	F/L and AB	4	3			
579P	99-FF-106	72	F	59	156	47.6	F/L and AB	11	4			
580P	99-FF-105	57	F	57	177	47.6	F/L and AB	0	0			
648P	**Hybrid III**					48.6	AB only			44	19	0.67
649P	**Hybrid III**					47.6	AB only			44	20	0.68
650P	00-FRM-124	40	M	47	150	48.9	AB only	4	2			
651P	00-FRM-121	70	M	57	176	48.6	AB only	0	0			
652P	00-FRM-118	46	M	74	175	49.7	AB only	0	0			
663P	**Hybrid III**					48.0	L/S and AB			45	40	0.89
664P	**Hybrid III**					48.0	L/S and AB			43	40	0.87
665P	99-FRM-112	55	M	85.3	176	48.0	L/S and AB	3	2			
666P	99-FRM-115	69	M	83.9	176	48.0	L/S and AB	3	2			
667P	00-FRF-120	59	F	79.4	161	48.0	L/S and AB	13	4			
668P	00-FRF-127	54	F	55.3	162	48.0	L/S and AB	23	4			
Tests Performed at the University of Heidelberg												
9001D	**Hybrid III**					47.0	L/S			43	22	0.69
9002D	**Hybrid III**					48.0	L/S			45	28	0.77
9013C	Cadaver	34	M	71	180	48.0	L/S	0	1			
9003D	**Hybrid III**					48.0	AB only			57	13	0.76
9014C	Cadaver	31	M	70	170	47.0	AB only	0	0			
9207C	Cadaver	25	M	74	184	49.0	AB only	0	0			
9212C	Cadaver	38	M	79	174	47.0	AB only	0	0			

Note: "P" in the test number represents tests performed with subjects in the right-front passenger position.

Another area for additional testing is the airbag-only restraint condition. Most of the original Kuppa and Eppinger (1998) tests that had to be removed from the database were performed with this restraint condition. This is a result of the relatively uncontrolled occupant motions that occur when a belt restraint is not used. Based on film analysis of the rejected tests, the dummy often does not represent the cadaver kinematics well in the absence of a belt and, as a result, many of these tests could not be used for the present study. The final database contains only nine airbag-only tests.

The database is also inadequate in the number of young cadavers. While there are obvious difficulties with obtaining young cadavers, age is known to be a significant predictor of injury risk. Additionally, the population of cadavers that have been tested under combined loading has a younger mean age and a lesser mean mass than the populations tested under the other two restraint conditions. These differences are not significant (≤ approximately one standard deviation), but may introduce bias into the restraint-dependency of the dummy measures.

Finally, the ratios of male-to-female cadavers are not equal among the three restraint conditions. In fact, there are no females tested in the airbag-only configuration. While the gender ratio of occupants receiving AIS 3+ chest injuries in the field is on the order of 1/1, the ratio in the sled test database is 43 males to 17 females.

Table 3 – Summary of Cadaver Database

	Existing database	New tests	Final Database
Total cadavers with a matching Hybrid III test	38	22	60
Number of these cadavers with AIS 3+ injury	35	9	44
Number of these cadavers w/o AIS 3+ injury	3	13	16
Tests with belt-dominated thoracic loading	29	1	30
Belt-dominated with AIS 3+ injury	28	0	28
Belt-dominated w/o AIS 3+ injury	1	1	2
Tests with airbag-dominated thoracic loading	3	6	9
Airbag-dominated with AIS 3+ injury	3	0	3
Airbag-dominated w/o AIS 3+ injury	0	6	6
Tests with combined thoracic loading	6	15	21
Combined with AIS 3+ injury	4	9	13
Combined w/o AIS 3+ injury	2	6	8
Females	9	8	17
Females with AIS 3+ injury	9	6	15
Females w/o AIS 3+ injury	0	2	2
Males	29	14	43
Males with AIS 3+ injury	26	3	29
Males w/o AIS 3+ injury	3	11	14
Under 56	15	9	24
Under 56 with AIS 3+ injury	14	2	16
Under 56 w/o AIS 3+ injury	1	7	8
Over 55	23	13	36
Over 55 with AIS 3+ injury	21	7	28
Over 55 w/o AIS 3+ injury	2	6	8
Driver-side tests	33	11	44
Driver-side tests with AIS 3+ injury	30	5	35
Driver-side tests w/o AIS 3+ injury	3	6	9
Passenger-side tests	5	11	16
Passenger-side tests with AIS 3+ injury	5	4	9
Passenger-side tests w/o AIS 3+ injury	0	7	7
Average age - belt-dominated tests	54.5	34.0	53.8
Standard deviation (years)	11.7	0.0	12.1
Average age - airbag-dominated tests	50.7	41.7	44.7
Standard deviation (years)	12.2	15.7	14.5
Average age - combined-loading tests	57.5	61.3	60.2
Standard deviation (years)	8.3	7.3	7.6
Average mass (kg) - belt-dominated tests	76.4	71.0	76.2
Standard deviation (kg)	18.0	0.0	17.7
Average mass (kg) - airbag-dominated tests	78.3	66.8	70.7
Standard deviation (kg)	3.8	12.3	11.4
Average mass (kg) - combined-loading tests	62.3	65.0	64.0
Standard deviation (kg)	9.9	12.3	11.4
Male/female ratio in belt-dominated tests	21/8	1/0	22/8
Male/female ratio in airbag-dominated tests	3/0	6/0	9/0
Male/female ratio in combined-loading tests	5/1	7/8	12/9

STATISTICAL ANALYSIS OF SLED TESTS (FINAL DATABASE)

METHODS: The data were analyzed by the generalized estimating equation method (GEE) (Liang and Zeger 1986). The outcome was a binary response, with cadaver thoracic AIS scores of 3 or greater considered as injury (Y=1) and scores <3 considered as non-injury (Y=0). Dummy-based injury criteria (A_{max}, C_{max}, and CTI), as well as characteristics of the cadaver [age (years), gender, and body mass (kg)] were treated as the set of potential predictors for thoracic injury. The data were deemed insufficient for an analysis of any interaction terms. The binary outcome was modeled by a generalized linear model with a logit link function. Model coefficients, β_i, were estimated by maximum likelihood and the variance-covariance parameters were estimated by the Huber-White estimator (Huber 1967, White 1982). The dummy test identification number was treated as a clustering factor, since in several instances multiple cadaver tests were performed under the same test configuration that was utilized during a single dummy test.

As a modeling strategy we fit 6 GEE models. Three of the models were specified so that only one of the dummy-based injury criteria was included along with the predictors associated with the cadaver (age, gender, and body mass). Model 1 used C_{max}, model 2 used A_{max}, and model 3 used CTI. Model 4 used both C_{max} and A_{max}. A baseline model (model 5) was developed, which included only the predictors associated with the cadaver (i.e., no dummy measures were included). Model 6 included C_{max} as well as consideration of the restraint condition (A,B,C).

The relative ranking of the predictors was based on a standardized version of the Wald chi-square statistic. By subtracting from the observed Wald chi-square statistic the total number of degrees of freedom associated with the statistic, the standardized value represents the difference between the observed chi-square statistic and the expected chi-square statistic under the null hypothesis of no association.

The comparison of the overall performance of the models was based on an internal bootstrap model validation (Efron and Tibshirani 1993). The bootstrap validation is a re-sampling procedure that corrects for the optimism (bias) in the observed value of model performance indices. Optimism is induced into the estimate of the model's overall performance as a consequence of utilizing the data at hand to estimate the model parameters. The bootstrap-adjusted index is bias-corrected and provides a better estimate of the model's performance when applied to a new sample of data.

The performance measure of interest in this analysis is the C statistic (Harrell 2001). The C statistic is an overall measure of the model's sensitivity and specificity with respect to discriminating between the subjects who fall into the category Y=1 and those who fall into the category Y=0. The C statistic can be interpreted in the following manner. For any random pair of subjects, one of which has realized outcome Y=1 and the other which has realized outcome Y=0, we would expect with probability C that the model-based prediction for the subject with realized outcome Y=1 to be greater in magnitude than the model-based prediction for the subject with realized outcome Y=0. For a set of predictors that produce a model C statistic of 0.50, the utility of the model is no better than the flip of a fair coin, while perfect discrimination produces a C statistic of 1.0. A set of predictors that produces a C statistic of 0.70 or greater is considered to have utility as a predictive tool (Harrell 2001).

RESULTS: Model 1, which includes C_{max} as a predictor, is a significant model of injury probability ($p < 0.0001$) (Table 4) and C_{max} is the most important covariate in that model (Figure 1). In contrast, model two, which includes A_{max} as the dummy measure, is not a significant model ($p = 0.0978$) and A_{max} is not a significant predictor ($p = 0.68$) in the model, ranking least important of the four covariates. Model 3, which includes CTI, is a significant predictor of injury probability ($p = 0.0004$) and CTI is the most important covariate in the model. To understand the importance of each component of CTI, consider model 4, which includes C_{max} and A_{max} as separate predictors. This model is a significant predictor of injury ($p < 0.0001$), but A_{max} is not a significant covariate ($p = 0.22$) and, in terms of relative ranking of predictors, ranks fourth out of the five predictors used in the model.

Table 4 – Comparison of Sled Test Injury Prediction Models

a) Model 1 ($p < 0.0001$)			b) Model 2 ($p = 0.0978$)			c) Model 3 ($p = 0.0004$)		
	β_i	p		β_i	p		β_i	p
Intercept	-0.6683	0.86	Intercept	-0.1733	0.95	Intercept	-6.5546	0.10
Gender	-0.8265	0.25	Gender	-1.4781	0.02	Gender	-2.2333	0.01
Mass (kg)	-0.05114	0.19	Mass (kg)	0.01088	0.64	Mass (kg)	-0.01867	0.52
Age (years)	-0.0009334	0.98	Age (years)	0.01540	0.60	Age (years)	0.03691	0.33
C_{max}	0.1652	0.00	A_{max}	0.01687	0.68	CTI	10.7951	0.00
Model C statistic = 0.8514			Model C statistic = 0.5484			Model C statistic = 0.8292		

d) Model 4 ($p < 0.0001$)			e) Model 5 ($p = 0.2516$)		
	β_i	p		β_i	p
Intercept	-2.6097	0.55	Intercept	0.6891	0.78
Gender	-1.3215	0.07	Gender	-1.3554	0.05
Mass (kg)	-0.05321	0.19	Mass (kg)	0.00976	0.67
Age (years)	0.004544	0.91	Age (years)	0.01173	0.71
C_{max}	0.1713	0.00			
A_{max}	0.05367	0.22			
Model C statistic = 0.8479			Model C statistic = 0.5776		

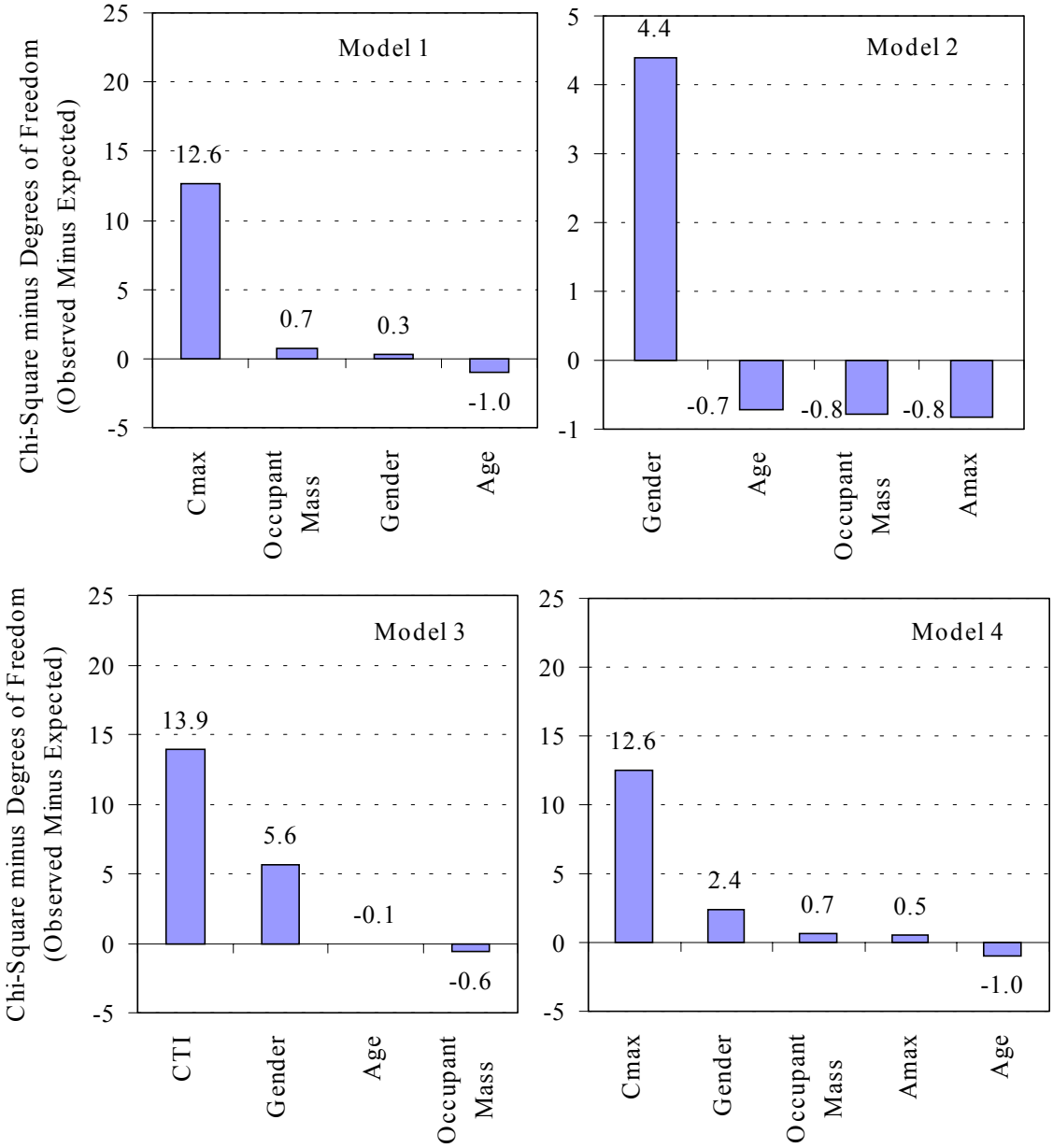


Figure 1 – Relative Ranking of Predictors Using Four Models of Sled Test Data

Since all models were identical except for the dummy criteria used as predictors, a comparison of models provides a means of comparing the efficacy of the various criteria. The C statistic is highest for model 1, indicating that, of the three criteria analyzed, C_{max} has the best injury sensitivity and specificity. The C statistic is lowest for model 2 (A_{max}) (Table 4). In fact, the addition of A_{max} in the model actually decreased the C statistic compared to model 5, which included no consideration of the dummy response. As expected based on this finding, the addition of A_{max}, either through the use of CTI (model 3) or the use of A_{max} as an additional covariate (model 4), decreased the C statistic relative to model 1. This lack of correlation between injury and A_{max} is illustrated in Figure 2. In this cross-plot, there is a clear relationship between C_{max} and the presence of 7 or more rib fractures. All tests with C_{max} below 25 mm resulted in fewer than 7 fractures. There is a transition zone between 25 mm and 40 mm in which some cadavers sustained greater than 7 fractures and some did not. Above 40 mm, all cadavers but one sustained greater than 7 fractures. A_{max} does not exhibit any

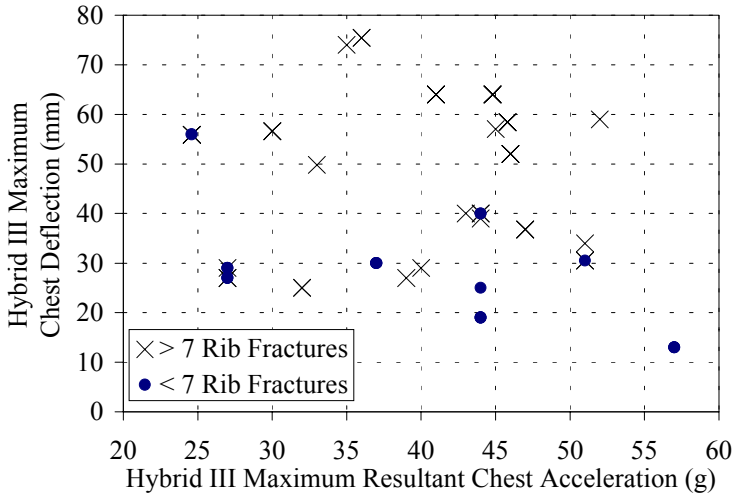


Figure 2 – Cross-plot of A_{max} and C_{max}, All Tests

relationship with the presence of 7 or more fractures. In fact, the test that resulted in the highest A_{max} (57g) resulted in no rib fractures. Figure 3 presents the test data and the resulting injury probability curves for C_{max}, A_{max}, and CTI.

The effect of the occupant characteristics is not clear. In model 1 and model 4, none of the occupant factors is a significant predictor of injury risk, though gender approaches significance ($p = 0.07$) in model 4. In model 2, gender is a significant covariate but, as mentioned above, the entire model is not a significant predictor of injury risk. In model 3, occupant gender is significant. The negative sign on the coefficient indicates that injury risk is higher for females ($x = 0$) than for males ($x = 1$). This trend is consistent for all models. Neither the occupant's mass nor the occupant's age is significant in any model, and there is no consistent trend among models (i.e., the sign of the coefficient is not consistent).

DISCUSSION

Our finding that C_{max} is a better rib fracture predictor than A_{max} for diverse types of restraint loading is in agreement with previous researchers. Well-distributed forces on the thorax can be much higher than concentrated forces (and thus result in much higher accelerations) without causing fractures. CTI results in no improvement over C_{max} because the acceleration term is poorly correlated with injury. Our findings confirm that rib fractures are a deformation-dependent injury and that C_{max} is a reasonably good predictor of rib fractures, even for diverse restraint conditions.

We have not, however, shown that the injurious level of C_{max} is constant regardless of restraint type. The data are insufficient to determine whether changes in the restraint type and therefore the force distribution on the thorax can change, for example, the value of C_{max} that corresponds to a 50% probability of AIS 3+ injury. We have demonstrated that C_{max} has reasonably good predictive ability for an entire dataset containing tests with diverse restraint conditions. In another paper (Kent et al. 2001), we showed that the injurious level of C_{max}, compared to other injury criteria, is relatively insensitive to the restraint condition. There is some sensitivity, however, so it is likely that a single, mid-sternal chest deflection measurement is insufficient to predict injury equally well for all restraint conditions. Advanced dummies, such as the Test device for Human Occupant Restraint (THOR), capable of measuring chest deformation at numerous locations, may allow for the identification of generally-applicable injury criteria that are not dependent on the restraint condition. The current measurement capabilities of the Hybrid III with its standard instrumentation package may not be sufficient for interpreting injury risk without consideration of the restraint. In support of this statement, consider another GEE model (model 6). In this model, the restraint condition (A,B,C) and the occupant's seating position (driver = 1, RFP = 0) are included as covariates in addition to C_{max} and the occupant characteristics age, gender, and mass. Despite the fact that C_{max} was the best of the three dummy criteria evaluated, the restraint condition is the dominant covariate in terms of injury prediction (Figure 4) and the C statistic (0.8789) is greater than the C statistic for model 1, indicating that consideration of the restraint improves model sensitivity and specificity. In this dataset, regardless of the dummy measurements, the belt-only restraint condition generated the greatest risk of injury, followed by the combined restraint condition. Due to the lack of concentrated belt loading, the airbag-only restraint condition was the least injurious.

The fact that no occupant characteristic was found to be a clear predictor of injury risk is an indication of the database's limitations. Of the three characteristics analyzed (gender, age, mass), age should be the strongest predictor of injury risk. Injury tolerance reductions with age are well documented. In this dataset, however, gender emerged as the most important occupant characteristic; it was the only characteristic to exhibit a consistent trend for all models and was the only characteristic to be a significant injury predictor in any model. Once gender-specific differences in age and size are accounted for, however, gender *per se* does not have a strong biomechanical basis as an indicator of hard tissue injury tolerance for a non-senescent population. The mean age of the subjects used in this study is over 50 years, however, so the observed gender effect is a reflection of the age-dependent differences in bone mineral density reduction.

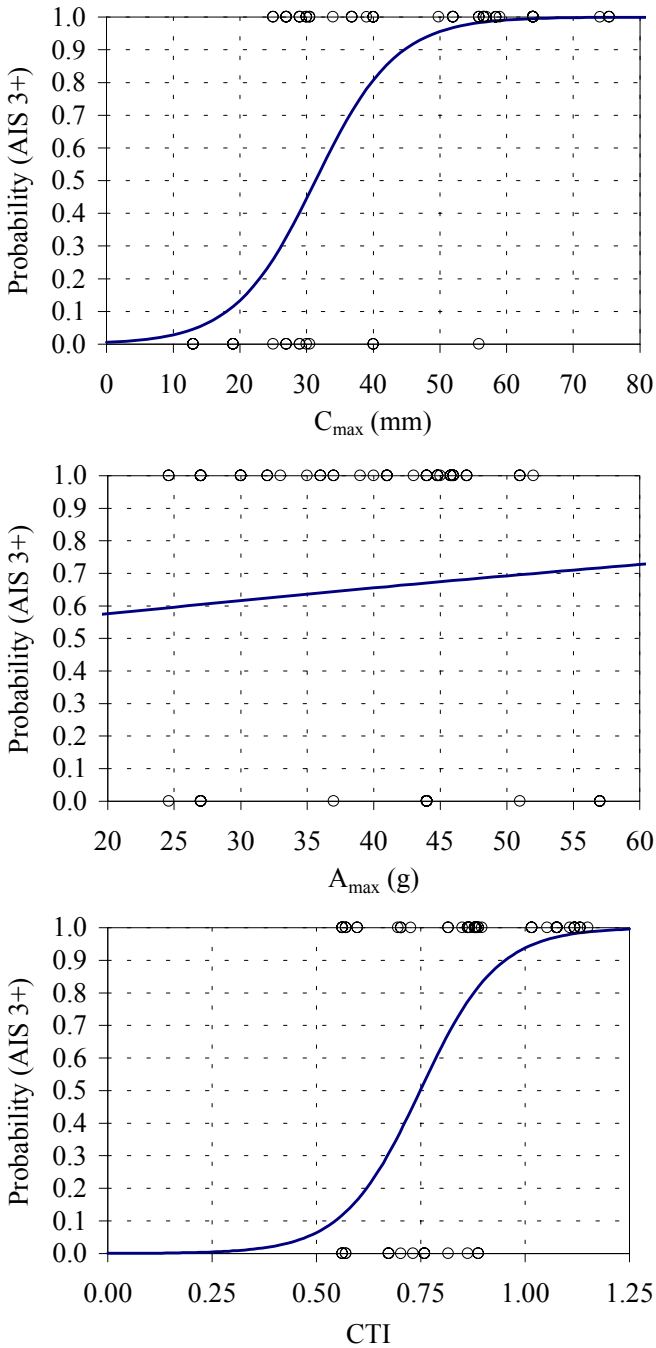
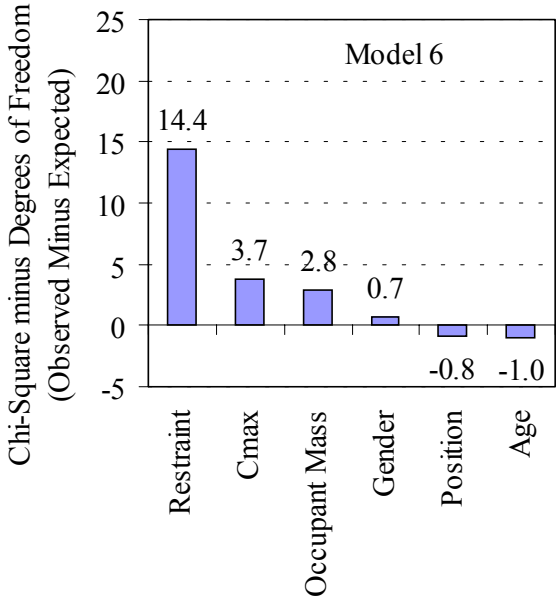


Figure 3 – Sled Test Model Results Showing Cadaver Injury Probability as a Function of Dummy-Measured C_{max}, CTI, and A_{max} for a 55 year-old, 71 kg, male (dataset mean age and mass)

The injury probability functions developed here are not applicable to a population of living humans because cadavers are generally more frail than the living, even for subjects of the same age. At present, we are aware of no studies that present a methodology for adjusting cadaver-based injury risk for application to the general population when diverse restraint conditions are considered. As a result, rather than predictors of living human injury risk, the statistical models presented here are useful primarily for evaluating the efficacy of the various dummy criteria and, to a lesser extent, for evaluating how critical values of the criteria change as a function of the occupant's age, gender, and mass. As a follow-up to this study, we will attempt to correlate dummy measures with injury risk to a population of living humans and in that way quantify the difference between living-human and cadaver injury risk.

In the tests presented here, the seatbelt was often the cause of rib fractures. This restraint dependence does not mimic the field experience of living humans. In the field, the addition of a belt restraint, rather than increasing thoracic injury risk, often increases occupant containment and precludes injurious strikes against hard vehicle components. Occupants in the field are also generally younger and less frail than cadavers used in sled tests and are thus less susceptible to rib fractures from belt loading (e.g., Martinez et al. 1994). The cadavers, then, may be representative of a population of elderly living humans, but exhibit different restraint-dependent injury trends than the general population of occupants. While it is important to consider the increased rib fracture risk for the elderly and the

increased morbidity and mortality associated with rib fractures in the elderly, the use of cadavers to generate thoracic injury criteria may result in over-prediction of the injury-causing potential of belt loading and thereby result in sub-optimal apportioning of belt and airbag loading for the whole population.



Model 6		
	β_i	p
Intercept	5.3670	0.45
Gender	-1.1060	0.19
Mass (kg)	-0.0953	0.05
Age (years)	-0.00157	0.98
C_{max}	0.1454	0.03
Restraint = B	-9.7397	0.00
Restraint = C	-2.8709	0.02
Position	0.3346	0.66
Model C statistic = 0.8789		

Figure 4 – Inclusion of Restraint and Seating Position in Injury Prediction Model

The use of cadaver injury to represent living-human injury is therefore a limitation of this analysis. In this, as in any cadaver-based dataset, the dominant thoracic injury type is rib fractures. The presence of multiple rib fractures is often associated with significant intrathoracic, intra-abdominal, and extra-cavity trauma (e.g., Pattimore et al. 1992). These injuries, therefore, are good indicators of the magnitude of overall thoracic trauma. They do not, however, correlate with all types of thoracic injuries sustained by the living, particularly soft-tissue injuries. For example, Lee et al. (1997) found that there is no clinically relevant correlation between thoracic skeletal injuries and acute traumatic aortic tear. It may be, therefore, that even a complete database of cadaver tests would not indicate the same injury risk factors, including dummy-based injury criteria, as a field-based set of data. For the purpose of evaluating injury criteria efficacy under diverse types of loading, however, cadaver testing is the most viable method currently available. In addition, while cadavers do not sustain soft tissue injuries representative of *in vivo* subjects, cadaver hard tissue injury is reasonably representative of the hard tissue injuries sustained by a frail population of living humans (Walfisch et al. 1985). Since approximately 61% of all AIS 2+ chest injuries are hard tissue injuries (rib fractures), and the maximum AIS is defined by fractures for approximately 72% of the occupants receiving MAIS 2+ chest injuries (Crandall et al. 2000), cadavers may be a reasonable representation of the living human population once differences in age and general physical condition are taken into account.

Injury criteria developed to predict rib fracture risk are an essential part of any vehicle design, assessment, or compliance testing effort. As optimized restraints and dynamically adaptable restraints are developed, however, it will become necessary to quantify the differences between cadaver-based injury and living-human injury since the use of a cadaver, as a conservative limit for the living human, is insufficient for robust optimization.

The finding that chest acceleration does not provide hard tissue injury-predictive value should not be interpreted to mean that the occupant's acceleration should not be limited in a crash. The human body does have an acceleration tolerance limit, but the types of injuries sustained at accelerations above this limit are not necessarily hard tissue injuries. Loss of consciousness, hemorrhaging, and other *in vivo* injuries, which have been associated with acceleration, currently cannot be evaluated using a cadaver model.

CONCLUSIONS

This paper presents analyses of 93 human cadaver and 50th percentile male Hybrid III dummy sled tests. Additional analysis, including more cadaver tests, is needed before a complete understanding of thoracic injury criteria efficacy and restraint dependence can be achieved. Even with the inclusion of 18 new cadaver tests and corresponding dummy tests, the sled test database is missing critical data points. The most important areas for additional testing are non-injury tests, tests with female subjects, tests with occupants in the RFP position, and tests with younger subjects.

The methodology outlined here eliminates some of the problems associated with thoracic injury criteria development from cadaver sled tests. The use of dummy measures collected using identical test conditions is of particular utility since it removes the largest source of systematic error (cadaver-mounted instrumentation) associated with cadaver sled tests. With this approach, it is critical to evaluate, using film and instrumentation, the dummy and cadaver kinematics to ensure that, first, the thoracic measures are due to thoracic loading rather than an unrelated artifact (e.g., a head strike) and, second, that the dummy's kinematics are similar to the cadaver's.

Relative ranking of injury prediction model covariates indicates that the injury-predictive effect of the restraint condition dominates that of the maximum dummy chest deflection or of any other dummy measure. Regardless of the dummy measures, the cadavers in this sample were more likely to be injured with a belt-dominated restraint condition than with a combined-loading restraint condition or an airbag-dominated restraint condition.

When the restraint condition was not considered, C_{max} was the best injury predictor of the three criteria considered. Consideration of the maximum chest acceleration, either as an independent covariate or as part of CTI, was found to weaken the injury model. The model that included A_{max} and not C_{max} was slightly worse than a model that included consideration of no dummy measure. This study is insufficient to exclude A_{max} as an injury criterion, however, since soft tissue injuries, particularly aortic injuries, which may be related to chest acceleration, are not considered.

Important caveats for cadaver test interpretation are given. In the laboratory, rib fracture risk increases when a belt is added to the restraint system because the belt results in concentrated force on the chest. In the field, the use of a belt controls occupant kinematics and often prevents injury-causing contacts against interior components. The validity of using cadaver-based injury criteria and risk functions to evaluate injury risk for a population of living, younger, and more robust occupants is currently not known and quantification of the differences between the living and the dead is an important area for research.

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