Restraint Dependency in Hybrid III Rib Fracture Prediction: A Comparison of Hybrid III and Cadaver Chest Deflection Response in Restrained Frontal Sled Tests

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ABSTRACT

Background and Objective: *The relationship between rib fracture risk and Hybrid III (H3) chest deflection in a frontal crash is dependent on the type of restraint employed. In contrast, the prediction of thoracic injury based on cadaver chest deflection is less dependent of the type of restraint employed. To investigate this limitation in H3 thoracic injury prediction, this study investigates the relationship between H3 and cadaver chest deflection under loading from various restraints in matched frontal sled tests.*

Methods: *Chest deflection results were examined for matched H3 and cadaver frontal sled tests performed at UVA using four different test configurations: A) driver's side position, force-limited 3-point belt plus airbag restraint; B) passenger's side, force-limited 3-point belt plus airbag restraint; C) passenger's side, standard (not force-limited) 3-point belt plus air bag restraint; D) lap belt plus air bag restraint. All tests were performed with target ΔV's of 48 km/h. For each cadaver test, R_d was defined as the ratio of maximum cadaver chest deflection (measured externally by chest bands) to maximum H3 chest deflection (measured internally by the H3 sternum "slider") under matched test conditions.*

Results: *The average (and 95% confidence interval) R_d values for each test configuration were as follows: A) R_{dAVG} = 0.66±0.30; B) R_{dAVG} = 0.87±0.17; C) R_{dAVG} = 0.75±0.14; D) R_{dAVG} = 0.59±0.07.*

Discussion: *These analyses suggest that the relationship between H3 and cadaver maximum chest deflection is dependent on the type of restraint employed. This may result in the observed restraint dependence of H3 chest-deflection-based thoracic injury prediction. It is shown that the H3-based thoracic injury risk function may be transformed, approximately, into the restraint-independent cadaver-chest-deflection-based injury risk function using the ratios described above.*

INTRODUCTION

Injuries and fatalities from automobile collisions represent a major public health concern worldwide. In the U.S. alone, more than 41,000 vehicular fatalities occur every year (NCIPC 2001). Thoracic injury is a contributing factor in nearly 70% of these fatalities (Mulligan et al. 1994). In the effort to mitigate vehicular injuries, safety systems may be evaluated using a variety of tools including tests with human volunteers, tests with human cadavers, and tests with animals. Due to the expense, limited repeatability, and ethical limitations of such tests, anthropomorphic test devices (ATDs, crash test dummies) are often used as human surrogates. The development of ATDs represents a significant biomechanical challenge. ATDs are designed to simulate the kinematic and dynamic responses of a human occupant in a collision. They are equipped with internal instrumentation to measure multiple dynamic occupant responses (e.g. head acceleration, chest deflection) that are correlated to a corresponding risk of injury to a human occupant. Such correlations are termed injury risk functions (IRFs), and may include non-dummy based variables such as characteristics of the crash (e.g. speed) or characteristics of a human occupant (e.g. gender, age). To serve as valid tools for vehicular safety evaluation, ATDs must be able to predict the kinematic, dynamic, and injury responses of human occupants for the range of loading conditions of interest.

The Hybrid III ATD is the current standard for evaluating vehicle crashworthiness and restraint performance in frontal impacts (Figure 1). Due to low seat belt use rates in the 1960s and 1970s, early thoracic injury threshold studies focused on simulating unrestrained driver contact with the steering wheel or instrument panel (e.g. Nahum et al. 1970, Kroell et al. 1971, Mertz and Gadd 1971, Kroell et al. 1974) using cadaveric experiments with blunt hub impacts to the chest. The data from these experiments were used to develop thoracic injury thresholds based on chest deflection (magnitude of displacement of the sternum towards the spine), which were implemented in the development of the Hybrid III ATD (Neathery et al. 1975, Mertz 1984, Viano and Lau 1988) and are still used in vehicle and restraint evaluation standards today.

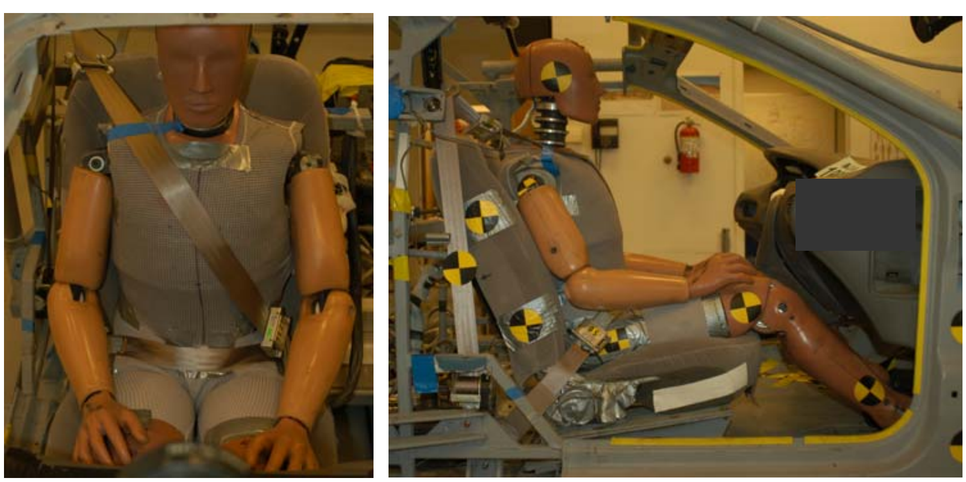

Figure 1: The Hybrid III frontal ATD positioned in a sled-type crash simulator.

The Hybrid III ATD is limited in its ability to evaluate rib fracture risk under loading from contemporary restraint systems. In a compilation of 93 matched Hybrid III and human cadaver sled crash tests, Kent et al. (2003a) showed that the risk of injury predicted by Hybrid III chest deflection is dependent on, among other factors, the type of restraint employed (i.e. seat belt, air bag, or combined seat belt plus air bag) and the test speed (Figure 2a). The observed Hybrid III dependence on restraint type suggests that for every type of restraint developed, there is a unique function relating Hybrid III chest deflection to injury risk. This eliminates the utility of the ATD by requiring cadaver testing of any new restraint to determine its unique injury risk function. This, however, is not true for human cadavers. Kent et al. (2003b) showed that human chest deflection, measured on cadavers, is an objective measure of rib fracture risk, independent of the type of restraint employed (Figure 2b). In a human, the re-distribution of force by the superficial soft tissues limits the effect of restraint load distribution on the deformation pattern of the rib cage (Kent et al. 2001a).

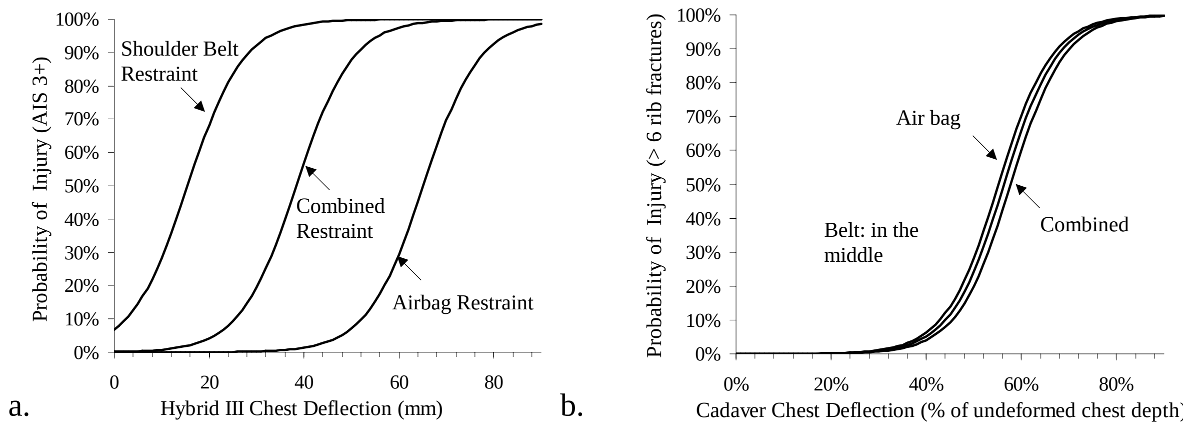

Figure 2: a. Functions relating Hybrid III chest deflection to risk of thoracic injury (Abbreviated Injury Score, AIS, 3+) for a 60 year old, 50th percentile male driver in a 48 km/h collision (from Kent et al. 2003a). b. Functions relating cadaver chest deflection to risk of thoracic injury (>6 rib fractures) for a 60 year old, 50th percentile male driver in a 48 km/h collision (from Kent et al. 2003b). The cadaver chest deflection injury risk functions are less sensitive to restraint condition than the Hybrid III injury risk functions.

To facilitate the prediction of rib fractures under diverse and novel loading conditions, it is desirable for an ATD to be able to predict injury based solely on measured responses, independent of the test conditions employed. It is possible that the test-condition-specific nature of the Hybrid III thoracic injury risk function is a result of, among other things, a restraint-dependent relationship between Hybrid III chest deflection and cadaver chest deflection. To investigate this, this study compared the measured maximum chest deflections in matched Hybrid III and cadaver frontal sled tests in a subset of the tests reported in Kent et al. (2003a, b). Test conditions varied by restraint type (e.g. 3-point belt plus airbag, lap belt plus airbag, etc.) and occupant seating position (driver or passenger).

METHODS

Identification of a Dataset

The datasets reported by Kent et al. (2003a, b) were examined to identify sets of matched cadaver-Hybrid III frontal sled tests with which to examine the relationship between cadaver and Hybrid III chest deflection as a function of test condition. Screening criteria included the following:

1) The tests were frontal, restrained sled tests.
2) Each set of cadaver tests had at least one corresponding (matched) Hybrid III test, performed under identical test conditions, in which internal Hybrid III chest deflection was measured.
3) Chest deflection must have been measured reliably in each cadaver test.
4) Cadaver tests resulting in more than 10 rib fractures were excluded.

Measurement of cadaver chest deflection in dynamic tests is commonly accomplished through the use of chest bands (a.k.a. External Peripheral Instrument for Deflection Measurement, Eppinger et al. 1989). These devices consist of a number of strain gauges bonded at discrete locations to a flexible steel strip, protected with an external layer of urethane rubber, that is wrapped around the chest of the test subject (e.g. model 4592, Robert A. Denton, Inc., Rochester Hills, Michigan). The strain gauges are sensitive to the long-axis bending of the strip; data from these gauges are then used to calculate the shape of the circumferential contour of the subject's chest during the test event, facilitating the determination of the maximum deflection of the subject's chest. The degree of curvature that the chest band can accurately reconstruct, however, is limited by the number strain gauges affected by the curvature. Early generation chest bands (e.g. those containing only 18 or 24 strain gauges along their length) lacked the resolution necessary to accurately reconstruct chest deformation contours under concentrated loading (e.g. belt loading) that resulted in small radii of curvature (Bass et al. 2000, Shaw et al. 2000). Therefore, in accordance with inclusion criterion #3, only cadaver tests using chest bands with relatively high resolutions (40+ gauges) were included in this study.

As noted in criterion #4, tests were screened for inclusion by the number of rib fractures that resulted from the test. Large numbers of rib fractures can destabilize the chest wall, resulting in a condition known as "flail chest". This condition allows artifactually high chest deflections that do not represent the response of an intact thorax. Thus, tests resulting in more than 10 rib fractures were excluded from this study.

Following the criteria above, 21 tests were identified for analysis (see Table 1 in Results). These tests were performed under four different test configurations:

A) Driver seating position; force-limiting, pre-tensioned 3-point belt plus full-powered airbag restraints

B) Right front passenger seating position; force-limiting, pre-tensioned 3-point belt plus depowered airbag restraints

C) Right front passenger seating position; standard (not force-limiting) 3-point belt plus depowered airbag restraints

D) Right front passenger seating position; lap belt only (no shoulder belt) plus full-powered airbag restraints

All tests were performed with a target ΔV of 48 km/h, with the occupant seated in a test fixture, or "buck", representing the interior of a mid-sized sedan (driver position tests – '93 Ford Taurus; passenger position tests – '97 Ford Taurus), mounted to a deceleration sled. Sled acceleration pulses were trapezoidal in shape, and were chosen to represent vehicle decelerations experienced in full frontal (passenger's side tests) and offset frontal (driver's side tests) barrier tests of the appropriate Ford Taurus. In all cadaver tests, deformation contours of the subjects' chests were recorded at two superior-inferior locations (nominally at the level of the 4^{th} and 8^{th} ribs, laterally) using 40-gauge chest bands. Maximum cadaver mid-sternal chest deflection was then calculated using the method described by Kuppa and Eppinger (1998). Specific details regarding the test procedures (including acceleration pulses, subject positioning, instrumentation, and data analysis) and the test subjects can be found in Shaw et al. 2000 (driver's side tests) and Kent et al. 2001a (passenger's side tests).

Data Analysis

The goal of this study was to examine the relationship between the injury predictor variables of the restraint-dependent Hybrid III thoracic IRF and the restraint-independent cadaver-based thoracic IRF (Kent et al. 2003a, b). Thus, the maximum internal (slider) chest deflection measurement (C_{H3max}, expressed in mm) was considered for the Hybrid III tests, and the maximum external mid-sternal chest deflection (C_{CADmax}, measured by chest bands and expressed as a percent of the undeformed chest depth) was considered for the cadaver tests. For each test condition, the average Hybrid III maximum internal chest deflection, $C_{H3maxAVG}$, was calculated. For each cadaver test the chest deflection ratio, R_d, was calculated as the ratio of C_{CADmax} to the $C_{H3maxAVG}$ associated with the test condition of interest (Equation 1). These R_d values were then examined to compare the relationship between cadaver external chest deflection and Hybrid III internal chest deflection across the test conditions of interest.

$$R_d = \frac{C_{CADmax}}{C_{H3maxAVG} \text{ (for the test condition of interest)}} \qquad [1]$$

RESULTS

The tests examined in this study, their resulting chest deflection measures, and their resulting R_d values are presented in Table 1.

Table 1: Summary of Tests and Chest Deflection Results

Test #*	Position	Restraint	ΔV (km/h)	Subject	Stature (cm)	Mass (kg)	Age/Gender	C_{CADmax} (%)**	C_{H3max} (mm)**	R_d (%/mm)
532[1]	Driver	FLB+AB	48.6	H3 50^{th}%	--	--	--	--	27	--
537[1]	Driver	FLB+AB	48.9	H3 50^{th}%	--	--	--	--	29	--
538[1]	Driver	FLB+AB	48.1	H3 50^{th}%	--	--	--	--	26	--
H3 Average	**Driver**	**FLB+AB**	**48.5±0.4†**	**H3 50^{th}%**	**--**	**--**	**--**	**--**	**27±1.5††**	**--**
533[1]	Driver	FLB+AB	48.6	Cad 104	163	63.5	67/F	15	--	0.53
534[1]	Driver	FLB+AB	48.4	Cad 76	175	50.8	47/M	18	--	0.67
544[1]	Driver	FLB+AB	49.2	Cad 83	169	56.0	59/F	24	--	0.88
545[1]	Driver	FLB+AB	48.1	Cad 103	184	74.0	67/M	16	--	0.57
571[2]	Passenger	FLB+AB	47.6	H3 50^{th}%	--	--	--	--	29	--
572[2]	Passenger	FLB+AB	48.1	H3 50^{th}%	--	--	--	--	28	--
576[2]	Passenger	FLB+AB	48.1	H3 50^{th}%	--	--	--	--	30	--
H3 Average	**Passenger**	**FLB+AB**	**47.9±0.3**	**H3 50^{th}%**	**--**	**--**	**--**	**--**	**29±1.0††**	**--**
577[2]	Passenger	FLB+AB	47.4	Cad 111	174	70	57/M	23	--	0.79
578[2]	Passenger	FLB+AB	47.6	Cad 107	155	52.5	69/F	25	--	0.86
580[2]	Passenger	FLB+AB	47.6	Cad 105	177	57	57/M	28	--	0.97
663[2]	Passenger	SB+AB	48.1	H3 50^{th}%	--	--	--	--	40	--
664[2]	Passenger	SB+AB	47.6	H3 50^{th}%	--	--	--	--	40	--
H3 Average	**Passenger**	**SB+AB**	**47.9±0.4**	**H3 50^{th}%**	**--**	**--**	**--**	**--**	**40±0.0††**	**--**
665[2]	Passenger	SB+AB	48.9	Cad 112	176	85.3	55/M	28	--	0.70
666[2]	Passenger	SB+AB	48.1	Cad 115	176	83.9	69/M	32	--	0.80
648[2]	Passenger	LB+AB	48.6	H3 50^{th}%	--	--	--	--	19	--
649[2]	Passenger	LB+AB	47.6	H3 50^{th}%	--	--	--	--	19	--
H3 Average	**Passenger**	**LB+AB**	**48.1±0.7**	**H3 50^{th}%**	**--**	**--**	**--**	**--**	**19±0.0††**	**--**
651[2]	Passenger	LB+AB	48.6	Cad 121	176	70	70/M	11	--	0.56
652[2]	Passenger	LB+AB	49.7	Cad 118	175	73.5	46/M	12	--	0.62

* Sources: 1 – Shaw et al. 2000; 2 – Kent et al. 2001a Note: All cadavers from source 1 were preserved by embalming, all from source 2 were preserved by freezing and refrigeration.

** C_{H3max} is the maximum deflection of the internal sternum slider of the Hybrid III, expressed in mm. C_{CADmax} is the maximum cadaver chest measured externally by chest bands, expressed as a percent of the undeformed chest depth.

† Average ± standard deviation

†† $C_{H3maxAVG}$ for the test condition shown, ± one standard deviation

Average values of R_d (and 95% confidence intervals, assuming a normal distribution) were calculated for each test condition (Figure 3). To facilitate the discussion of these results in the context of the Hybrid III IRF (described in the Discussion below), average (and 95% confidence interval) R_d values were also calculated for the set of data including all passenger's side tests with combined 3-point belt and airbag restraints (both FLB+AB and SB+AB, this is termed the "combined" passenger's side data set). Figure 3 shows that R_{dAVG} for the passenger's side tests with combined 3-point belt and airbag restraint differed significantly from that of the passenger's side tests with lap-belt and airbag restraint ($p<0.05$). In contrast, because of its large confidence interval, it cannot be stated that the R_{dAVG} for the driver's side tests differed significantly from the R_{dAVG} for any of the passenger's side tests.

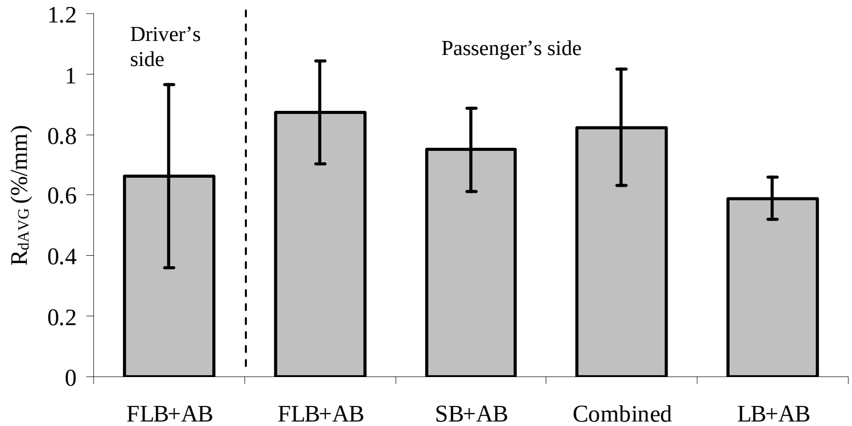

Figure 3: Average and 95% confidence intervals for R_d for each test condition (FLB+AB = pretensioned, force-limiting 3-point belt plus airbag; SB+AB = standard 3-point belt plus airbag; LB+AB = lap belt plus airbag). Note that the "combined" passenger's side condition is a combination of the passenger's side FLB+AB and SB+AB data sets.

DISCUSSION

Role of R_d in Hybrid III Injury Risk Functions

Kent et al. (2003b) reported that chest deflection measured on a cadaver was a significant predictor of rib fractures, independent of the type of the test condition employed. This implies that cadaver chest deflection is related to the cause of rib fractures (e.g. the magnitude of strain within the ribcage), in a manner that is independent of the method or distribution of thoracic loading. For the purposes of this paper, predictor variables of this type (i.e. test condition independent) are termed "objective predictors". As a result, any other objective predictor of thoracic injury would also be an objective predictor of cadaver chest deflection, and vice versa. In contrast, Hybrid III chest deflection is neither an objective predictor of thoracic injury, nor an objective predictor of cadaver chest deflection. Kent et al. (2003a) showed that thoracic injury risk predicted from Hybrid III maximum chest deflection in frontal sled tests is dependent on, among other things, the speed of the test and the type of restraint employed. The analyses presented here suggest that the relationships between Hybrid III and cadaver maximum chest deflection are also dependent on, at least, the type of restraint employed. This is consistent with the restraint-dependency observed in the Hybrid III IRF.

If cadaver chest deflection is a cause of, and objective predictor of, osseous thoracic injury, then the Hybrid III IRF should be expressable as a transformation of the cadaver IRF using some function relating Hybrid III chest deflection to cadaver chest deflection. Such a cadaver-to-Hybrid III IRF transformation was explored by Laituri et al. (2005) for crashes involving restraint from 3-point belts only (no airbags). That study transformed a cadaver-chest-deflection-based IRF into a Hybrid III based IRF using a non-linear function relating Hybrid III chest deflection to cadaver chest deflection. Given the restraint specificity observed in the Kent et al. (2003a) Hybrid III IRF, however, it is likely that such a transfer function would also be a function of the restraint conditions employed (this was not considered in Laituri et al. 2005 because only one restraint condition was considered). To examine this, the inverse transformation was performed. The restraint-dependent Kent et al. (2003a) Hybrid III IRFs (Equation 2) were expressed in terms of cadaver chest deflection using R_{dAVG} (and its corresponding 95% confidence intervals) for each test condition (Equations 3 and 4).

$$P(AIS3+) = \frac{1}{1+e^{-(q'+\beta_8 C_{H3max})}} \qquad [2]$$

$$P(AIS3+) = \frac{1}{1+e^{-(q'+\beta_8 \frac{C_{CADmax}}{R_{dAVG}})}} \qquad [3]$$

$$P(AIS3+)bounds = \frac{1}{1+e^{-(q'+\beta_8 \frac{C_{CADmax}}{R_{dAVG} \pm 1.96 \times R_{dSD}})}} \qquad [4]$$

where

$$q' = \beta_0 + \sum_{i=1}^{7} \beta_i x_i \qquad [5]$$

and where x_i are the predictor variables defining the risk function for a given set of test conditions and subject characteristics, and β_i are the model coefficients (model coefficients and definitions of predictor variables can be found in Kent et al. 2003a). Predictor variables regarding test conditions included occupant position, test ΔV, and restraint condition (shoulder belt dominated restraint, airbag dominated restraint, or combined shoulder belt and airbag restraint). Predictor variables regarding characteristics of the test subject included age at death, mass, and gender. It is worth noting that the bounds calculated in Equation 5 represent only uncertainty in R_{dAVG}, and do not include the uncertainty in the original Hybrid III injury risk function. The true confidence bounds for the transformed IRF would be much larger than those presented in this study.

As the Kent et al. (2003a) statistical model did not distinguish between FLB+AB loading and SB+AB loading, it was necessary to use the “combined loading” passenger’s side model for both sets of passenger’s side tests that included shoulder belt and airbag restraint. Also, as the lap belt plus airbag restraint (LB+AB) resulted in the thorax being loaded solely by the airbag, this was considered to be an “airbag dominated” restraint condition. Finally, to facilitate comparison across test conditions, the occupant-based predictor variables were set to the same values for each IRF (male occupant, 60 years old, mass of 77 kg).

These models were then compared to the cadaver-chest-deflection-based IRF developed reported by Kent et al. (2003b) (Figure 4). To accommodate the differences in predicted variables of the two models (Hybrid III model – predicts AIS 3+ injury; cadaver model – predicts 6+ rib fractures), the dependent axes for each plot in Figure 4 are presented as the "Probability of Severe Injury".

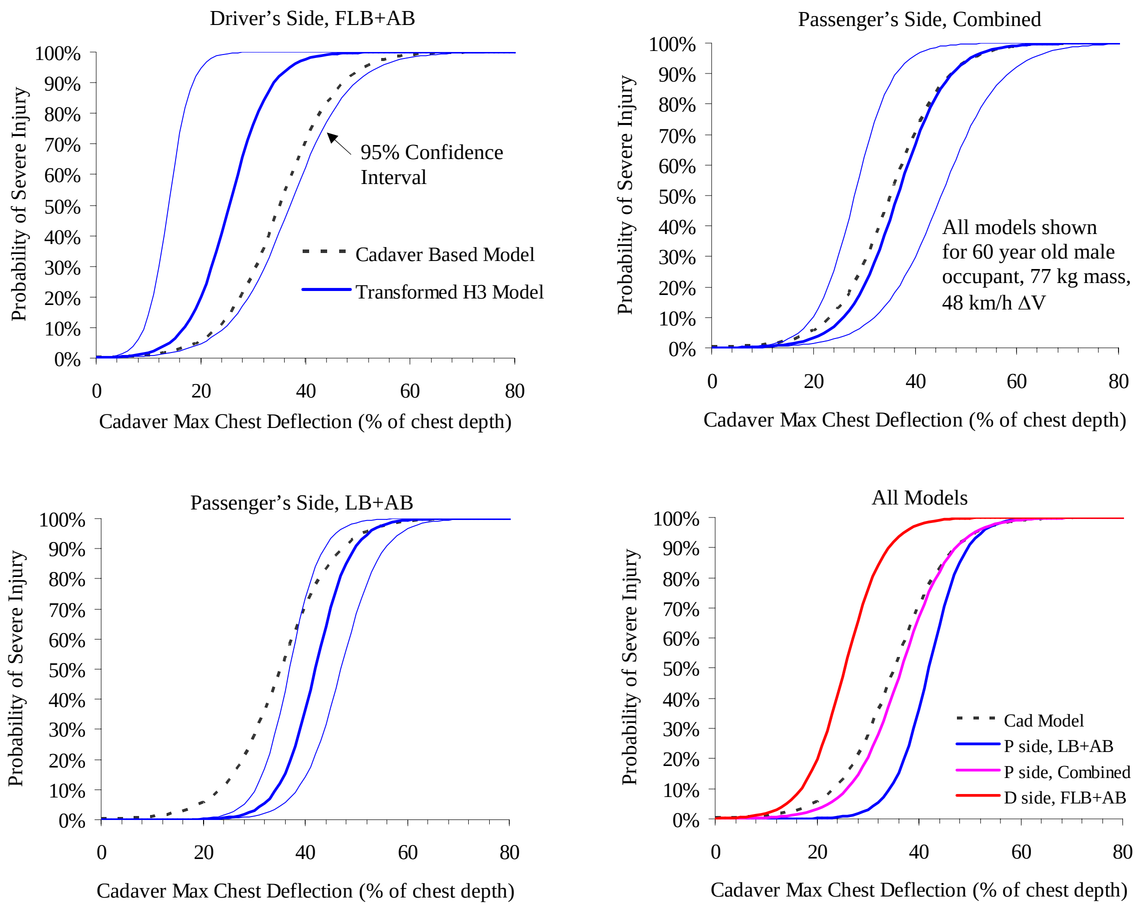

Figure 4: Plots of the cadaver-chest-deflection based injury risk curves developed here through transformation of the Hybrid III injury risk curves described by Kent et al. (2003a), using the chest deflection ratios, R_{dAVG}. These are compared to the restraint-independent cadaver-chest-deflection-based injury risk function described by Kent et al. (2003b).

The transformation described above reduces the restraint dependency of the Hybrid III injury risk function. The plots above suggest reasonable agreement between the Hybrid III IRFs (transformed into functions of cadaver chest deflection) and the cadaver-chest-deflection-based IRF. On inspection, the passenger's side, combined restraint condition appears to show the closest agreement between the two IRF types. In contrast, the average driver's side, FLB+AB Hybrid III model appears to deviate from the cadaver model. Because of the wide range of the model bounds, however, the cadaver model still falls within the 95% confidence interval for the Hybrid III model for this test condition. Furthermore, it is possible that the mode of preservation used in the driver's side tests (embalming) contributed to the discrepancy shown here. Embalming may have stiffened the thoraces of these subjects relative to the subjects of the other tests (Crandall et al. 1994), resulting in less cadaver chest deflection and smaller values for R_d. Lastly, the cadaver-based IRF lies outside the confidence intervals for the transformed Hybrid III passenger's side, LB+AB (airbag dominated loading) IRF at the beginning portion of the curve. This may be due to the limited fidelity of the airbag-only portion of the Hybrid III IRF. Very few airbag-only tests were included in the original development of this function, and none of these tests resulted in MAIS3+ injury to the occupant. Thus, the Hybrid III IRF is not well defined for airbag dominated thoracic loading, and the accuracy of the transformed LB+AB Hybrid III IRF is suspect.

Caution must be taken when using the plots above to compare the transformed Hybrid III IRFs to the cadaver IRF. First, it is likely that the relationship between Hybrid III and cadaver chest deflection is non-linear (e.g. Laituri et al. 2005). In other words, this relationship is likely a function of the magnitude of deflection, as well as the test conditions. This non-linearity in R_d would cause deviations in the transformed Hybrid III IRF, the nature of which are unknown at this time, at magnitudes of chest deflection different than those presented here. Furthermore, the force-deflection responses of both the human and Hybrid III thoraces are visco-elastic, or loading-rate dependent (Kent et al. 2001b, 2002). This may result in a loading-rate dependence in the Hybrid III-cadaver chest deflection relationship, the nature of which was not investigated here. Second, as mentioned above, neither the transformed Hybrid III models nor the cadaver-based models shown here include the uncertainty associated with the original models. The true confidence intervals for the transformed Hybrid III IRFs are likely larger than those presented here. Lastly, the transformed Hybrid III IRF and the cadaver-based IRF predict different classifications of injury (Hybrid III model – predicts AIS 3+ injury; cadaver model – predicts 6+ rib fractures). Although these classifications may be similar, this difference may have resulted in some shift in the IRFs relative to one another.

Despite the limitations above, however, the analyses performed here illustrate the role of the restraint-dependence in R_d in affecting the restraint dependence in the Hybrid III IRF. This analysis has combined the restraint-dependent Hybrid III IRFs with the restraint-dependent chest deflection relationships, and arrived, approximately, at the restraint-independent cadaver-chest–deflection based IRF. This suggests that the magnitude and nature of restraint-dependency observed in the Hybrid III injury risk functions are comparable to those observed in the Hybrid III-cadaver chest deflection relationship. Although the analyses presented here are of limited scope, future work is planned to expand on the data set presented here and investigate this issue further.

Possible Causes of Observed Restraint-Dependency

Although a detailed discussion of the causes of the observed Hybrid III chest deflection and IRF restraint-dependency is beyond the scope of this paper, it is worth noting some factors that may contribute to this phenomenon. First, this may be caused by restraint-dependent differences in the mechanical characteristics of the Hybrid III and cadaver thoraces. For example, Kent et al. (2002, 2004) showed that the stiffness (imparted force vs. resulting deflection) of cadaver and Hybrid III thoraces varies with the type of loading employed (e.g.

belt-like loading, distributed loading). Similarly, the relationship between cadaver and Hybrid III thoracic stiffness also varies according to loading condition (Forman et al. 2005, Cesari and Bouquet 1994). This may be due, in part, to differences in the geometry of the upper thorax. When a cadaver thorax is loaded by a shoulder belt, part of the load is borne by the clavicle and stiff upper ribs, off-loading the more compliant lower ribcage. The Hybrid III, however, has no clavicle to act as an alternate load path, causing the shoulder belt force to be borne entirely by the rib cage. This structural difference would likely contribute more to situations in which loading was concentrated over the shoulder (e.g. belt loading) than to situations in which loading was spread over the entire chest (e.g. airbag loading), resulting in restraint-dependent differences in cadaver and Hybrid III thoracic stiffness. This would result in restraint-dependent differences in Hybrid III and cadaver thoracic deflections for a given magnitude of applied restraining force.

The observed restraint-dependencies may also be due to restraint-dependent differences in Hybrid III and cadaver kinematics. In a frontal crash, the engagement of the occupant's thorax by a restraint system is dictated by the magnitude of forward motion of the thorax (excursion) relative to the interior of the vehicle. If the forward excursion or rotation of the Hybrid III torso did not match that of the cadavers, then the Hybrid III would not experience an interaction with the shoulder belt and airbag representative of that experienced by the cadavers. For example, the forward motion of the thorax is influenced by the restraint of the pelvis by the lap belt. While the lumbar spine of the human likely exhibits compliance when loaded in shear, flexion, and axial extension, the lumbar spine of the Hybrid III is a rigid steel connection between the pelvis and the thorax. This rigid connection would likely influence the forward excursion of the thorax, the amount of restraining force borne by the thorax, and the resulting Hybrid III thoracic deflection. As occupant kinematics are dependent on the type of restraints employed, it is likely that differences in occupant kinematics are also dependent on the types of restraints employed. This may result in restraint-dependent differences in occupant thoracic loading and chest deflection.

Lastly, it is possible that the observed restraint-dependencies may result from systematic errors in the chest deflection measurement techniques employed. Bass et al. (2000) and Shaw et al. (2000) reported the resolution of chest bands limits their accuracy as their measured radii of curvature decreases. As a result, chest bands may be more accurate under distributed loading (e.g. airbag-like loading) that results in large radii of curvature than under concentrated loading (e.g. belt-like loading) that results in smaller radii of curvature. Because of the relatively high resolution of the chest bands employed in the tests investigated here, however, it is believed that this would have negligible effect on the results presented here. Perhaps more importantly, it is possible that the accuracy of the Hybrid III sternum slider is dependent on the type of restraint employed. In a study of Hybrid III internal chest deflection in frontal sled tests, Butcher et al. (2001) suggested that the chest deflection measured by the sternum slider deviated from the true value by as much as 23% under standard 3-point belt plus airbag restraint, and by as much as 40% under lap belt plus airbag restraint. These deviations were caused, in part, by rotation of the sternum relative to the base of the slider due to loading concentrated on the superior portion of the sternum. Such deviations are indicative of systematic, restraint-dependent errors in Hybrid III chest deflection measurement that may have resulted in the restraint-dependent chest deflection relationships investigated here.

CONCLUSIONS

Chest deflection data from 21 cadaver and Hybrid III frontal, restrained sled tests were analyzed. Test conditions varied by occupant position (driver or passenger) and restraint type (force-limiting 3-point belt plus airbag-FLB+AB, standard 3-point belt plus airbag-SB+AB, lap belt plus airbag=LB+AB); all tests were performed with ΔV's of 48 km/h. These analyses suggested that the ratio of maximum Hybrid III internal chest deflection (measured by the sternum slider) to maximum cadaver chest deflection (measured externally by chest bands) may vary based on the restraint condition employed. Furthermore, the IRF transformation analysis presented here suggested that the nature and magnitude of the restraint-dependency observed in the Hybrid III-cadaver chest deflection relationship was comparable to the restraint-dependency of the Hybrid III injury risk functions. Future work is planned to expand the data set presented here to include different types of restraint systems (e.g. 3-point belt only) and tests at different speeds.

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