DEVELOPMENT OF AN AGE-DEPENDENT THORACIC INJURY CRITERION FOR FRONTAL IMPACT RESTRAINT LOADING

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Paper No. 72

ABSTRACT

Data from 93 human cadaver tests (age range 17-86 years, mean 60.2, S.D. 13.3) were used to develop thoracic injury risk functions for frontal loading. The set of potential predictors included the maximum chest deflection, the age of the cadaver at death, the cadaver's gender, and the loading condition on the anterior thorax: blunt hub (41 tests), seat belt (26 tests), air bag (12 tests), and combined belt-and-bag (14 tests). Predicted outcomes were the probability of any rib fractures (onset of injury) and the probability of greater than six rib fractures (severe injury). Linear logistic regression models were used with the outcome modeled as a binary response (injury, no injury). It is shown that the injury risk function is not dependent on the loading condition (e.g., the 50% risk of injury does not change when the loading condition changes), but that the injury risk function is strongly dependent on the age of the cadaver at death. A significant injury risk model with good ability to discriminate injury from non-injury tests ($p < 0.0001$, Chi-square = 21.49, area under ROC = 0.867, Kruskal's Gamma = 0.732) is presented using only maximum chest deflection and cadaver age as predictors of injury risk. The 50% risk of any rib fractures is found to occur at 35% chest deflection for a 30-year-old, but at 13% deflection for a 70-year-old. The 50% risk of severe injury is shown to occur at 33% chest deflection for a 70-year-old, but at 43% for a 30-year-old.

INTRODUCTION

The posterior displacement of the anterior chest relative to the posterior chest under frontal loading is often used to describe thoracic injury risk. This displacement, commonly referred to as chest deflection, is measured by contemporary frontal impact anthropomorphic test dummies (ATDs) and a chest deflection limit is specified in Federal Motor Vehicle Safety Standard (FMVSS) 208 – Frontal Impact Protection. Many cadaver-based studies have shown an increasing risk of injury as chest deflection increases (Kent et al. 2001a, Morgan et al. 1994, Kuppa and Eppinger 1998) and the National Highway Traffic Safety Administration has published an injury risk function quantifying this increase for one data set (Figure 1).

The injury risk associated with a particular magnitude of chest deflection is not, however, constant for all conditions. For example, Zhou et al. (1996) showed that chest deflection tolerance under blunt hub loading decreases with increasing age. This decrease results from several characteristic anatomical and material changes that are strongly related to aging. It is well established that the mineral density, fracture toughness, and failure strain of bone decreases with increasing age, starting at about age 30 (e.g. Yamada 1970). Furthermore, aging is related to a change in the proportion of cortical bone in the rib. 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 (failure force (N) = 254 – 3.34*age (years), $p<0.001$). 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 (cortical bone area (mm^2) = 32.9 – 0.19*age (years), $p < 0.001$). The progressive calcification of the costal cartilage may also tend to decrease the chest deflection tolerance of the rib cage by reducing the failure strain of the cartilage. Furthermore, age-related anatomical changes that may influence chest deflection tolerance have been observed anecdotally (Wang 2003). For example, the slope of ribs in the sagittal plane may decrease with increasing age (this slope can be seen in Figure 2b). This decrease results in the "barrel chest" condition that is often observed in computed tomography (CT) scans of older patients or cadavers, but is not often seen in younger or middle-aged subjects. Compared to a younger person, this decreased rib slope may result in increased strain

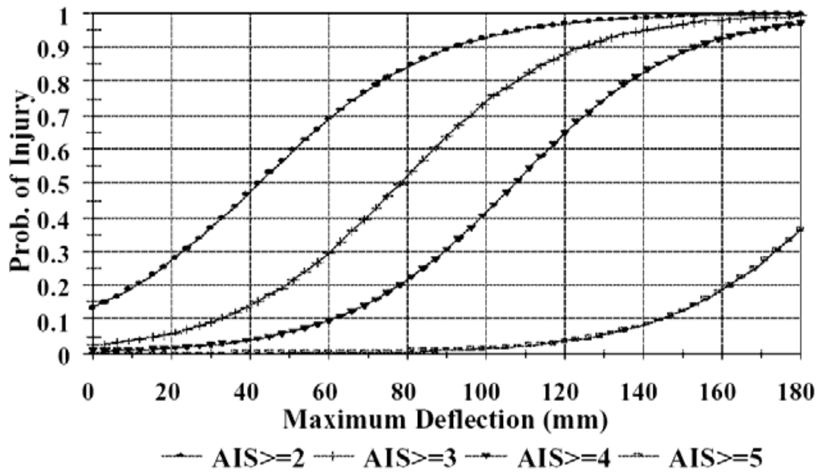
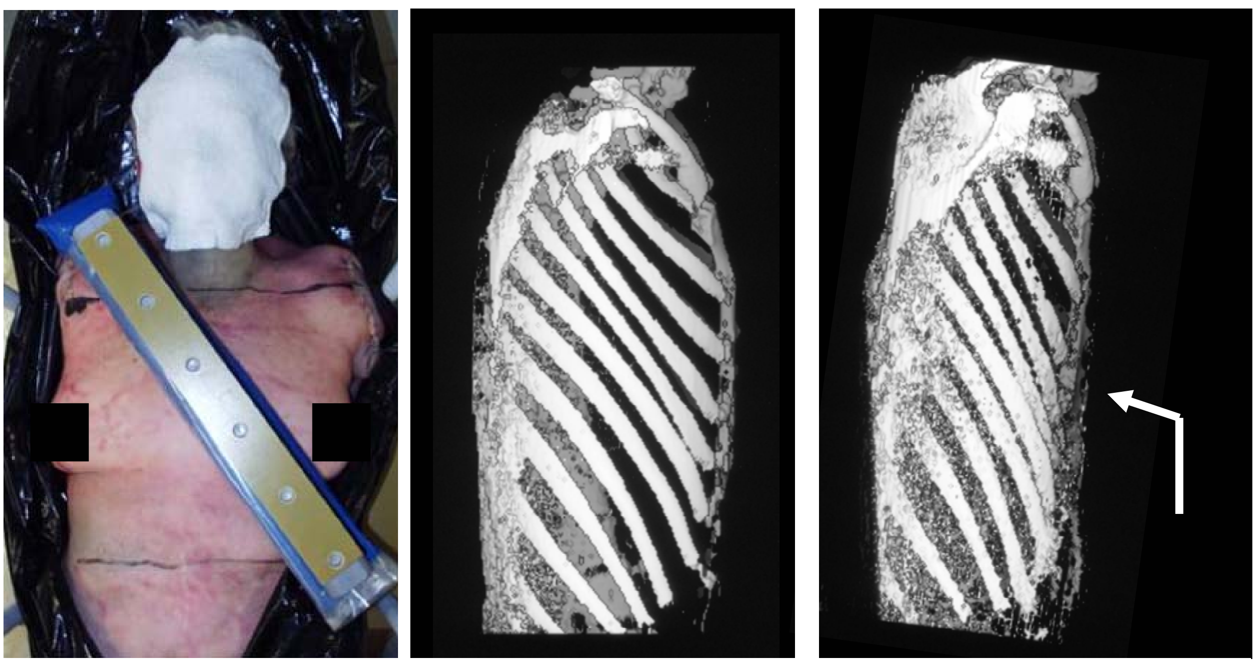


Figure 1. Chest deflection injury risk curve published by the NHTSA (Eppinger et al. 1999) showing increasing risk of all injury levels as maximum chest deflection increases.



right left
a. Anterior view of loading condition used in CT study

posterior anterior
b. Right lateral view of unloaded thorax

posterior anterior
c. Right lateral view of loaded thorax

Figure 2. CT study showing rib deformation pattern under concentrated "belt-like" loading. Note especially ribs 3 through 6 exhibiting concentrated deformation unlike that seen with a distributed "air bag-like" load (see Kent et al. 2001a).

in the rib for a given level of chest deflection since the rib's tendency to rotate inferiorly about the costo-vertebral articulation is decreased.

One purpose of this paper, therefore, is to quantify how the chest deflection injury tolerance changes with age. A second purpose is to evaluate how the nature of the loading on the chest influences the chest deflection injury tolerance. Contemporary restraints generate a complex loading environment on the chest. Interaction with an air bag results in a well-distributed, nearly constant pressure field on essentially the entire anterior chest, the neck, and the head. In contrast, the seat (shoulder) belt generates relatively concentrated forces on fewer anatomical structures (clavicle, sternum, ribs). Steering wheel loading involves yet a third distinct pattern. Further complicating the issue is the fact that most collisions will involve a combination of different loading patterns (e.g., belt and bag or bag and wheel). If the chest deflection injury tolerance varies significantly as the loading environment changes from belt to hub to air bag loading, then the assessment of restraint systems in the laboratory must consider this sensitivity. It is currently not known, however, how the chest deflection injury tolerance is affected by a change in the load distribution (loading condition). Rib fracture patterns vary depending on the restraint condition (e.g., Patrick 1974, Yoganandan et al. 1998, Kent et al. 2002a) and a CT scan of a loaded thorax will reveal a loading condition-specific strain distribution in the rib cage (Figure 2). Based on these observations, it may be assumed that the chest deflection tolerance will change as the restraint condition changes. We have hypothesized; however, that variation in individual tolerance and the load-distributing effects of the soft tissues may be sufficient to effectively mask the changes in deflection tolerance associated with a change in restraint type (Kent et al. 2001a). That previous study was limited, however, in the number of cadavers considered and was unable to conclude definitely that the tolerance was insensitive to restraint condition. The current study expands on that analysis by including an expanded dataset sufficient for a robust statistical analysis of the hypothesis.

METHODS

Dataset Development

The literature was searched to identify cadaver tests having certain characteristics. Five inclusion criteria for the dataset were established:

1. The cadaver must have been subjected to anterior loading (i.e., lateral impacts were not considered).
2. The age and gender of the cadaver must be known.
3. The number of rib fractures resulting from the loading must be known.
4. The loading must be concentrated through the sternum. This eliminated cases where, for example, the lower steering wheel rim generated rib fractures in an unbelted subject with an air bag. It is apparent that chest deflection measured at the mid-sternum is inappropriate for this type of loading (Prasad (1999) and Kent et al. (2000) discuss this type of loading in more detail).
5. The chest deflection must have been measured reliably.

These final two requirements precluded the inclusion of many cadaver sled tests. Driver-side sled tests with an air bag and no belt were not included since these have exhibited pronounced interaction between the inferior thoracic cage and the lower steering wheel rim. Furthermore, as discussed in some detail by Kent et al. (2001b), Bass et al. (2000), Shaw et al. (2000), and in an unpublished study by Hagedorn and Burton (1993), the use of chestbands to resolve thoracic deflections in a sled test is subject to error in some instances. For example, chestbands, especially the early-generation 18-gage and 24-gage bands, have insufficient resolution to track the thoracic contour under concentrated belt loading. Chestbands are, however, considered to be accurate at resolving thoracic contours under distributed loading (Pintar et al. 1997). As a result, we excluded most available sled tests from consideration. We did not include any sled tests involving belt-only loading. We also did not include any sled test that involved a chestband having fewer than 40 gages. This left relatively few sled tests available for inclusion. Sled tests that were used in the analysis involve high-resolution (40+ gage) bands and include tests with combined belt-and-air bag loading (both standard and force-limiting belts) and tests with air bag loading on the passenger side where the confounding effect of the lower steering wheel rim is not present.

Ninety-three tests were available for consideration after eliminating those that failed one or more of the criteria listed above (Appendix A). The tests can be conveniently grouped into four primary loading conditions:

1. Blunt hub loading (41 tests)
2. Seatbelt loading (26 tests)
3. Distributed loading (12 tests)
4. Combined belt-and-bag loading (14 tests)

The blunt hub loading condition involved cadavers with either a fixed or free back loaded by a 15.2-cm diameter circular hub at the approximate location of the 4th interstitial space. All but one of these tests were first described by Kroell et al. (1971, 1974), but the values used here were taken from Viano (1978), who summarized the tests in a convenient form. One blunt hub test was performed by the University of Virginia (UVA) and publication of this test is pending. Chest deflection in the Kroell tests was measured visually using high speed film and a rod passing through the thorax and in the UVA test using a potentiometer attached to the hub.

The seatbelt loading condition involved cadavers positioned supine on a flat loading table with a narrow belt passing diagonally over the anterior thorax (Cesari and Bouquet 1991, Cesari and Bouquet 1994). In the Cesari and Bouquet tests, force was applied via a cable system and an impactor. In a single UVA test (publication pending), force was applied via a hydraulic-and-cable system and a high-speed material test machine. Chest deflection was measured via an array of potentiometers. The mid-sternal potentiometer data were used for this analysis.

The distributed loading tests were of three types. Seven tests come from the series of static, out-of-position air bag deployment tests performed by Crandall et al. (1999). Three tests come from the series published by Kent et al. (2001a). This series involved passenger-side sled tests with an air bag and no shoulder belt. In the Crandall and Kent tests, chest deflection was measured using high-resolution chestbands. Finally, two table-top tests were used (Kent et al. 2002b). These tests involved a 20.3-cm wide belt oriented laterally over the thorax, with chest deflection measured by a potentiometer.

The combined belt-and-bag loading condition involved cadaver tests from several sources. Eight tests come from the series presented by Kent et al. (2001a), four come from the series presented by Shaw et al. (2000), and three come from the series presented by Kallieris et al. (1995). In all cases, chest deflection was measured using high-resolution chestbands.

Analysis

Four mulivariate linear logistic regression models were used to evaluate whether the injurious level of maximum chest deflection (C_{max}) was sensitive to the age of the cadaver at death or to the loading condition (Table 1). Two outcomes were modeled. In both cases, the outcome variable was a binary response. In one set of models, cases with any rib fractures (fx > 0) were coded as "injury" (Y = 1) and cases with no rib fractures (fx = 0) were coded as "no injury" (Y = 0). In another set of models, cases with more than six rib fractures (fx > 6) were coded as "severe injury" (Y = 1) and cases with six or fewer fractures (fx < 7) were coded as "no severe injury" (Y = 0). Abbreviated Injury Scale (AIS) coding could not be used because many of the tests were published prior to the advent of the AIS scale and the fracture locations were not presented in sufficient detail to determine AIS. The two outcomes were evaluated so that the C_{max} level corresponding to the probability of the onset of injury and to the probability of severe injury could be determined.

For each outcome level (injury onset, severe injury), a "full" model was developed. The full models included all available parameters that have some biomechanical justification for potentially influencing injury threshold: C_{max}, the age of the cadaver at death, the loading condition, gender, and an age*gender interaction term. Based on the results of the full models, "reduced" models were developed for each outcome level using only those predictors that were found to be significant in the full model. Generalized Wald tests were performed to verify the validity of removing these variables. For all models, the relative importance of each covariate was assessed using the covariate's Wald chi-square statistic minus its degrees of freedom (DOF). The chi-square statistic indicates the amount of variance in the outcome that is explained by each covariate and subtracting the DOF accounts for the fact that covariates with different DOF were included in the same model.

Table 1. Models Developed and Evaluated

Model	Outcome	Predictors
1	> 6 rib fractures	age, gender, loading
2	> 0 rib fractures	cond., C_{max}, age*gender
3	> 6 rib fractures	age, C_{max}
4	> 0 rib fractures	

Gender and the loading condition (blunt hub, belt, distributed, combined) were treated as classification variables in the analysis, while age (years) and C_{max} (percent of initial anterior-posterior chest depth measured externally) were treated as continuous covariates.

The logit of the probability of thoracic injury (both levels of "injury"), P(I), was modeled as a linear function of the value of the predictors, x_i:

$$P(I) = \frac{1}{1+e^{-q}} \qquad [1]$$

where

$$q = \alpha + \sum_i \beta_i x_i \qquad [2]$$

is the logit function, α is the intercept, x_i are the model predictors, and β_i are the coefficient associated with each predictor.

Several parameters were utilized to evaluate the predictive ability of the various models, including percent concordance and discordance, Kruskal's Gamma, and the area under the receiver operator characteristic (ROC) curve. A pair of observations with different outcomes (injury and non-injury) is concordant if the model predicts a higher risk of injury for the case with injury than for the case without. A pair of observations is discordant if the injury case has a lower model-predicted risk than the non-injury case. Kruskal's Gamma is defined by the number of concordant and discordant pairs in the dataset, so it is a measure of the model's ability to discriminate injury from non-injury cases:

$$\gamma = \frac{N_{conc} - N_{disc}}{N_{conc} + N_{disc}} \qquad [3]$$

where N_{conc} is the number of concordant pairs and N_{disc} is the number of discordant pairs. A Gamma value of zero indicates that the model has no predictive ability, while a value of one indicates perfect prediction.

The value for the area under the receiver operator characteristic (ROC) curve (Metz 1978, Metz 1986, Hanley 1989) was also used as a criterion for judging the ability of the predictor to distinguish between the injured and non-injured specimens. A unitless measure, the value of the area under the ROC curve summarizes the functional relationship between the sensitivity and the specificity of the discrimination tool to classify observations into one of two distinct groups: those that have the characteristic of interest (injury in this case), and those that do not. The value of the area under the ROC curve can be interpreted as the probability of a randomly selected observation that has the characteristic of interest (injury or non-injury) being viewed as more likely (model-based predicted probability) of having the characteristic than a randomly selected observation that does not have the characteristic. The advantage of the area under the ROC over Kruskal's Gamma is that ties are intrinsically considered. Kruskal's Gamma considers only concordance and discordance (see Equation [3]). In this analysis, there are no ties, so these two parameters indicate the same trends in model performance.

For a predictor or set of predictors that produces an area under the ROC curve value of 0.50, the utility of the model to correctly classify the outcomes is no better than basing the classification on the flip of a fair coin, while perfect discrimination corresponds to an area under the ROC curve of 1.0. As a general guideline, models that produce an area under the ROC curve within the range of 0.50 to 0.60 are considered to have little or no utility as a discriminating tool, 0.60 to 0.70 poor utility, 0.70 to 0.80 moderate utility, 0.80 to 0.90 good utility, and 0.90 to 1.0 excellent utility.

RESULTS

The dataset includes 93 tests, of which 71 resulted in at least one rib fracture and 39 resulted in more than 6 rib fractures. The age range is 17 to 86 years (median 60.2, standard deviation 13.3) and the majority of the subjects (66) are male. A distribution of injury and non-injury (both levels of injury) cases is present for each loading condition (Figure 3).

The model coefficients for Model 1 and their standard errors are presented in Table 2 and the Analysis of Variance (ANOVA) for Model 1 is presented in Table 3. Model 1 is a significant model of the outcome ($p = 0.0033$), but only two of the covariates (age and C_{max}) are significant to the $p = 0.1$ level. Gender and loading condition are not significant covariates. The age*gender interaction term approaches significance ($p = 0.1471$). The chi-square minus DOF analysis indicates that C_{max} is the most important covariate in terms of predicting severe injury outcome, followed by age, gender, and age*gender. The loading condition is the least important of the predictors.

The Model 2 coefficients and standard errors are presented in Table 2, and the ANOVA summary is presented in Table 3. Model 2 is a significant predictor of the outcome. The ANOVA results are similar to the findings for Model 1: age and C_{max} were the only significant covariates and ranked as the most important predictors in the model. Again, the loading condition is the least important predictor in the model. Figure 4 illustrates the load condition insensitivity of the injury risk functions.

Based on the findings from Model 1 and Model 2, we performed a generalized Wald test for dropping the variables gender, loading condition, and age*gender from the models. In both cases, the removal of these variables did not significantly change the predictive ability of the model (Model 1 $p = 0.6390$ and Model 2 $p = 0.3456$). The model coefficients and standard errors for these reduced models (Model 3 and Model 4) are presented in Table 4 and the ANOVA results are presented in Table 5. Both reduced models were significant. Age and C_{max} were significant predictors in both models and C_{max} remained a more important predictor than age.

While the removal of non-significant variables simplified the models, the most important determinant of the validity of our model reduction strategy is whether the reduced models remained equally able to discriminate tests with injury from those without. As shown in Table 6, both Model 1 and Model 2 have good utility as injury discriminators. Comparison of the performance of

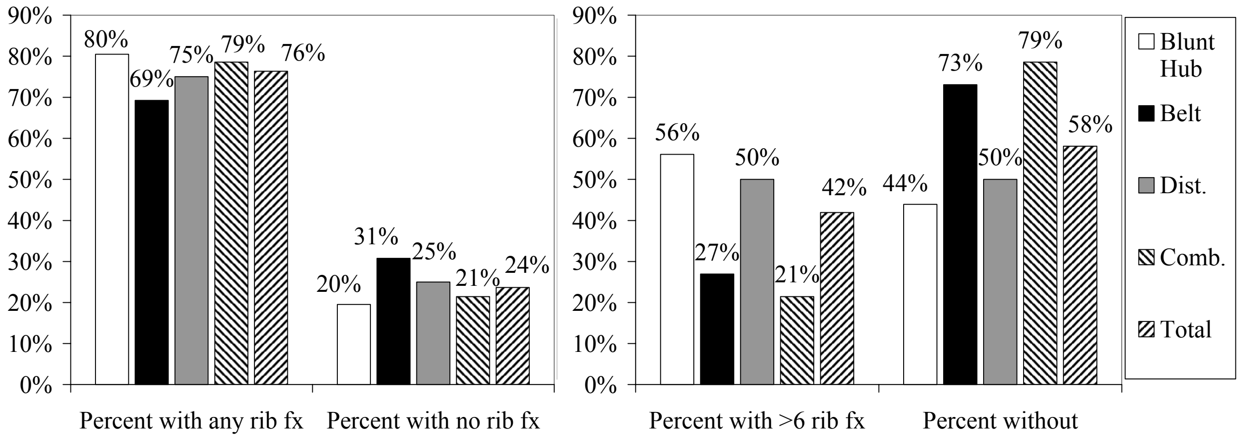


Figure 3. Injury distribution by loading condition for the 93 tests used in this analysis.

Table 2. Model Coefficient Estimates (β) and Standard Errors (SE) for Model 1 and Model 2

Coefficient	Model 1 (rib fx. >6) Estimate (β)	SE (β)	Model 2 (rib fx. >0) Estimate (β)	SE (β)
Intercept	-7.0223	2.7625	-14.6737	5.2166
Age (years)	0.0204	0.0289	0.1870	0.0785
Gender=male	-5.1708	3.1596	6.0046	4.2944
Load Cond. = Seatbelt	-0.1178	1.0003	1.4037	1.0143
Load Cond. = Distributed	0.0895	0.7581	1.4564	0.9362
Load Cond. = Combined	-0.3674	1.0413	2.5619	1.6376
C_{max} (%)	0.1781	0.0499	0.1875	0.0503
Age x Gender=male	0.0720	0.0497	-0.1273	0.0804

Table 3. ANOVA Wald chi-square Statistics for Model 1 and Model 2

	Model 1 (rib fx. >6)				Model 2 (rib fx. >0)			
Predictor	Chi-Square	DOF	Chi-Square - DOF	p value	Chi-Square	DOF	Chi-Square - DOF	p value
Cmax	12.7631	1	11.7631	0.0004	13.8960	1	12.8960	0.0002
Age	5.8047	2	3.8047	0.0549	9.2504	2	7.2504	0.0098
Gender	3.1399	2	1.1399	0.2081	2.5064	1	1.5064	0.1134
Age x Gender	2.1023	1	1.1023	0.1471	2.9023	2	0.9023	0.2343
Load Cond.	0.1966	3	-2.8034	0.9781	3.6775	3	0.6775	0.2985
Total	21.3644	7	14.3644	0.0033	17.7368	7	10.7368	0.0132

these full models with the reduced models, however, reveals that very little injury predictive ability is lost with the removal of the non-significant covariates. The indicators of discrimination decrease slightly for the injury onset models (Model 2 vs. Model 4) when the non-significant covariates are removed while, for the severe injury outcome (Model 1 vs. Model 3), the performance actually increases slightly when the non-significant covariates are removed.

The injury risk functions from Model 3 and Model 4 are plotted in Figure 5. The top plot shows the risk functions for both injury outcomes along with their confidence intervals. The bottom plot illustrates the age sensitivity of both outcomes.

DISCUSSION

This study has shown that the chest deflection injury threshold is strongly dependent upon the age of the subject. This is true regardless of whether injury onset or severe injury is considered. A 30-year-old has a 50% risk of sustaining one rib fracture at a chest deflection level of 35%. The 50% threshold drops to 13% deflection for a 70-year-old. A 30-year-old has a 50% risk of sustaining more than six rib fractures at a deflection level of 43%, while a 70-year-old can tolerate only 33% deflection before being at 50% risk of more than six rib fractures. This finding is consistent with other studies (e.g., Zhou et al. 1996) and is presumably due to multiple characteristics of aging. First, the failure strain of both cortical and trabecular bone decreases with age.

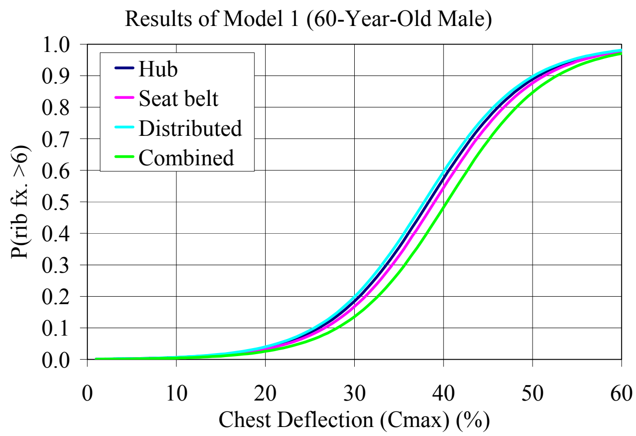


Figure 4. Results of Model 1 showing load condition insensitivity of C_{max} injury threshold.

Table 4. Model Coefficient Estimates (β) and Standard Errors (SE) for Model 3 and Model 4

Coefficient	Model 3 (rib fx. >6) Estimate (β)	SE (β)	Model 4 (rib fx. >0) Estimate (β)	SE (β)
Intercept	-9.3189	2.0965	-6.7508	1.8814
Age (years)	0.0474	0.0215	0.0720	0.0230
C_{max} (%)	0.1838	0.0423	0.1302	0.0345

Table 5. ANOVA Wald chi-square Statistics for Model 3 and Model 4

	Model 3 (rib fx. >6)				Model 4 (rib fx. >0)			
Predictor	Chi-Square	DOF	Chi-Square - DOF	p value	Chi-Square	DOF	Chi-Square - DOF	p value
Cmax	18.8401	1	17.8401	<0.0001	14.2762	1	13.2762	0.0002
Age	4.8644	1	3.8644	0.0274	9.8290	1	8.8290	0.0017
Total	21.4868	2	19.4868	<0.0001	17.7748	2	15.7748	0.0001

Table 6. Predictive Performance of All Four Models

	Model 1	Model 2	Model 3	Model 4
Percent Concordance	86.562	90.269	86.610	87.708
Percent Discordance	13.438	9.731	13.390	12.292
Percent Ties	0	0	0	0
Kruskal's Gamma	0.731	0.805	0.732	0.754
Area under ROC	0.866	0.903	0.867	0.877

Second, geometric changes associated with aging may predispose ribs to fracturing for older subjects under conditions where they might deflect non-injuriously in a younger subject. These geometric changes include a decrease in the proportion of the rib cross-section that is cortical bone and a general decrease in rib slope. Finally, material changes such as calcification of the costal cartilage and decreasing bone mineral density also are likely contributors to the decreased chest deflection tolerance.

The second important finding of this study is that the chest deflection injury tolerance is insensitive to the loading condition. This finding greatly simplifies the relative assessment of different restraint systems

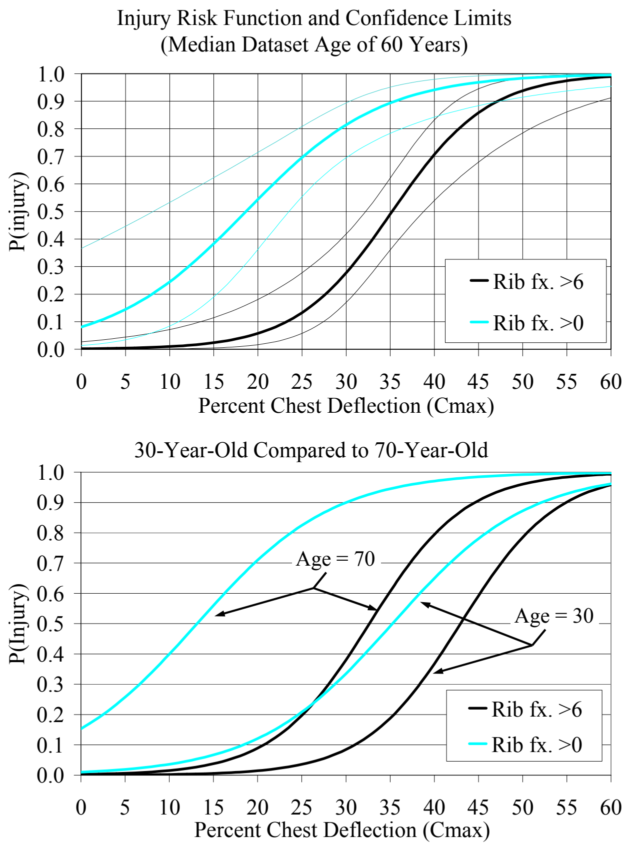


Figure 5. Injury risk functions from Model 3 and Model 4. Top plot shows injury onset risk function and severe injury risk function along with confidence intervals for a 60-year-old. Bottom plot compares injury onset risk and severe injury risk for two ages.

(e.g., belt vs. air bag) since C_{max}, as measured on the cadaver, can be considered to be an objective injury criterion for different restraint conditions (as long as the loading is concentrated through the sternum rather than, for example, through the abdomen and lower rib cage). Unfortunately, we have not shown that chest deflection, as measured by a dummy, has a relationship to injury threshold that is insensitive to the restraint type. Preliminary tests in our laboratory have indicated that the degree to which both the Hybrid III and the THOR dummy are biofidelic depends on the restraint condition. Since the cadaver-based chest deflection tolerance is not sensitive to the restraint type but the force required to achieve a given chest deflection on the dummies is less human-like for some restraints than for others, it follows that the chest deflection injury tolerance, as measured on a dummy, is sensitive to the restraint condition. Unfortunately, we are not yet able to quantify this sensitivity.

As more data become available, it may become possible to develop models that include significant covariates for gender and an age*gender interaction. Future studies should also consider the effect of cadaver mass, body composition (such as changes in the depth of superficial soft tissues), and pulmonary cycle. Furthermore, a C_{max} sensitivity to load condition may eventually be shown once sufficient data are collected to overcome the inherent variability in cadavers, which currently overwhelms any sensitivity to loading condition. Based on differing rib fracture patterns for different restraint conditions and on medical imaging studies such as that illustrated in Figure 2, it is clear that the strain distribution in the rib cage is sensitive to restraint condition. Preliminary studies of rib fracture threshold using a finite element thorax have shown a slight C_{max} sensitivity to restraint condition, with rib fracture onset occurring at slightly (~5%) lower levels of chest deflection under seat belt loading than under distributed loading. Work is ongoing to evaluate the validity of the existing thoracic finite element models for this type of study, however, so this finding should be evaluated in that light. Based on the currently available cadaver data, there is no justification for considering different rib fracture thresholds for belt loading and air bag loading.

CONCLUSIONS

The chest deflection injury tolerance is strongly dependent upon age and this study has quantified this dependence for two levels of injury severity: rib fracture onset and greater than six rib fractures. This study has also shown that the chest deflection injury tolerance is insensitive to the loading condition on the chest within the range of conditions considered (blunt hub, seat belt, air bag, combined belt-and-bag). This insensitivity does not necessarily apply to chest deflection as measured by a dummy, however, since the degree to which both the Hybrid III and the THOR dummy are biofidelic appears to be sensitive to the restraint type. Additional research is needed to quantify how the dummies' chest deflection injury tolerance changes for different restraint conditions.

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Appendix A – Dataset of Cadaver Tests Used in the Analysis

Load Cond.	Cmax (%)	No. Rib Fractures	Age	Gender	Test ID	Reference
Blunt hub	32.9%	9	59	m	172/43fm	Viano (1978)*
Blunt hub	32.1%	0	61	m	171/42fm	Viano (1978)*
Blunt hub	31.5%	10	64	m	177/45fm	Viano (1978)*
Blunt hub	26.9%	9	66	m	200/60fm	Viano (1978)*
Blunt hub	25.7%	3	75	m	189/53fm	Viano (1978)*
Blunt hub	37.3%	4	53	m	203/63fm	Viano (1978)*
Blunt hub	37.1%	6	72	m	204/64fm	Viano (1978)*
Blunt hub	39.3%	9	80	m	69/15fm	Viano (1978)*
Blunt hub	44.4%	12	81	m	65/13fm	Viano (1978)*
Blunt hub	42.0%	14	67	f	61/12ff	Viano (1978)*
Blunt hub	43.5%	6	76	f	66/14ff	Viano (1978)*
Blunt hub	32.8%	6	48	m	104/37fm	Viano (1978)*
Blunt hub	45.9%	11	51	m	93/31fm	Viano (1978)*
Blunt hub	42.5%	16	65	m	86/24fm	Viano (1978)*
Blunt hub	45.8%	13	75	m	94/32fm	Viano (1978)*
Blunt hub	28.9%	17	60	m	47/5fm	Viano (1978)*
Blunt hub	32.5%	11	83	m	50/6fm	Viano (1978)*
Blunt hub	44.7%	11	64	m	96/34fm	Viano (1978)*
Blunt hub	40.7%	7	49	f	190/54ff	Viano (1978)*
Blunt hub	41.8%	11	58	f	85/23ff	Viano (1978)*
Blunt hub	39.5%	10	65	m	87/25fm	Viano (1978)*
Blunt hub	35.0%	6	66	m	219/120fm	Viano (1978)*
Blunt hub	37.0%	10	69	m	218/119fm	Viano (1978)*
Blunt hub	18.5%	0	75	m	88/26fm	Viano (1978)*
Blunt hub	19.4%	0	54	m	92/28fm	Viano (1978)*
Blunt hub	31.0%	3	52	f	92/30ff	Viano (1978)*
Blunt hub	37.7%	0	60	m	187/51fm	Viano (1978)*
Blunt hub	39.4%	3	65	m	192/56fm	Viano (1978)*
Blunt hub	48.6%	9	65	m	188/52fm	Viano (1978)*
Blunt hub	43.1%	10	66	m	186/50fm	Viano (1978)*
Blunt hub	39.0%	4	68	m	196/58fm	Viano (1978)*

Appendix A – Dataset of Cadaver Tests Used in the Analysis (cont.)

Load Cond.	Cmax (%)	No. Rib Fractures	Age	Gender	Test ID	Reference
Blunt hub	39.7%	0	69	m	182/48fm	Viano (1978)*
Blunt hub	37.5%	0	19	m	77/19fm	Viano (1978)*
Blunt hub	35.0%	0	29	m	79/20fm	Viano (1978)*
Blunt hub	41.7%	17	72	m	83/22fm	Viano (1978)*
Blunt hub	41.8%	14	78	m	76/18fm	Viano (1978)*
Blunt hub	31.0%	0	46	m	178/46fm	Viano (1978)*
Blunt hub	34.6%	7	52	m	99/36fm	Viano (1978)*
Blunt hub	40.7%	8	46	f	191/55ff	Viano (1978)*
Blunt hub	37.0%	9	58	m	220/123fm	Viano (1978)*
Blunt hub	41.4%	6	54	m	145	pending
Combined	23.0%	0	57	m	577	Kent et al. (2001a)
Combined	25.0%	4	69	f	578	Kent et al. (2001a)
Combined	34.0%	11	72	f	579	Kent et al. (2001a)
Combined	28.0%	0	57	m	580	Kent et al. (2001a)
Combined	28.0%	3	55	m	665	Kent et al. (2001a)
Combined	32.0%	3	69	m	666	Kent et al. (2001a)
Combined	36.0%	13	59	f	667	Kent et al. (2001a)
Combined	14.5%	1	67	f	533	Shaw et al. (2000)
Combined	18.3%	4	47	m	534	Shaw et al. (2000)
Combined	31.5%	16	57	f	535	Shaw et al. (2000)
Combined	15.7%	3	67	m	545	Shaw et al. (2000)
Combined	26.7%	1	63	f	C11	Kallieris et al. (1995)
Combined	23.8%	2	58	m	C12	Kallieris et al. (1995)
Combined	28.2%	0	50	m	C13	Kallieris et al. (1995)
Seat belt	27.1%	6	60	m	THC 75	Cesari, Bouquet (1994)
Seat belt	26.1%	6	64	f	THC 77	Cesari, Bouquet (1994)
Seat belt	28.7%	3	43	m	THC 79	Cesari, Bouquet (1994)
Seat belt	27.4%	10	63	m	THC 93	Cesari, Bouquet (1994)
Seat belt	25.2%	2	63	m	THC 91	Cesari, Bouquet (1994)
Seat belt	11.8%	0	64	f	THC 76	Cesari, Bouquet (1994)
Seat belt	11.8%	0	43	m	THC 78	Cesari, Bouquet (1994)
Seat belt	10.2%	0	63	m	THC 90	Cesari, Bouquet (1994)
Seat belt	12.1%	0	63	m	THC 92	Cesari, Bouquet (1994)
Seat belt	34.1%	8	47	f	THC 11	Cesari, Bouquet (1990)
Seat belt	35.4%	0	17	f	THC 12	Cesari, Bouquet (1990)
Seat belt	25.1%	2	86	f	THC 13	Cesari, Bouquet (1990)
Seat belt	29.8%	17	69	m	THC 14	Cesari, Bouquet (1990)
Seat belt	29.1%	3	60	m	THC 15	Cesari, Bouquet (1990)
Seat belt	35.4%	4	59	m	THC 16	Cesari, Bouquet (1990)
Seat belt	30.0%	7	71	m	THC 17	Cesari, Bouquet (1990)
Seat belt	9.5%	0	72	m	THC 61	Cesari, Bouquet (1990)
Seat belt	14.4%	0	71	m	THC 64	Cesari, Bouquet (1990)
Seat belt	11.3%	0	40	m	THC 68	Cesari, Bouquet (1990)
Seat belt	36.3%	6	67	m	THC 18	Cesari, Bouquet (1990)
Seat belt	28.4%	4	83	f	THC 19	Cesari, Bouquet (1990)
Seat belt	30.1%	18	70	m	THC 20	Cesari, Bouquet (1990)

Appendix A – Dataset of Cadaver Tests Used in the Analysis (cont.)

Load Cond.	Cmax (%)	No. Rib Fractures	Age	Gender	Test ID	Reference
Seat belt	22.8%	4	72	m	THC 62	Cesari, Bouquet (1990)
Seat belt	36.1%	10	71	m	THC 65	Cesari, Bouquet (1990)
Seat belt	30.6%	1	40	m	THC 69	Cesari, Bouquet (1990)
Seat belt	40.8%	14	63	f	147	pending
Distributed	29.6%	9	61	f	386	Crandall et al. (1999)
Distributed	71.0%	29	45	f	387	Crandall et al. (1999)
Distributed	36.6%	4	34	f	388	Crandall et al. (1999)
Distributed	43.7%	25	68	f	421	Crandall et al. (1999)
Distributed	40.6%	17	67	f	422	Crandall et al. (1999)
Distributed	42.2%	13	51	f	423	Crandall et al. (1999)
Distributed	55.2%	20	55	f	424	Crandall et al. (1999)
Distributed	0.35	4	69	m	116	Kent (2002b)
Distributed	0.35	0	29	f	143	Kent (2002b)
Distributed	0.0%	4	40	m	650	Kent et al. (2001a)
Distributed	11.0%	0	70	m	651	Kent et al. (2001a)
Distributed	12.0%	0	46	m	652	Kent et al. (2001a)

* These tests were originally published earlier, but were compiled in a convenient form by Viano.