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Development of thoracic injury risk functions for the THOR ATD

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A R T I C L E I N F O

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A B S T R A C T

The Test Device for Human Occupant Restraint (THOR) 50th percentile male anthropomorphic test device (ATD) aims to improve the ability to predict the risk of chest injury to restrained automobile occupants by measuring dynamic chest deflection at multiple locations. This research aimed to describe the methods for developing a thoracic injury risk function (IRF) using the multi-point chest deflection metrics from the 50th percentile male THOR Metric ATD with the SD-3 shoulder and associating to post-mortem human subjects (PMHS) outcomes that were matched on identical frontal and frontal-oblique impact sled testing conditions. Several deflection metrics were assessed as potential predictor variables for AIS 3+ injury risk, including a combined metric, called PC Score, which was generated from a principal component analysis. A parametric survival analysis (specifically, accelerated failure time (AFT) with Weibull distribution) was assessed in the development of the IRF. Model fit was assessed using various modeling diagnostics, including the area under the receiver operating characteristic curve (AUC). Models based on resultant deflection consistently exhibited improved fit compared to models based on x-axis deflection or chord deflection. Risk functions for the THOR PC Score and C_{max} (maximum resultant deflection) were qualitatively equivalent, producing AUCs of 0.857 and 0.861, respectively. Adjusting for the potential confounding effects of age, AFT survival models with C_{max} or PC Score as the primary deflection metric resulted in the THOR injury risk models with the best combination of biomechanical appropriateness, potential utility and model fit, and may be recommended as injury predictors.

1. Introduction

The development and refinement of injury criteria and injury risk functions are primary applications of injury biomechanics research (e.g., Viano and Lau, 1988; Kent and Funk 2004; Petitjean et al., 2012). An injury criterion is a measurable parameter or combination of parameters, whose magnitude can be correlated with the risk of an injury occurring. An injury risk function quantifies the relationship between the magnitude of the injury criterion and the probability of a certain injury occurring. Factors that modify this relationship, such as the age or size of an individual, may be incorporated into the function as covariates (Funk et al., 2002; Laituri et al., 2005; Kent and Patrie, 2005). Injury criteria and injury risk functions are used in a range of applications, including the design of safety systems, the retrospective analysis of injury causation in a field event, consumer information testing of vehicles, policy prioritization, and regulation.

1.1. THOR and thoracic injury

The Test Device for Human Occupant Restraint, or THOR, 50th percentile male dummy represents the next generation anthropomorphic test device (ATD) for predicting injury risk for restrained automobile occupants in frontal collisions (Ridella and Parent, 2011). The design of the THOR has seen several iterations throughout its development cycle. The THOR used in this research represents the THOR Metric with SD-3 as defined in Parent et al. (2013); which is a 50th percentile male ATD with metric dimensions and fasteners that includes the SD-3 shoulder assembly. When tested in the blunt thoracic impact condition, the THOR Metric w/SD-3 showed excellent biofidelity (Parent et al., 2013). All subsequent references to "THOR" in this document pertain to this model of the dummy.

The prediction of thoracic injury risk using ATD measurements has been the topic of extensive research, starting with the acceleration-based metrics of Stapp (1951, 1970) and Mertz and Gadd (1971). Based on these studies and the work by Eiband (1959), the National Highway

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Traffic Safety Administration (NHTSA) proposed a peak chest acceleration as a requirement in Federal Motor Vehicle Safety Standard (FMVSS) 208. Deformation-based metrics have also been proposed, starting with Kroell et al. (1971), Kroell and Schneider (1974). These have included the rate of deformation (e.g., Lau and Viano, 1981), the direction and location (e.g., Kuppa and Eppinger, 1998) of deformation, and the nature of the loading environment that caused the deformation (e.g., Horsch et al., 1991; Kent et al., 2003a; Petitjean et al., 2003). The need for ATD-specific thoracic injury criteria has been reported by several researchers (e.g., Kent et al., 2003a; Laituri et al., 2005; Kuppa et al., 2003). The THORAX project of the European Commission Directorate-General for Research and Innovation (DG RTD) Seventh Framework Programme (Hynd et al., 2013) has also conducted research on the development and use of multi-chest deflection parameters in predicting injury using finite element human body model to inform their process (Davidsson et al., 2014).

The purpose of the current study is to describe a method for the development of a function used to predict the probability of thoracic rib injury using multiple chest deflection measurements made with restraint loading of the THOR in frontal and nearside-frontal-oblique sled tests. Past research has identified sensitivity to experimental conditions, such as airbag fitment, seating position, and impact speed, in the relationship between a dummy's measurements and the risk of injury (e.g., Kent et al., 2003a; Petitjean et al., 2003), so a diversity of conditions was considered in the current development. The added chest sensors in THOR increase the number of ways to describe chest deflection and dynamic loading patterns. This manuscript presents the methods behind the development of a new injury risk function that can take in to account the added information from THOR's multi-point chest deflection values and patterns.

2. Methods

The general approach for this IRF development was to match injury outcomes observed in post-mortem human subjects (PMHS) to chest deflection data generated from THOR tests.

2.1. Post mortem human subject testing conditions

Data from previously conducted frontal-impact and nearside-frontal-oblique sled tests performed on un-embalmed (i.e., fresh-frozen) PMHS were reviewed and compiled to complement the THOR testing conditions, and to provide outcome-level data used in the estimated thoracic injury risk function. The general selection criteria for these PMHS tests were based on crash type, restraint system and subject anthropometry, described as follows:

- *Crash type*: Zero degree frontal impact sled tests, and up to 30° nearside-frontal-oblique sled tests, with PMHS
- *Restraint system*: Restrained with a 3-point seatbelt (with or without an airbag), a lap-belt plus an airbag, or a 3-point seatbelt with a belt-integrated airbag.
- *Anthropometry*: Subject height ranging from 150 to 195 cm; mass ranging from 45 to 110 kg (ranges based on available PMHS)
- *Matching Tests*: Test conditions for which matched tests with the THOR are available.

The resulting dataset contained 13 different frontal sled testing conditions listed in Table 1. All attempts were made to include a wide variety of restrained frontal and nearside-oblique sled test conditions in order to cover as large a portion of the real-world exposure as possible, but the number of cases was limited by the availability of experimental data. Detailed descriptions of the test conditions may be found in the noted references, however, a brief summary of the conditions is included in Appendix A.

From these 13 testing conditions, 45 PMHS tests were initially selected for inclusion in the IRF analysis. One observation, a 40 year old male subject tested in condition #3 from Kent et al. (2001a) was removed from final analyses due to outlying effects and questionable pre-existing physical conditions (scleroderma) that may have increased the subject's risk of injury. Therefore, the working database for developing the IRF consists of 44 observations from 40 PMHS that had undergone at least one of the thirteen described test conditions. Four subjects were exposed to two tests in different impact velocity conditions, after determining a non-injurious outcome in the first (low speed) test.

2.2. THOR sled testing

Sled tests were performed with the THOR, matching experimental conditions to the PMHS testing conditions presented in Table 1. THOR chest deflections were measured using InfraRed Telescoping Rod for Assessment of Chest Compression (IR-TRACC) 3D displacement sensors mounted at four locations (upper left, upper right, lower left, lower right) (Rouhana et al., 1998). The raw IR-TRACC signals were processed to calculate displacements in the ATD's x, y, and z directions in the local spine coordinate system (Shaw et al., 2014; Parent et al., 2013). IR-TRACC and rotational potentiometer channels were filtered at CFC 180 in accordance with the THOR Instrumentation Processing Manual (NHTSA, 2005). While SAE J211 (1995) recommends a CFC 600 filter for chest deflections, it is not clear whether this filtering is appropriate for IR-TRACCs and rotational potentiometers used to measure THOR chest deflection. In the interest of robustness, the THOR chest instrumentation data were also processed at CFC 600 and the difference in the relevant output metrics was found to be less than 1%.

Two to three THOR tests were performed for each test condition. Using the maximum x-deflection, the average coefficient of variation among all conditions was 3.4%. The standard deviation of the maximum x-deflection was calculated for the four chest sensors in each test condition, with an overall average standard deviation of 1.29 mm (range: 0.94, 2.12 mm). Together, these values indicate high repeatability in the condition-specific tests and absence of any test anomalies such that the results were averaged to give a single set of chest deflection measures for each test condition.

2.3. Defining experimental injury outcome and chest deflection parameters

The outcome of interest for this study was defined as a serious or greater rib fracture injury, based on the 2008 update of the 2005 version of the Abbreviated Injury Scale (AIS 3+) (AAAM, 2008). This update defines AIS 3+ rib cage injury as any occurrence of three or more fractured ribs (including costal cartilage fractures), unilateral or bilateral. Thus, any PMHS that exhibited three or more fractured ribs was classified as exhibiting an AIS 3+ injury. Note that this most recent definition (AIS 2005/08) does differ from older versions of the AIS. For example, in the 1998 update of AIS1990, absent a corresponding hemo-/pneumothorax at least four ribs must be fractured to be considered a serious (AIS 3+) injury (AAAM, 1998). In all cases, injuries were identified via autopsy.

This study focused on chest deflection as a predictor of injury, as chest deflection has previously been shown to be an indicator of injury with dummy models (Laituri et al., 2005; Kent et al., 2003a; Mertz et al., 1991) and PMHS models (Kent and Patrie, 2005), while also relating to strain in the ribcage both computationally (e.g., Campbell et al., 2007) and experimentally (e.g., Kemper et al., 2011). Chest deflection measures are often analytically evaluated according to the x-axis, resultant, or "D" (chord length) deflections, displayed in Fig. 1. The x-axis deflection is defined as the compression of the ribcage along the x-axis of the local spine coordinate system. The resultant deformation is defined as the square root of the sum of the square changes in position along the three axes (Fig. 1). The "D" deflection value (i.e., change in chord length) can be described as the change in the distance from a point on the anterior ribcage to the origin of the coordinate

Table 1
PMHS test conditions, anthropometry and injury outcome.

Buck	Restraints	Speed (km/h)	Condition Ref. #	Test #	NHTSA Test #	Age	Gender	Height (cm)	Weight (kg)	BMI	# Fx'd Ribs	AIS Code[a]
Gold Standard Driver	Standard Belt	10	10	1397	–	59	F	167	80	28.7	0	0
				1401	–	69	M	178	84	26.5	0	0
				1404	–	60	M	191	81	22.2	0	0
		40	11	1398	–	59	F	167	80	28.7	11	450203.3
				1402	–	69	M	178	84	26.5	13	450213.4
				1405	–	60	M	191	81	22.2	5	450203.3
Gold Standard	Standard Belt (UVA custom)	40	8	1294	9546	76	M	178	70	22.1	6	450203.3
				1295	9547	47	M	177	68	21.7	17	450214.5
				1358	–	54	M	177	79	25.2	10	450203.3
				1359	–	49	M	184	76	22.4	8	450203.3
				1360	–	57	M	175	64	20.9	5	450203.3
				1378	11014	72	M	184	81	23.9	8	450203.3
				1379	11015	40	M	179	88	27.5	8	450203.3
				1380	11016	37	M	180	78	24.1	2	450202.2
	3 kN FL Belt	30	13	S0028	11468	59	M	178	68	21.5	0	0
				S0029	11469	66	M	179	70	21.8	0	0
				S0302	11509	67	M	178	72	22.6	4	450203.3
				S0303	11510	67	M	170	70	24.2	7	450203.3
				S0304	11511	74	M	178	73	22.9	0	0
	3 kN FL Belt	30 (@ 30° oblique)	14	S0313	11518	69	M	173	69	23.1	7	450203.3
				S0314	11519	66	M	172	76	25.8	5	450203.3
				S0315	11520	67	M	177	64	20.5	0	0
1997 Taurus Passenger	FL + PT Belt plus Depowered AB	48	2	577	8371	57	M	174	70	23.1	0	0
				578	8372	69	F	155	53	21.7	4	450203.3
				579	8373	72	F	156	59	24.3	11	450203.3
				580	7374	57	M	177	57	18.2	0	0
	LB + Full Powered AB	48	3	*650*[b]	*8377*	*40*	*M*	*150*	*47*	*20.9*	*4*	*450203.3*
				651	8378	70	M	176	70	22.6	0	0
				652	8379	46	M	175	74	24.1	0	0
	Standard Belt + Depowered AB	48	4	665	8382	55	M	176	85	27.5	3	450203.3
				666	8383	69	M	176	84	27.1	3	450203.3
				667	8384	59	F	161	79	30.5	12	450203.3
	Standard Belt	29	5	1094	–	49	M	178	58	18.3	0	0
				1095	–	44	M	172	77	26.1	0	0
				1096	–	39	M	184	79	23.5	0	0
		38	6	1110	–	44	M	172	77	26.1	0	0
2004 Taurus Rear Seat	Standard Belt	48	7	1262	9337	51	M	175	55	17.9	9	450203.3
				1263	9338	57	F	165	109	40.0	18	450203.3
				1264	9339	57	M	179	59	18.4	9	450203.3
	FL + PT Belt	48	9	1386	–	67	M	175	71	23.2	8	450213.4
				1387	–	69	M	171	60	20.5	1	450201.1
				1389	–	72	M	175	73	23.8	10	450213.4
	Inflatable Belt w/3 kN FL	48	12	1427	–	72	M	173	88	29.2	7	450213.4
				1428	–	69	M	175	69	22.7	0	0
				1429	–	40	M	186	83	24.0	2	450202.2

FL: force limiting; PT: pre-tensioner; AB: airbag; LB: lap belt.
[a] Maximum injury severity to the thoracic rib cage, as per the Abbreviated Injury Scale (2008).
[b] Test #650 removed from analyses due to potential outlying effects.

system on the spine (Fig. 1), and is measured on THOR as the distance between the anterior and posterior attachment points of the respective 3D IR-TRACC assembly. Based on preliminary regression analyses, the resultant deflection consistently produced better model fit compared to the x-axis deflection and chord deflection. Therefore, resultant deflection was retained as the basis for further analysis.

Given that THOR has the ability to record deflection measures at multiple points of the chest that represent the anatomical 4th and 8th ribs (Ridella and Parent, 2011), several measures (and combinations thereof) were considered for their utility in predicting chest injury. One of the target criteria for choosing predictors was that the resulting injury risk function would be symmetric left-to-right, meaning that the injury risk prediction would not be dependent on whether loading is concentrated on the left side vs. the right side of the chest. The practical result of this is that individual deflection terms cannot be combined as isolated independent variables in a regression model − if each deflection term has its own regression coefficient, then there is no constraint to ensure left-right symmetry. Instead, it is necessary to transform those individual deflection terms into composite measures that retain the model's left-right symmetry when model coefficients are assigned. In the case of the combined-deflection model developed here, this was accomplished by combining each left-right deflection pair using a summation term and a difference term, each of which was then treated as a potential predictor variable. Through this, all deflection magnitude information is retained, and model coefficients are constructed in a manner that retains prediction symmetry. To follow are descriptions for each of these deflection metrics, including a brief discussion of their biomechanical interpretation. As noted above, the deformations along coordinate axes were used to calculate resultant measures of deformation at each point. An example of how these deflection values may be used is provided at the end of the results section.

The Maximum Peak Deflection (C_{max}) of the resultant can be defined as the maximum peak deflection magnitude (absolute value) out of the four deflection measurement locations (measured in millimeters).

The Sum of the Upper (or Lower) Chest Deflection (UP_{tot} or LOW_{tot}) is the time independent sum of the maximum left and maximum right deflection magnitudes (e.g., sum of the peak values), for the upper or lower measurement locations, respectively. The sums of these

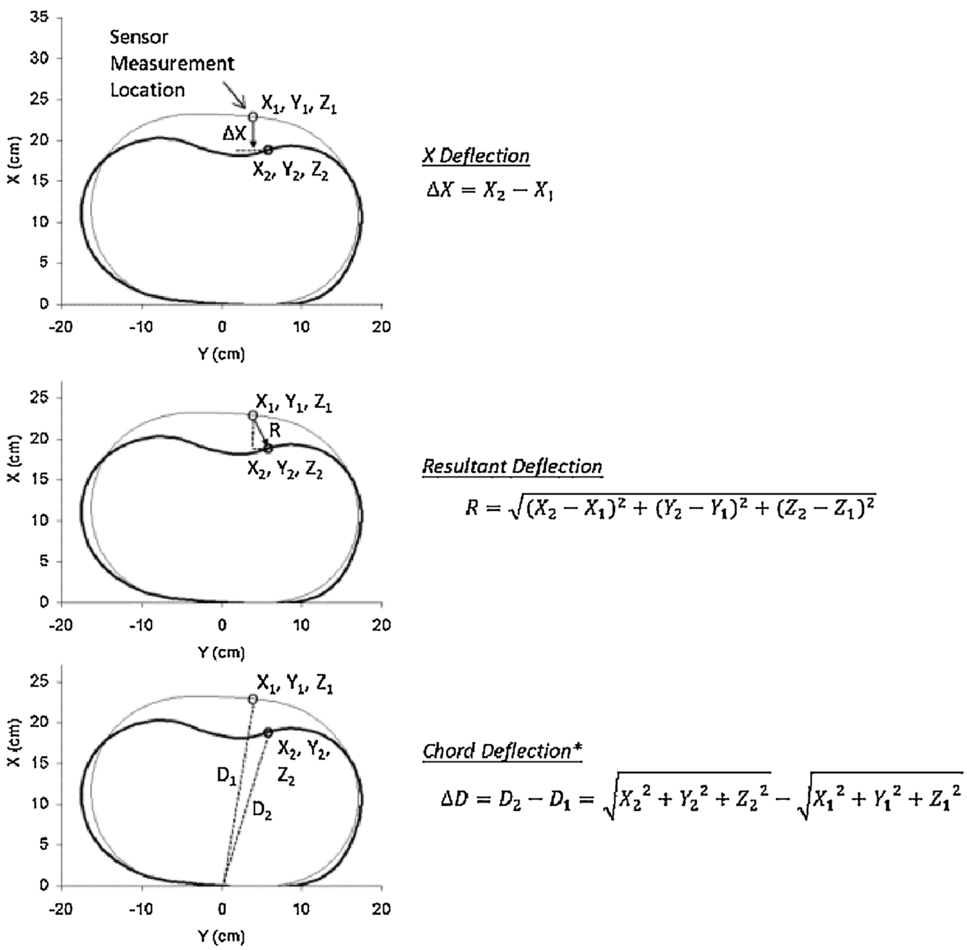

Fig. 1. Illustration of the three methods used for calculating deflection. (* For the chord deflection, note that with the THOR IR-TRACCs the 0,0 reference point for each sensor is located at the base of the sensor, as defined by Parent et al., 2013).

deflection are surrogate indicators of the average of the upper (or lower) maximum deflection.

The Upper (or Lower) Maximum of the Left-Right Difference (UP_{dif} or LOW_{dif}) is equal to the maximum magnitude (absolute value) of the difference observed between the left and right deflections in-phase, for either the upper or lower sensors, regardless if peak values were included (i.e., these values are time dependent). Differential deflection is intended to capture localized curvature of the chest brought about by asymmetric loading.

As the only published composite chest deflection measure for THOR, an evaluation of the deflection metric developed by the THORAX project was also conducted. This metric, termed DcTHOR, is a linear combination of deflection values, which also includes a conditionally-defined term, and can be found in Davidsson et al. (2014). Deflection values from THOR and matched to PMHS frontal sled tests for this research can be found in Appendix B.

2.4. Combined deflection and principal component analysis

In addition to the above, it was desired to explore a combined deflection metric similar to one assessed in the THORAX project that captures the effects of both the average chest deflection across the chest (accomplished with the sum of terms) and the difference in the chest deflection between the left and right aspects, which is analogous to an asymmetry in the loading between the left and right chest. The simplest form of a combined deflection term that results in symmetric injury risk prediction (i.e., injury risk that is independent of the chest aspect to which deflection is applied) would reflect the following:

$$CombinedDeflection = \beta_1(UP_{tot}) + \beta_2(LOW_{tot}) + \beta_3(UP_{dif}) + \beta_4(LOW_{dif}) \quad (1)$$

Note that the generic form of the combined deflection term in Eq. (1) contains four independent deflection terms, with four corresponding coefficients (β_1 through β_4). There is an approximate rule of thumb in regression analysis that each predictor variable requires 10 events (AIS 3+ injuries) and 10 non-events for a reliable model fit (Vittinghoff et al., 2012). Given the sample size of this dataset, the maximum number of independent variables for injury prediction should be limited to two. Age is considered an important confounding factor for injury risk and was deemed an essential predictor in the statistical model. Thus, with only one available term remaining to serve as the primary predictor of chest deflection, it was necessary to reduce the four deflection measures (UP_{tot}, LOW_{tot}, UP_{dif}, LOW_{dif}) into a single term to include in the injury risk function developed from this dataset.

Principal Component Analysis (PCA), a standard technique used to reduce the dimensionality of a dataset, was conducted in an effort to retain as much information about deflection patterns as possible, while limiting the number of predictors needed in the model. PCA describes the directions of maximal variation in the data through an eigen-decomposition of the correlation matrix; the eigenvectors, or principal component directions, describe the direction of variation, and the eigenvalues describe the magnitude of variation in the corresponding directions. PCA was performed to describe the variation in the four variables described above (UP_{tot}, LOW_{tot}, UP_{dif}, LOW_{dif}), and the weighting factors generated by PCA define scalar coefficients for each component, generating a linear combination such as Eq. (1) above.

PCA scores for each data point reflect variation around the sample mean, and therefore contain negative values. Survival (or time-to-event) analyses require inputs to be non-negative. In order to ensure that the resulting PCA scores are non-negative, the raw score were translated to be made positive. This translation (explained in Eq. (3), below) simply shifts the PCA values, ensuring that a value of zero corresponds to zero stimuli (e.g., deflection), realigning the data with the underlying model assumptions. This weighted, combined metric is termed the PC Score, and is described in detail in the results section.

2.5. Statistical analyses and the injury risk function

The association between chest deflection and serious injury was assessed using a parametric survival regression analysis, specifically, an accelerated failure time (AFT) model with a Weibull distribution. The association with injury risk for each of the deflection variables described above was assessed univariately and in an age-adjusted model. Logistic regression models and survival analyses were recently shown to produce similar results and are largely equivalent when data are assumed doubly-censored (McMurry and Poplin, 2015). By treating the outcomes of the four subjects with repeated measures as interval censored, AFT models add information to the statistical model and also meet the assumption that observations are independent of each other.

Statistical model fit and assessment of assumptions was evaluated using standard post-estimation diagnostic strategies for survival regression modeling. Log-likelihood is a measure of model fit that is equivalent to the commonly used Akaike information criterion (AIC) as long as the number of parameters in the model remains fixed. When using the log likelihood, the largest value (in this case the value closest to zero as all values are negative) is indicative of improved model fit. The area under the receiver operating curve (AUC) was also generated and assessed to further discriminate between each statistical model. All statistical analyses were conducted using R Statistical Software (version 3.0.1, Vienna, Austria).

3. Results

3.1. PMHS dataset summary

In the final dataset, males accounted for 87.5% (N = 35) of tested subjects. In general, age and weight were similar between genders, however, due to their shorter height (approximately 6 inches), females had a significantly increased BMI in comparison to male PMHS ($p = 0.033$). Despite the fact that a greater proportion of females (100%) sustained AIS 3+ injuries than males (57%), there were too few female PMHSs (N = 5) to determine any significant difference in injury risk between genders. While 25 of the 40 PMHSs sustained an AIS 3+ injury, Table 2 demonstrates no meaningful differences in physical characteristics between PMHS with and without serious injury.

3.2. Combined deflection metric by principal component analysis

A correlation analysis of the four primary deflection parameters used for PCA (UP_{tot}, LOW_{tot}, UP_{dif}, LOW_{dif}), indicated strong correlation with values ranging between 0.762 and 0.938. The loadings values (or weighting factors) for the individual components in the resulting PCA are listed in Table 3. As indicated, almost all the variability (87.1%) in the test data can be explained by a single principal component. Moreover, that principal component is an approximately equally weighted sum of the four inputs scaled to have mean 0 and variance 1.

Table 2
Differences in anthropometric measures between injury and uninjured PMHSs.

	Total mean (SD)	AIS 3+ mean (SD)	AIS 1–2 mean (SD)	T-test*	Rank-sum†
N	40	25	15		
Age (years)	59.9 (11.0)	62.1 (9.3)	56.2 (12.8)	0.1018	0.1567
Height (cm)	175.2 (7.1)	174.0 (8.2)	177.3 (4.0)	0.1529	0.2175
Weight (kg)	73.1 (10.9)	74.9 (12.2)	70.0 (7.8)	0.1753	0.1499
BMI	23.9 (3.9)	24.8 (4.5)	22.3 (2.2)	0.0441	0.0489

* p-value for two-group mean comparison test using unequal variances.
† p-value for Mann-Whitney *U* test.

Table 3
Loading values (weighting factors) for each variable of the four principle components.

	Component			
	First	Second	Third	Fourth
UPtot	0.486	0.718	0.454	−0.206
LOWtot	0.492	0.132	−0.833	−0.216
UPdif	0.496	−0.664	0.309	−0.466
LOWdif	0.526	−0.160	0.069	0.833
Variance %	**87.1**	**6.4**	**5.4**	**1.1**

3.3. Calculating the principal components score (PC score)

Given that the first principal component explains almost all of the variance in the deflection patterns observed in the data set (87.1%), the weighting factors from this component were used as the basis for a combined deflection term that will serve as the primary injury predictor. The first principal component can be calculated by:

$$PC_1 = l_1\left(\frac{UP_{tot}-m_1}{s_1}\right) + l_2\left(\frac{LOW_{tot}-m_2}{s_2}\right) + l_3\left(\frac{UP_{dif}-m_3}{s_3}\right) + l_4\left(\frac{LOW_{dif}-m_4}{s_4}\right) \quad (2)$$

where l_1 through l_4 are the principal component loadings (first column in Table 3), with values of:

$l_1 = 0.486$ $l_2 = 0.492$ $l_3 = 0.496$ $l_4 = 0.526$,

the means of the four input deflection metrics (described in Eq. (2)) are:

$m_1 = 59.856$ $m_2 = 47.417$ $m_3 = 27.389$ $m_4 = 26.345$,

and the standard deviations of the four input deflection metrics (described in Eq. (2)) are:

$s_1 = 17.439$ $s_2 = 14.735$ $s_3 = 9.672$ $s_4 = 12.384$.

As previously mentioned, the principal component values calculated by Eq. (2) are not guaranteed to be non-negative, which makes them inappropriate for use with a survival model (described below). However, since the principal component loadings are all positive, there is a natural translation to guarantee every value is non-negative and to ensure that zero deflection corresponds to a score and injury risk of zero. The translated principal components score (PC Score) score is given by:

$$\begin{aligned} PCScore &= l_1\left(\frac{UP_{tot}-m_1}{s_1}\right) + l_2\left(\frac{LOW_{tot}-m_2}{s_2}\right) + l_3\left(\frac{UP_{dif}-m_3}{s_3}\right) + l_4\left(\frac{LOW_{dif}-m_4}{s_4}\right) \\ &\quad + l_1\left(\frac{m_1}{s_1}\right) + l_2\left(\frac{m_2}{s_2}\right) + l_3\left(\frac{m_3}{s_3}\right) + l_4\left(\frac{m_4}{s_4}\right) \\ &= l_1\left(\frac{UP_{tot}}{s_1}\right) + l_2\left(\frac{LOW_{tot}}{s_2}\right) + l_3\left(\frac{UP_{dif}}{s_3}\right) + l_4\left(\frac{LOW_{dif}}{s_4}\right) \end{aligned} \quad (3)$$

3.4. Injury risk analyses

When analyzed individually, each deflection metric was positively and statistically significantly ($p < 0.05$) associated with AIS 3+ injury risk in univariate and age-adjusted models. Age was statistically significant in all but one of the models (UP_{tot}, sum of upper maximums). In addition, the beta-coefficients (i.e. effect measure) for deflection consistently increased when age was included in the model, denoting an importance to adjust for the confounding effect of age.

The translated PC Score and maximum resultant peak deflection

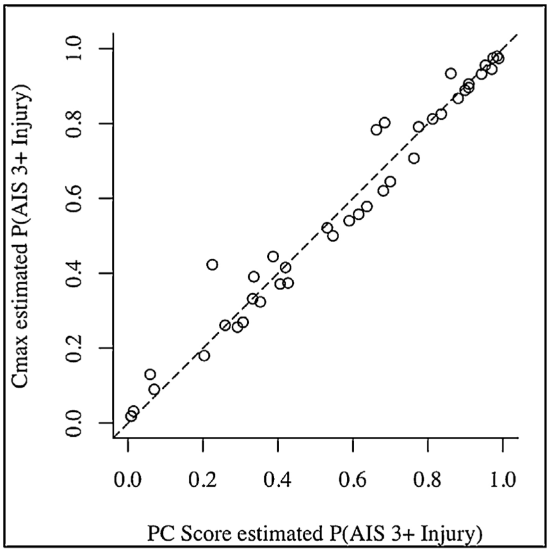

Fig. 2. Correlation plot of predicted injury risk between THOR PC Score and C_{max} injury risk functions.

(i.e., C_{max}) were identified as the independent variables with the best predictive potential and were selected as candidates for primary predictors for further analysis. To test if any of the modelling assumptions were violated with the fitted data, and to identify any potential outlying data observations, post-estimation diagnostic plots for residual versus fitted values, leverage, and scale-location were assessed. AFT models produced non-significant Hosmer-Lemeshow goodness of fit (GOF) statistics, which is an indicator of good model fit. In addition, DFBETA values for the fitted modes were calculated and four subjects were identified as having potentially high influence on the coefficient estimate (test numbers 1095, 1379, 1387, and 1427 in Table 1); however, there was no identifiable physical basis for removing these observations.

As displayed in Fig. 2, AFT models for PC Score and C_{max} produce similar injury risk prediction capabilities, with a Pearson correlation value of (0.987). In addition, PC Score and C_{max} generate comparable areas under receiver operating characteristic curves (AUC) with values of 0.857 and 0.861, respectively. The AUC for DcTHOR was 0.827, using AFT modeling.

3.5. Proposed THOR injury risk models

Based on the considerations above, the current recommended model is an accelerated failure time model with a Weibull distribution (Eq. (4)). It estimates the probability a subject is injured in an event with predictor variable *x*, where *x* is either the PC Score or C_{max} (based on resultant deflection).

$$P(Inj|age, x) = 1 - e^{-\left[\frac{x}{e^{\beta_0+\beta_1 age}}\right]^{\lambda}}, \quad (4)$$

where lambda (λ) is equal to 1/0.302, and 0.302 is the scale parameter value from the survival model output attributed to PC Score (Table 4).

Output from the fitted AFT models, adjusting for age effects and accounting for repeated measures are presented in Table 4 for both PC Score and C_{max}, respectively.

Pointwise 95 percent confidence intervals (CI) were calculated for the combined deflection score needed to produce a specified probability of injury. As illustrated, and for example, the impact required to produce a 50% probability of injury in a 55 year old passenger has a translated PC Score of 5.84, (95% CI: 4.89, 6.96). Similarly, the 50% probability of injury when using C_{max} as the predictor variable equates to a value of 41.6 mm, (95% CI: 35.1, 49.3). The respective injury risk curves and confidence intervals are shown in Fig. 3.

3.6. Example of the PC score IRF use

Suppose a frontal sled test was run using the THOR. The associated AIS 3+ injury risk can be estimated through the following stepwise process:

(1) Calculate UP_{tot}, LOW_{tot}, UP_{dif}, and LOW_{dif} as defined under the “deflection metrics” subsection. For the remainder of the example, we use the values:

$$UP_{tot} = 75.206 \; LOW_{tot} = 51.751 \; UP_{dif} = 33.461 \; LOW_{dif} = 36.753$$

(2) Calculate the translated PC Score as given by Eq. (3). Use the values for UP_{tot}, LOW_{tot}, UP_{dif}, and LOW_{dif} calculated in Step 1, and the values for l_i and s_i given under Eq. (2).

$$PCScore = 0.486\left(\frac{75.206}{17.439}\right) + 0.492\left(\frac{51.751}{14.735}\right) + 0.496\left(\frac{33.461}{9.672}\right) + 0.526\left(\frac{36.753}{12.384}\right) = 7.101$$

(3) Calculate the probability of injury using Eq. (4). Substitute PC Score calculated in Step 2 for *x*, and input the occupant's age. For this example, we assume the occupant is 60 years old.

$$P(AIS3{+}Injury|age = 60, x = 7.101) = 1 - e^{-\left[\frac{7.101}{e^{2.868-0.018*60}}\right]^{1/0.3022}} = 0.833$$

In this example, a 60 year old subject is estimated to have an 83.3% chance of AIS 3+ injury, given the deflections defined above.

4. Discussion

As noted above, this study focused on chest deflection as a predictor of injury since it has been shown in numerous contemporary studies to be an indicator of thoracic injury risk under loading rates and magnitudes consistent with a restrained occupant involved in a frontal collision (Laituri et al., 2005; Kent et al., 2003a; Mertz et al., 1991; Kent and Patrie, 2005; Campbell et al., 2007; Kemper et al., 2011). With THOR’s ability to measure multiple points of chest deflection pattern on a triaxes basis, there are myriad of deflection measures, and combinations thereof, to use as potential explanatory variables for characterizing injury risk. For the THOR, the composite deflection value developed in this research, PC Score (with a resultant basis), produced qualitatively similar results in terms of overall model fit and injury predication as compared to the maximum deflection term, C_{max}.

Age was found to be a significant contributor to thoracic injury risk, and may be examined by observing how age affects the magnitude of association for each of the predictor variables. This is consistent with existing knowledge on the effect of aging on thoracic injury tolerance. It is well established that older individuals tend to exhibit a greater incidence of rib injuries than younger people, including in collisions of reduced severity (e.g., Kent et al., 2005a; Morris et al., 2002; Morris et al., 2003; Kuppa et al., 2005). Kent et al. (2009) and others have attributed this to an increased fragility of the chest and ribs in advanced age. Indeed, ribs undergo a number of material, geometric, and morphological changes with age (such as increased porosity and decreased thickness in the rib cortical bone (Evans, 1975; Lindahl and Lindgren, 1967; Stein and Granik, 1976) that tend to increase the risk of fracture for a given amount of chest deflection (Kent et al., 2005b; Forman et al., 2012; Kent and Patrie, 2005). Interestingly, though, due to counterbalancing changes in the geometry of the ribcage − such as a shallower rib angle − increased age does not appear to affect the overall stiffness of the ribcage (Kent et al., 2005b, 2003b). Overall, age has consistently appeared as a significant covariate in the development of thoracic injury risk functions, when it is considered (Laituri et al., 2005; Kent et al., 2003a). The functions presented may be used as a tool to define injury risk curves for specific target ages, as defined by the end user. For illustration, the injury risk curves presented in the results section

Table 4
Age-adjusted output for THOR PC Score and C_{max} (mm) using the AFT survival model.

Variable	PC Score			C_{max}		
	β-coefficient	Std. Error	p-value	β-coefficient	Std. Error	p-value
Age	−0.0181	0.00785	0.0215	−0.0171	0.00773	0.0271
(Intercept)	2.8677	0.47407	< 0.0001	4.7775	0.46819	< 0.0001
Log(scale)	−1.1975	0.32246	< 0.0001	−1.2101	0.31508	< 0.0001
	Scale parameter = 0.302			Scale parameter = 0.298		
	λ = 1/scale = 3.311			λ = 1/scale = 3.356		
	Log-likelihood:			Log-likelihood:		
	Age-adjusted = −19.2			Age-adjusted = −19.8		
	Intercept-only = −22.6			Intercept-only = −22.9		

N_{PMHS} = 40.

(Fig. 3) represents a 55 year old. Fig. 4 demonstrates complementary risk curves applied to ages 45, 55, and 65.

4.1. Deflection patterns

The process of the principal component analysis provides information on the deflection patterns, specifically the relationship between the four deflection terms used in the PCA. In this particular dataset, there is substantial correlation between each of these measures (as evidenced by dominance of a single component from the PCA, Table 3). As a result, any one of the individual measures, or any combination of a subset of the individual measures, will exhibit similar predictive ability to the others. Thus, it is simply not productive to examine each possible combination as they are bound to give similar results in terms of model fit. By basing the injury risk function on the PCA's linear combination of terms, we remove a majority of the arbitrary predictor decisions from the IRF development and the potential to include overly-influential data points from a single deflection value (e.g., UP_{tot}, LOW_{dif}). Even if not used as the basis for an injury risk function, PCA may be useful as a tool to evaluate deflection patterns observed in future tests and compared to those observed in this dataset. This may provide an objective tool for assessing loading distribution, which may allow for risk prediction adjustment by restraint type in a quantitative (non-categorical) manner, if necessary. In addition, it may be possible to gain insight into the similarities (or differences) in deflection patterns in new loading conditions using the principal component analysis described here.

The first principal component from this research weighted the normalized deflection terms relatively equally (Table 3). To put this in the context of the physical characteristics of the deflection pattern, this first component (which accounted for 87.1% of the variation) is indicative of a chest deflection pattern that contains both an overall increase in the sum of the four deflection measures, as well as an increase in the difference in deflection between the left and right aspect of the chest. This is consistent with the prevalence of 3-point belt restraints in the dataset studied, which tend to cause a combination of distributed and concentrated deflection within the chest (i.e., belt loading tends to cause both an increase in the average deflection across the chest, and a difference in deflection between the left and right chest due to loading concentrated along the belt path).

The second principal component contains positive terms for the upper total and lower total, and negative terms for the upper difference and lower difference. Thus, the second component (accounting for 6.4% of variation) captures loading where there is an increase in the deflection sums, and a decrease in the deflection differences (i.e., an increase in the total summed deflection, but less asymmetry in the left and right deflections). This may be indicative of more evenly distributed loading across the chest.

The third principal component accounts for 5.4% of the variation and may be influential when a greater portion of the deflection is in the lower chest versus the upper chest. This component corresponds to a decrease in upper deflection sum and upper deflection difference, but an increase lower deflection sum and lower deflection difference. Interpreting the fourth component with much certainty is not recommended and not considered meaningful in the current research, as it accounts for approximately 1% of the total variation.

The use of the principle components helps minimize the concern for (over)fitting the model with a single dataset, as it reduces the number of predictors without using the outcome (injury/no injury) data (Hastie et al., 2009). It is important to note that while a new data set would generate slightly different principle component scores, predictions made using the present model with principle component scores calculated as presented would not be invalidated.

4.2. DcTHOR

The previously-developed DcTHOR term resulted in a similar (albeit slightly inferior) model fit and outcome prediction compared to the C_{max} and PC Score terms. The utility of the DcTHOR metric, however, should be considered in the context of its development method. One concern is the conditional terms imposed on an ad hoc basis to improve model consistency across different restraint types and testing conditions (Hynd et al., 2013). The left-right differential deflection terms have two somewhat arbitrary thresholds. Differential deflections of less than 20 mm are treated as producing no additional risk of injury, of which

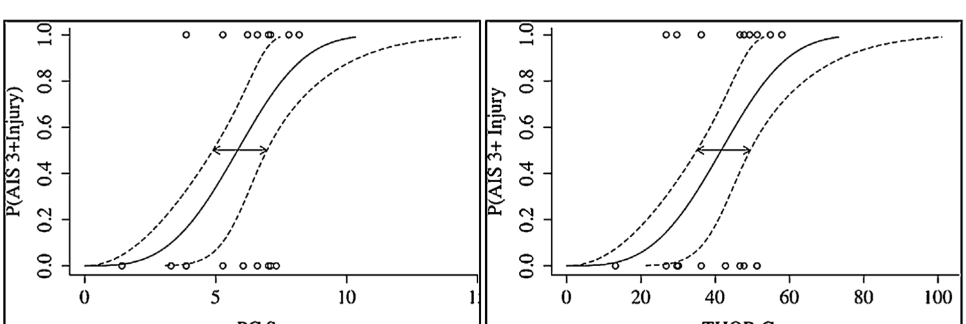

Fig. 3. THOR Injury risk curves for a 55 year old with 95% pointwise horizontal confidence limits for both PC Score and C_{max} (mm). The arrow denotes the confidence interval for a 50% risk of injury.

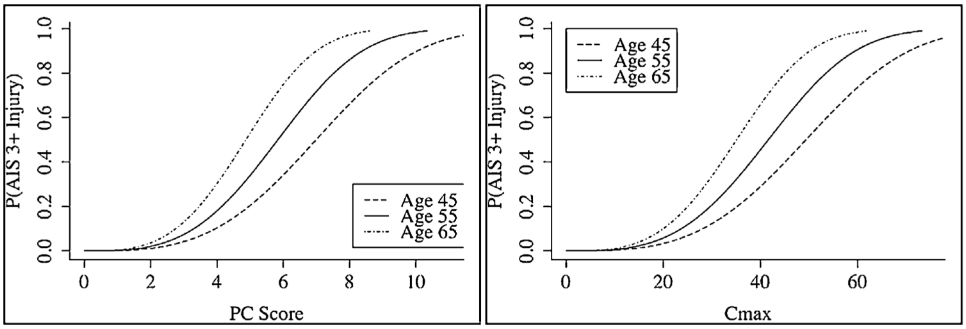

Fig. 4. C_{max} (mm) and PC Score functions for ages 45, 55 and 65 years.

there is no physical reason to assume as true. In addition, if the absolute deflection on either side is less than 5 mm, the differential deflection term is nullified. These thresholds function as additional statistical parameters in the model because they are chosen to improve model fit. While there is nothing wrong with additional parameters per se, they do require additional data observations. Also dissimilar to the PC Score, the DcTHOR term treats the lower and upper deflections identically, even though it is not expected that deflections in the upper ribs produce the same injury risk as deflections of the same magnitude in the lower ribs. Given the ambiguity in the conditional terms included in calculating DcTHOR, it is difficult to define a procedure that can adjust the DcTHOR metric as new data becomes available. In contrast, both the C_{max} and the PC Score injury risk functions described here were developed in a manner that may be refined more readily following the procedures outlined, should new data become available.

4.3. Anthropometry

There are several factors to consider when developing an injury risk function, of which anthropometry is particularly interesting. One may either chose to restrict the observed data to a narrow anthropometry range matching the anthropometry of the dummy used, or may include a wider anthropometry range to allow prediction covering a wider range of the population. While the THOR has an anthropometry of a 50th percentile male, use of the THOR should provide some idea of risk response over a range of potential occupants in the population, including occupants within a reasonable variation in anthropometry. Anthropometry can play a role in individual injury risk, thus subjects with a range of anthropometries were included in the dataset to ensure that the resulting injury risk function is predictive of injury probability within as large a range of the population as is reasonable, and not artificially restricted to a very narrow band of 50th percentile adult males. Undoubtedly, however, there does come a point where anthropometry may be so different as to render injury risk grossly inconsistent with observations made with the THOR (for example, with severely obese subjects or with very small, frail subjects). Instead of trying to identify such cases based on subjective intuition or arbitrary cutoff points, it is more appropriate to evaluate the data for potential outliers using quantitative, objective means. In this study, post-estimation statistical diagnostics were used to evaluate the influence of individual observations, through which only one case was noted as a potential outlier. Based on an examination of the specimen's medical information, it is likely that this specimen exhibited an outlying influence due to an underlying skeletal pathology, and not due to anthropometry. There was no quantitative evidence to warrant exclusion of any specimens based on anthropometry alone. With the methods described here, the effects of anthropometry may be further investigated in future studies with a subset of the dataset provided, by including additional observations, or with datasets covering other types of crash modes.

4.4. Limitations

The PC Score term includes maximum deflection terms that are independent of their relative timing within a test. In other words, as defined, the PC Score term may include peak deflection and maximum difference values that occur at different times in a test. In contrast, an alternative metric could sum all deflection terms at each point in time during a test and identify the point in time that corresponds to a maximum of the combined values. However, if there are meaningful differences in phasing, treating the combined metric purely as time-dependent would, by definition, result in a combined metric of lesser magnitude than would be calculated using the current methodology. In the absence of other information, treating the relationship as time-independent represents a conservative approach for predicting injury risk.

As noted above, sample-size limitations required that the injury risk functions be restricted to two independent variables. Since age was assumed a priori to be a potential confounder (and was confirmed by the results), this left room to include a single independent dummy measure as a potential predictor in each of the investigated injury risk models. Despite the noted sample size limitations, examination of the post-estimation diagnostics indicates that the resulting PC Score and C_{max} injury risk functions model the data in the available dataset relatively well. A true evaluation of the predictive ability of the injury risk functions, however, requires validation against an independent dataset. While limited, there are some independent data available from restrained PMHS sled tests that were not included in current analysis (e.g., Petitjean et al., 2002; Vezin 2002a,b; Rouhana et al., 2003), because matching tests with a THOR ATD of the same design level were not available. Should such matching tests be performed, the resulting data could serve as an independent validation dataset for the injury risk functions developed here.

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Appendix A. Supplementary data

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