Evaluating the Abdominal Response of a Porcine Surrogate to Lap Belt Loading

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ABSTRACT

To better understand the biomechanical response of pediatric occupants undergoing abdominal belt loading, a porcine model (sus scrofa domestica) was developed to represent the abdomen of a 6-year-old human. A custom test fixture was designed to replicate two-point transverse belt loading across the anterior abdomen at rates up to 7 m/s. Five independent parameters were varied—abdominal compression, belt loading velocity, location of belt loading, loading waveform (ramp-hold vs. ramp-release), and the presence of abdominal muscle stimulation—for a total of 21 unique conditions and 47 total dynamic tests. The upper abdomen tests directly loaded the lower ribs, liver, spleen, and stomach, while lower abdomen tests involved direct loading of the small and large intestines. Quasi-static compression tests were also performed to model the force-deflection response of the abdomen with and without active muscle tensing.

The effect of loading location, loading velocity, and the presence of muscle stimulation on the force-deflection response was compared between the quasi-static and the dynamic tests. The upper abdomen produced a stiffer response than the lower abdomen in the quasi-static tests, but this effect was muted at dynamic rates. The effect of active muscle stimulation was similarly noticeable only in quasi-static tests, due to the lower reaction forces. Injury risk functions were created to evaluate the robustness of various predictive criteria, and the Goodman-Kruskal gamma was calculated for each criterion to assess its predictive ability. Force- and compression-based criteria, such as maximum posterior reaction force, maximum belt force, and maximum abdominal deflection, proved to more predictive than velocity or viscous criteria such as $(V^*C)_{max}$*. The most common injuries associated with the upper abdomen tests were rib fractures, liver lacerations, splenic lacerations, and kidney contusions. The lower abdomen tests tended to produce mesenteric lacerations and contusions of the small intestine, and ruptures of the large intestine. These findings will later be used to develop a reusable, biofidelic abdominal insert for the 6-year-old Hybrid III ATD.*

INTRODUCTION

After the head, the abdomen is the second-most commonly injured region in children using adult seat belts, and is associated with significant health care costs and extended hospitalization (Arbogast *et al*., 2005). Loading from adult belts causes a wide array of abdominal and spinal injuries collectively known as "seat belt syndrome" (Durbin *et al.*, 2001), and these injuries patterns occur most frequently during the transition between booster and adult seats (Nance *et al*., 2004). A survey of over 200,000 crashes found that the likelihood of MAIS 2+ injury among 4-8-year-old children is 2.6 times greater than for children 9-15 years old, and almost 25 times than for children 0-3 years old (Arbogast *et al.,* 2004). Seat belt syndrome is commonly attributed to two phenomena: submarining, where the pelvis moves under the belt with the torso reclined, and jackknifing, where the pelvis slides under the belt while the torso flexes forward, wedging the belt between the pelvis and the upper abdomen. The geometry of the pediatric pelvis and abdomen of younger children, and the proportionally higher center of gravity in the pediatric abdomen relative to the adult abdomen, places them at elevated risk for these injuries.

Automotive safety engineers are limited, though, in their tools to address this injury pattern. There currently exists no pediatric dummy capable of quantifying the structural response of the abdomen under belt loading. This is largely due to the lack of available experimental data on the pediatric response of a child's abdomen compared to an adult abdomen. Absent sufficient human cadavers, a suitable animal model must be identified and tested in order to establish the biomechanical response of the pediatric abdomen. Using an approach similar to that described in Rouhana *et al.* (2001), this data can then guide the development of a reusable, biofidelic, abdominal insert for the 6-year-old Hybrid-III dummy. This insert will provide automotive engineers with a pediatric ATD that responds more realistically to belt loading on the abdomen.

METHODS

Porcine Model Development

A porcine *(sus scrofa domestica)* model was developed to examine the abdominal response of a 6-year-old under lap belt loading at collision speeds. Porcine models have been previously used for thoracic and abdominal injury characterization in both adult (Stalnaker et al. 1973, Trollope et al. 1973, Miller 1989) and juvenile (Prasad and Daniel, 1984, Kent *et al.,* 2004, Woods *et al.*, 2002) pigs. The studies by Miller specifically focused on belt loading to the abdomen, but were developed for an adult pig, and the other cited studies involved blunt, hub, or air-bag loading conditions. No studies have been published to date specifically investigating lap-belt loading on juvenile pigs.

Following the methodology outlined in Arbogast *et al.* (2005), an extensive radiology and necropsy study was used to establish a porcine model that corresponds in size and developmental level to a 6-year-old human. Anthropometric measurements and organ masses were taken on 25 porcine subjects, ranging in age from 14 to 429 days (4-10 kg whole body mass), and these measurements were taken as a fraction of the corresponding measurements for the 50th percentile 6-year-old human. These measurements were then compared to anthropometric data on humans taken from a number of sources—the GEBOD database (Snyder *et al.,* 1977), the University of Michigan Anthrokids project (Owings, *et al.,* 1975, Snyder *et al.,* 1977), and the Children's Hospital of Philadelphia (Arbogast *et al*., 2005)—and to mass data compiled by Stocker and Dehner (2002). A multiple-linear regression was used to identify the subject whose age and mass most closely matched the 50th percentile 6-year-old while being constrained to lie along the age/mass trendline of the average pig surrogate. This analysis established a 77-day-old, 21.4 kg subject as the optimal subject for these tests.

Test Procedure

To establish the response of the porcine model to a lap belt loading scenario, 47 porcine subjects were loaded transversely across the abdomen. A custom test fixture was designed and built to produce two-point belt loading across the abdomen of the subject resting supine on the text fixture. A schematic of this test fixture in oblique and lateral views is shown in Figure 1.

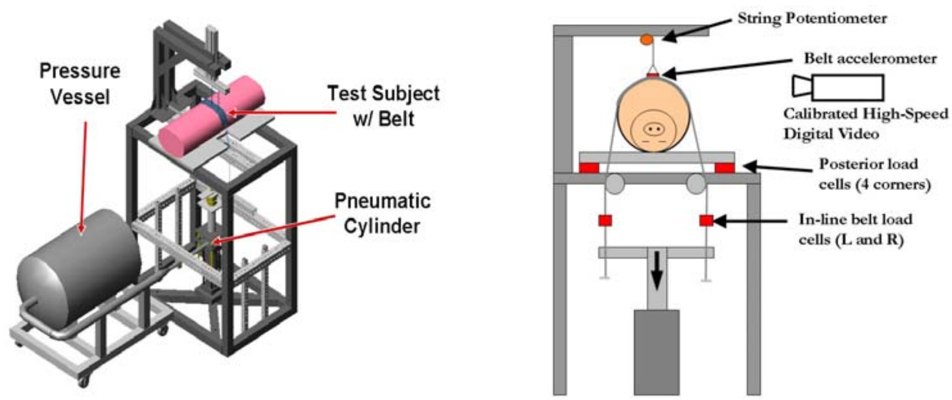


Figure 1: Oblique (left) and lateral (right) view of test fixture, with subject positioned as during the test.

The fixture consisted of a subject table, a piston/cylinder assembly, a pressure vessel, and a nitrogen tank to drive the piston. The belt was attached to either side of a horizontal cross-bar which was attached to the cylinder piston; this ensured symmetric left-right abdominal loading. During the loading sequence, the sealed vessel is pressurized with nitrogen. Once the vessel was filled to the appropriate pressure, an external trigger opened a quick-fire solenoid valve, venting the pressure vessel and causing the cylinder to stroke. This downward piston motion, in turn, compressed the subject's abdomen.

The test fixture was designed to allow five parameters to be varied during the tests:

1. Belt loading velocity. This was nominally controlled by regulating the pressure in the vessel prior to opening the quick-fire valve. The target velocities were 3m/s and 6m/s.
2. Abdominal compression. This was nominally controlled by adjusting the length of permissible stroke of the cylinder piston and the pressure in the vessel. The target compressions were 25%, 50%, and 65% of the subject's total abdominal depth while supine on the test table.
3. Belt loading location. The response due to loading on the upper, primarily solid, organs (liver, stomach) has been shown to be different that loading on the lower, primarily hollow (small intestine, bowel, bladder) lower organs (Rouhana 2002). The belt on the upper abdomen tests was positioned such that the superior edge of the belt was located halfway between the subject's xiphoid process and umbilicus. Lower abdomen tests had the belt positioned such that the belt's inferior edge rested just superior to the illac crest of the pelvis to avoid belt/pelvis entanglement.
4. Loading waveform. The belt loading was varied between displacement ramp-release tests and ramp-hold tests. For the ramp-release tests, the abdomen was unloaded immediately

after the maximum abdominal deflection was achieved (within 0.2 seconds of loading), whereas the ramp-hold tests held the abdomen at the maximum deflection for 120 seconds in order to capture the time-dependent relaxation behavior.

5. The presence of muscle tensing in the lower abdomen. An electrical stimulus was applied to the internal, external, and oblique muscles on the anterior surface of the abdomen on selected tests, to produce active muscle tensing during the loading.

These five parameters were varied in 21 unique combinations over the 47 subjects, with one subject per test. Four load cells were mounted beneath the table surrounding the subject to capture the posterior reaction force produced by the belt. Load cells were placed in-line with the belt, and a string potentiometer and belt accelerometer were mounted at the midline of the outer surface of the belt to capture the belt's displacement and acceleration. Video at 3000 frames per second (fps) was used to capture the displacement, and digital photography was used to extensively document the pre- and post-test conditions.

After each dynamic test, a full necropsy was performed to document and categorize the injuries. Injuries were classified from 0 to 6 using the Association for the Advancement of Automotive Medicine's (AAAM) Abbreviated Injury Scale (AIS - update 1998). During the positioning stages of the test, the subjects were under general anesthesia, and the subjects were euthanized immediately prior to the test. All testing was overseen by personnel from the UVA Center for Comparative Medicine and Department of Emergency Medicine, and all procedures complied with the guidelines of the Animal Welfare Act and Public Health Policy on the Humane Care and Use of Laboratory Animals.

On thirteen of the tests prior to the dynamic tests, the subjects were loaded quasi-statically such that the applied force did not exceed 300N. This force nominally matched that used by Chamouard *et al*. (1996) in human child tests. Chamouard *et al*. loaded six child volunteers (mean age: 6.1 years) using a 2-point lap belt oriented transversally across the lower abdomen. Corridors based on data from the Chamouard *et al*. study were used to assess the correlation of the porcine model to actual human response.

Analytical Methods

To determine the structural response of the abdomen, force-displacement plots were generated for both the dynamic and static tests. Force-displacement plots were generated using both the sum of the posterior reaction forces and the average belt force. Both of these forces were digitally filtered at 200Hz prior to further analysis. The displacement was calculated in three ways: by using the filtered displacement data from the string potentiometer, by integrating the filtered belt accelerometer signal twice, and by tracking the motion of the belt surface over time using the digital video. The velocity was calculated by differentiating the string potentiometer displacement, integrating the belt acceleration, and differentiating the position taken from digital video. The displacement and velocity calculated from the video was used in further analysis of the dynamic data, because the other two measurements of displacement were subject to some error. The string potentiometer tended to overshoot the actual displacement, often significantly, during the loading portion of the dynamic tests. The belt accelerometer was uni-axial, and thus was subject to misalignment errors due to rotation of the accelerometer during loading. Additionally, the integration of the accelerometer to get velocity and displacement was subject to some numerical instabilities. For the quasi-static analyses, the displacement from the string potentiometer was used directly. As there tended to be minimal movement between the anterior surface of the abdomen and the belt during the loading portion of the waveform, the velocity of the top surface of the belt was used as the rate of abdominal compression, *i.e.* the loading rate.

Injury risk functions were developed based on the injuries found during the post-test necropsies, with an "injurious" test being defined as a test which involved an abdominal injury at AIS level 3 or higher. Several injury risk criteria were examined, including maximum abdominal compression C_{max}, maximum compression velocity V_{max}, the viscous criterion $(V^*C)_{max}$ (Lau and Viano, 1986), maximum posterior reaction force (F_{max}), and maximum belt force. Each of the compression and velocity criteria was normalized by the initial abdominal depth, rendering C_{max} in units of %, V_{max} in units of %/s, and $(V^*C)_{max}$ in units of $\%^2$/s. Survival analysis methods were used to generate continuous injury risk functions of the logistic form

$$P(X) = \frac{1}{1+e^{-(\beta_0+\beta_1 X)}}$$

where X is the predictor under consideration and $P(X)$ is the probability of injury. The coefficient β_1 and intercept β_0, and the Goodman-Kruskal gamma (a measure of the injury predictor's ability to adequately discriminate between injurious and non-injurious tests) were calculated and reported for each predictor.

RESULTS AND DISCUSSION

Structural Response Analysis

The quasi-static loading tests generally showed behavior consistent with that expected in humans, despite the anatomic and geometric differences between the porcine abdomen and the human abdomen. The porcine spleen is long and slender and is often described as "tongue-like", while the human spleen is a compact "bean-like" organ. Also, while the human colon has clearly defined ascending, transverse, and descending regions, the porcine colon is wrapped around itself much like the small intestine. Further, the porcine rib cage and pelvis are more narrow relative to the abdominal width than those of humans. The force-deflection plots in Figure 2 (upper left plot) showed that the upper abdomen (red traces) was, on average, stiffer than the lower abdomen (blue traces), consistent with findings published by Rouhana (2002). These force-deflection traces mostly fit within the corridors established for human volunteers by Chamouard *et al.* (1996), shown as black dash-dot lines in the figure. The presence of active muscle tensing (dotted lines) appeared to produce a slight stiffening effect over the unstimulated abdomen (solid lines) after 10-30mm of compression of the lower abdomen (Figure 3). This finding suggests that the porcine abdomen is a reasonable model for the pediatric human abdomen, at least at quasi-static loading rates.

Many of the effects found at quasi-static rates became less pronounced in the dynamic tests. In the dynamic tests, it is difficult to discern the effect of the abdominal loading position; the upper abdomen force-deflection traces straddle the lower abdomen traces at the slow (less than 4 m/s), medium (4-6 m/s), and high (greater than 6 m/s) loading rates (Figure 2). The effect of active muscle stimulation on the abdomen's structural response is similarly somewhat muted. Figure 3 (right) shows the force-deflection characteristics for two tests whose conditions differ only by the presence (dotted lines) or absence (solid lines) of stimulation. While two of the tests presented here show a stiffening effect due to the muscular stimulation, the other two show a negligible effect. This result does not conclusively indicate that active muscle tensing does not play any role in the dynamic response of the porcine model, but rather suggests that structural variability between subjects, even under matched test conditions, contributes an effect that dilutes the effect due to muscle tensing. Also, the significantly higher reaction forces experienced under the dynamic loading condition may be a factor; the benefit gained from the muscle stimulation may be insignificant compared to the significantly larger reaction forces.

All subjects experienced significant rate-dependent stiffening under dynamic loading compared to quasi-static loading. The maximum posterior reaction force for any of the quasi-static tests was 363N, while the posterior reaction forces under dynamic loading ranged from 1297N to 11963N at comparable deflections. Over the experimental range of maximum velocities for the dynamic tests (2.87m/s to 7.75 m/s), the force-deflection response showed no significant trend with increasing maximum velocity (Figure 4). This finding does not conclusively imply that loading rate becomes insignificant at dynamic speeds; inter-subject variability may be obfuscating this trend, especially given the narrow range of velocities tested.

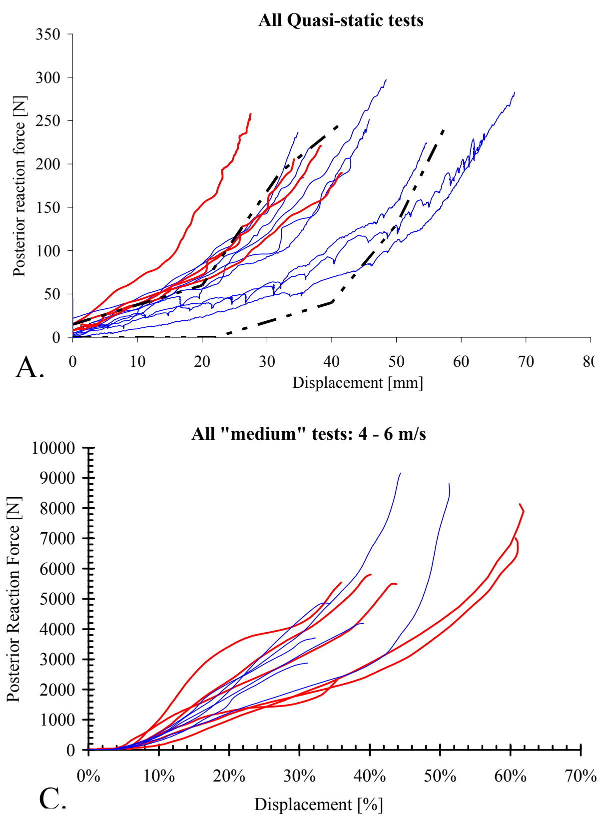


Figure 2: Force-displacement curves for quasi-static and dynamic loading to the upper (red) and lower (blue) abdomen, at (a) quasi-static, (b) slow, (c), medium, and (d) fast rates.

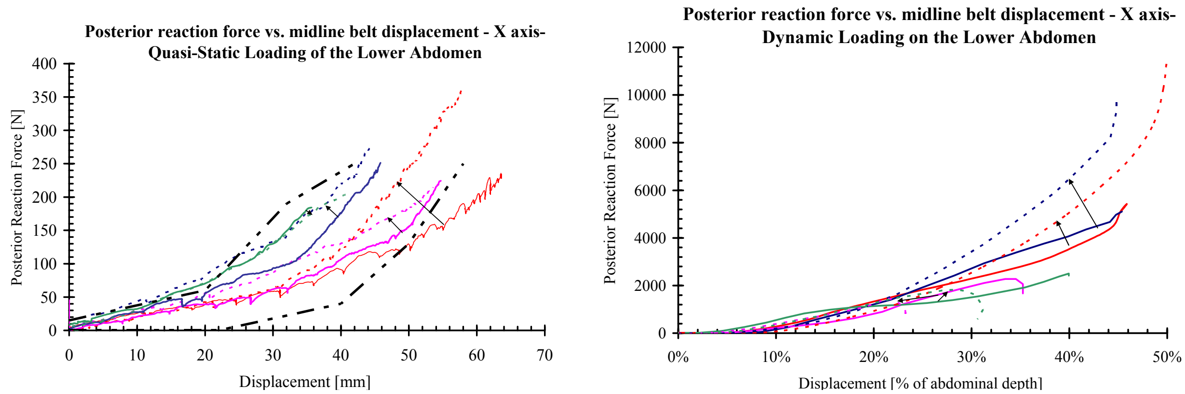


Figure 3: Effect of active muscle stimulation on the
response of the lower abdomen during quasi-static (left) and dynamic (right) loading.

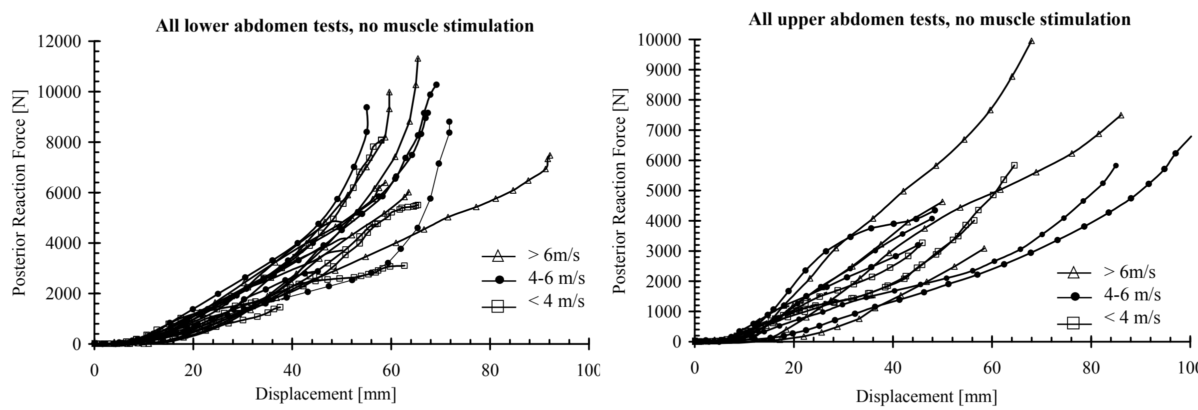


Figure 4: Sensitivity of structural response to velocity
for upper (left) and lower (right) abdomen tests under dynamic loading.

Injury Analysis

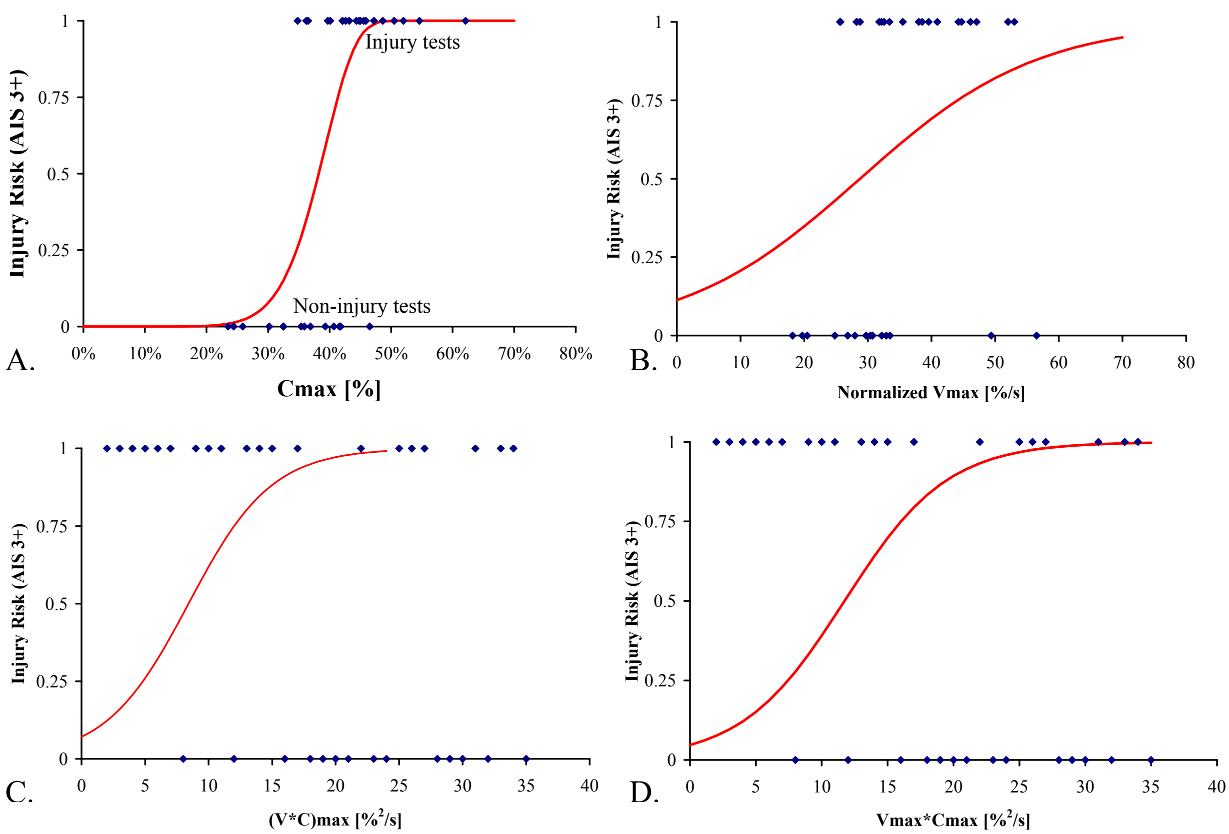
The most common injuries from the dynamic tests were injuries to the spleen, kidneys, and small intestine. The kidneys most often showed a narrow, linear contusion along the border of either the upper or lower pole, but not both. The spleen most often experienced small capsular lacerations and contusions to major stellate lacerations and complete transections (typically at the attachment of the helium). Stomach injuries most often took the form of small petechial hemorrhages along the surface of the greater curvature of the stomach, and most often occurred symmetrically anteriorly-posteriorly. Intestinal injuries ranged from lacerations and contusions of the mesentery to contusions and large perforations (most commonly at the cecum).

Several relevant injury risk functions were developed according to the method above and are reported here (

Table 1 and Figure 5). From examination of the Goodman-Kruskal gamma, velocity-based criteria appeared to be less robust than force- or compression-based criteria. The weakness in velocity-based criteria is perhaps due to the narrow range of velocities afforded by our test setup; this range is much tighter than the 3m/s-30m/s deformation velocity range used by Lau and Viano (1986) in the original development of the viscous criteria.

CONCLUSIONS

This paper presents an intermediate report on the development of a porcine model to study the response and injury patterns of a human pediatric abdomen. A necropsy and radiology study was used to identify the optimally-sized surrogate to represent a 50th 6-year-old human. A test fixture was designed to produce seatbelt loading on the anterior surface of the porcine abdomen was conducted, and 47 porcine surrogates were loaded quasi-statically and dynamically. The sensitivity of loading location and loading velocity to force-deflection response was investigated for quasi-static and dynamic loading rates. The predictive ability of some commonly used injury criteria were examined, and it was found that force and compression based criteria were more predictive than velocity-based criteria. Further statistical analysis of these injury risk functions will identify the contributions of loading velocity and loading location to injury risk. These findings will guide the development of an abdominal insert for the Hybrid-III 6-year-old ATD, providing a robust, reusable experimental device for automotive engineers to design safer seatbelts for children.



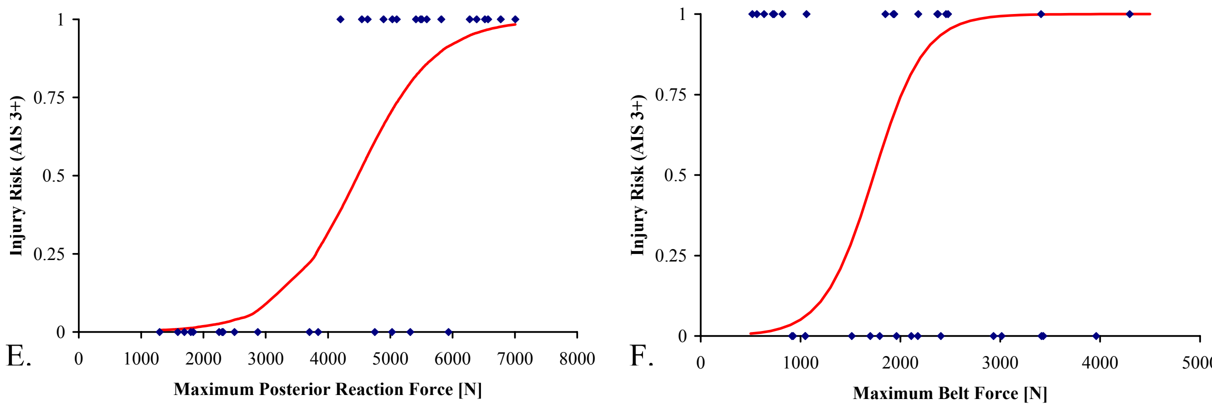


Figure 5: Injury Risk Functions for AIS 3+ injuries, based on (a) C_{max}, (b) V_{max}, (c) $(V*C)_{max}$, (d) $V_{max}*C_{max}$, (e) maximum posterior reaction force, and (f) maximum belt force.

Table 1: Summary of Logistical Injury Risk Function Parameters. *This value is close to the 50% injury risk value of $F_{max}*C_{max}$ = 1.79kN for belt loading on adult pigs reported by Miller *et al.* (1989).

Injury Predictor	Intercept β_0	Coefficient β_1	Goodman-Kruskal Gamma	50% injury risk
V_{max} [%/s]	-2.41457	0.0820631	0.42	28.76
$V_{max}*C_{max}$ [%^{2}/s]	-3.00962	0.25655	0.54	11.73
$(V*C)_{max}$ [%^{2}/s]	-2.57566	0.305938	0.56	8.42
C_{max} [%]	-14.2097	35.8234	0.84	0.40
$F_{max}*C_{max}$ [kN]	-5.45819	0.0031242	0.87	1.75*
F_{max} [kN]	-7.25295	0.0016198	0.87	4.48
Maximum Belt Force [kN]	-6.90658	0.0039834	0.89	1.73

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