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INJURY BIOMECHANICS RESEARCH
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Response of the Pediatric Abdomen to Seatbelt Loading using a Porcine Model

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ABSTRACT

This study presents the development of a porcine (sus scrofa domestica) model to represent the abdomen of a 6-year-old human for biomechanical testing, and utilizes this model to quantify the mechanical response and injury tolerance of the pediatric abdomen to seatbelt loading. The long term goal of this study is to develop a biofidelic abdominal insert for the 6-year-old Hybrid III crash test dummy which will provide automotive design engineers with a tool to quantify the response of the abdomen during belt loading in order to mitigate abdominal injuries in pediatric occupants. Five loading parameters were controlled independently (abdominal deflection, loading rate, loading waveform, active abdominal muscle tensing, and belt location) to yield 47 distinct, repeated tests on 47 subjects. A custom-built loading frame was used to generate forced displacements using a two-point, transversely oriented seatbelt over the anterior abdomen.

*The injuries produced under the test conditions are the same types of seatbelt-induced abdominal injuries that are observed in pediatric occupants in real world automobile crashes. Specifically, the belt location is highly correlated to the type of injury seen, with upper abdominal tests resulting in more frequent liver, and spleen injuries, while lower abdominal tests more frequently generate injury to the large and small intestines. Preliminary examination of the data indicates that belt force, posterior reaction force, and maximum abdominal compression are predictive of abdominal injury. However, belt velocity was not predictive of abdominal injury, and the inclusion of velocity in a mathematical injury criterion (such as V*C) was not an improvement over belt force or maximum abdominal compression alone.*

INTRODUCTION

The abdomen is the second most commonly injured body region in children using adult seat belts and injuries to this region are part of the complex of injuries known as “seat belt syndrome”(Durbin et al., 2001). The fact that children have a proportionally higher center of gravity than adults coupled with improper and uncomfortable restraint “fit,” puts children at particular risk for the jackknife motion attributed to the mechanism of these injuries. The lap portion of the seatbelt is able to “ride up” over the bony pelvis, and load the abdomen directly. Nance et al. (2004) showed a correlation between optimal restraint and the risk of an AIS2+ abdominal injury. This trend is most strongly evident for children 4-8 years old, which corresponds to the period where pediatric occupants are transitioning from booster seats into adult belts. This illustrates the importance of incorporating design features into adult belts that consider pediatric occupants and their increased risk of sustaining a seatbelt-induced abdominal injury.

Automotive design engineers, however, are limited in the tools available in their efforts to design restraint systems that mitigate abdominal injuries to children. No current pediatric anthropometric dummy has the capability to quantify the engineering response of the abdomen attributed to belt loading. This is largely due to the lack of available data on the structural and injury responses of the pediatric abdomen. This project addresses that gap by designing a test methodology to define the biomechanical response of the pediatric abdomen using a well-controlled porcine model. Using an approach similar to Rouhana et al. (2001), the data collected during these tests will guide the development of a biofidelic, reusable abdominal insert for the Hybrid III 6-year old child dummy. An in-depth necropsy and medical imaging study were conducted to establish the correlation between porcine and human age based on several anthropometric measurements. The porcine subjects used in each test were the optimal size and age that best represented a 6-year-old human. It is expected that this information will expand upon and clarify the mechanism and tolerance of seatbelt-induced abdominal injuries.

METHODS

Development of the porcine model

Absent the availability of pediatric human cadavers, a porcine model was used for testing. The porcine model has been used previously for thoracoabdominal injury characterization in both adults (Stalnaker et al., 1973; Trollope et al., 1973; Gogler et al., 1977; Miller, 1989) and children (Aldman et al., 1974; Prasad and Daniel, 1984; Kent et al., 2003a, 2004; Woods et al., 2002). To establish a reasonable model for a 6-year-old human, the methodology outlined in Arbogast et al. (2005) was used. In brief, a series of anthropometric measurements and organ masses on 25 porcine subjects, ranging from 14 to 429 days (4-10 kg whole body mass) were taken. The porcine measurements and masses for each subject were then expressed as a percentage of the corresponding human dimensions and masses, using anthropometric data from 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 mass data compiled by Stocker and Dehner (2002). These percentages were then correlated to age and mass, and a multiple-linear regression approach was used to identify the subject whose age and mass most closely matched 100% of human measurements while being constrained to lie along the actual porcine age/mass relationship. A 77-day-old, 21.4 kg subject was identified as the "optimal" subject for these tests.

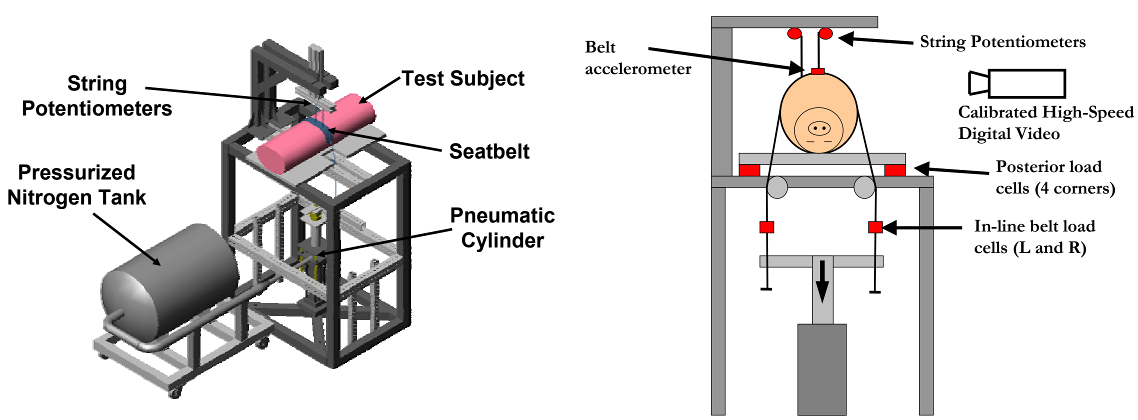

Figure 1: Schematic drawings of the test fixture showing major components and instrumentation.

The test fixture, shown in Figure 1, was designed to produce two-point, transverse belt loading on the abdomen using a 5 cm wide seatbelt. The test subject was placed supine onto the fixture table, and the belt was positioned across the anterior abdomen (the exact position of the belt was dictated by the test matrix). The belt was attached to a loading bar to produce symmetric left-right loading, and the loading bar itself was attached to a pneumatic cylinder beneath the test fixture. The pneumatic cylinder was driven by a pressurized nitrogen tank, which powered the forced belt displacement. A string potentiometer and uni-axial accelerometer were attached to the outer surface of the seatbelt at the subject's midline. The displacement and acceleration signals were differentiated and integrated respectively to calculate belt velocity. Additionally, high speed digital video was captured for each of the tests. The digital imager recorded at 3,000 frames per second (fps), and the position of the seatbelt could be tracked through a calibrated reference frame to yield another measurement of velocity. Two in-line load cells (one on each side) were used to measure the tension in the belt, and there were four load transducers beneath the four corners of the table top which measured the posterior reaction force. Additionally, in several of the tests, pressure transducers were placed in the abdominal aorta, thoracic aorta, aortic arch, and urinary bladder. Digital video and still photography were used to document the test conditions. Figure **2** shows a schematic representation of the instrumentation and nominal positions of the pressure transducers. During the positioning and instrumentation, all subjects were under general anesthesia, and were euthanized immediately prior to the test.

The Institutional Review Board and Institutional Animal Care and Use Committee of the University of Virginia approved the conduct of relevant components of this project. All testing was overseen by personnel from the UVA Center of Comparative Medicine and Department of Emergency Medicine. All procedures comply with the guidelines of the Animal Welfare Act and Public Health Policy on the Humane Care and Use of Laboratory Animals.

Development of the test methodology

Five parameters were identified to control during each test. These factors were:

1. The loading location (upper abdomen vs. lower abdomen). The response due to loading on the upper, primarily solid, organs (liver, spleen) has been shown to be different than loading on the lower, primarily hollow (small intestine, large intestine, bladder) organs (Rouhana, 2002). An upper abdomen test placed the superior edge of the belt halfway between the umbilicus and the xiphoid process. A lower abdomen test placed the inferior edge of the belt just superior to the pelvis, to avoid pelvic engagement.
2. The target abdominal compression. The loading piston was controlled to displace to a nominal magnitude of 25%, 50%, or 60% of the initial abdominal depth.
3. The displacement waveform. A ramp-hold (RH) waveform was used to develop the long-term force relaxation behavior of the abdomen up to 120 seconds after impact, while a ramp-release (RR) waveform was used to develop abdominal stiffness corridors and injury tolerances.
4. The presence or absence of active muscle tensing.
5. The peak compression rate (3 m/s or 6 m/s). The rate of piston stroke, and therefore the rate of belt displacement into the abdomen, was regulated.

In order to maximize information gained from the test series while minimizing the number of test subjects used, 21 unique combinations of the above five parameters were identified and tested. Each of these tests was then repeated, for a total of 42 matched tests. Prior to the loading sequence, a reservoir was pressurized to a known pressure calibrated to match target abdominal deflection and loading rate. To initiate loading, a solenoid valve was opened, allowing the pressurized gas to fill the cylinder and stroke the piston. This piston motion, in turn, loaded the abdomen after some initial "run-up" which allowed the piston to reach the target speed before loading the abdomen. For the ramp-release tests, the abdomen was unloaded immediately after the maximum abdominal deflection was achieved, whereas the ramp-hold tests held the abdomen at the maximum deflection for 120 seconds in order to capture the time-dependent relaxation behavior. A detailed necropsy was performed immediately following each test to identity and document injuries sustained during each test.

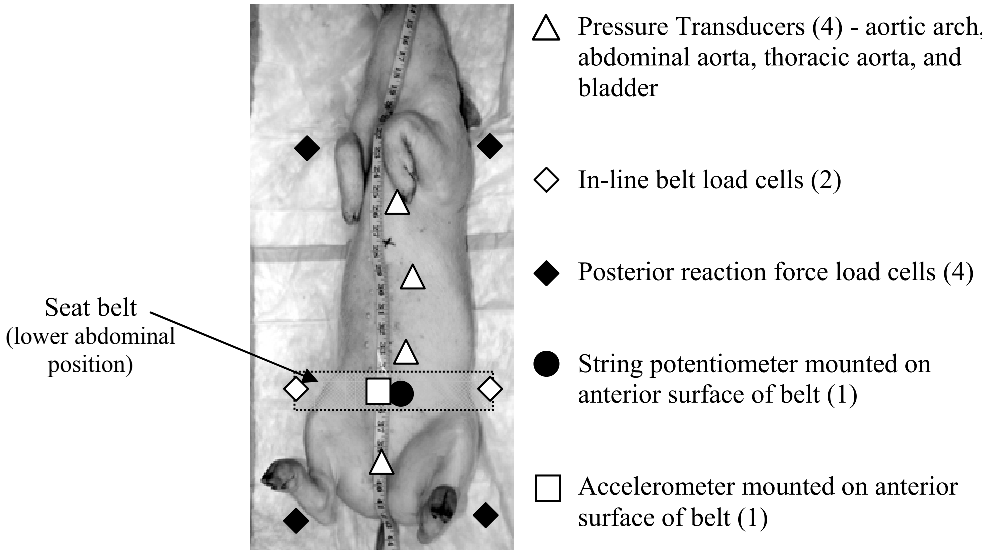

Figure 2: Schematic showing the nominal position of the instrumentation and pressure transducers.

Prior to the dynamic test event, a quasi-static, non-injurious ramp release was performed on approximately one-third of the subjects. Each of these quasi-static tests was limited such that the belt force did not exceed 250 N (approximately the same nominal force Chamouard et al. (1996) used for their child volunteer tests). The purpose of these quasi-static tests was to compare the response of the porcine abdomen to the response of human child volunteers. Chamouard et al. (1996) conducted a series of quasi-static tests with 6 child volunteers (mean age: 6.1 years) using a 2-point seatbelt oriented laterally across the lower abdomen. Corridors representing the upper and lower bounds of the force/deflection plots published by Chamouard et al. (1996) were plotted against the data from our quasi-static porcine tests (Figure 3).

After each test, a detailed necropsy was performed to identify and classify injuries according to the Association for the Advancement of Automotive Medicine's (AAAM) Abbreviated Injury Scale (AIS - update 98). Each injury was recorded and documented with digital photography. Several metrics were evaluated for their potential as injury predictors, including maximum abdominal compression and the viscous criterion $(V*C)_{max}$. Injury risk functions were developed for each metric, with "injury" defined as any injury of AIS 3 or greater. Logistic regression models were used to predict injury given V_{max}, C_{max} and $(V*C)_{max}$. From this, injury risk functions were developed for each injury criterion. Injury risk functions for V_{max}, C_{max} and $(V*C)_{max}$ are shown in Figures 4, 5 and 6, respectively.

Goodman-Kruskal Gamma (γ) was used to measure the degree of utility for each injury criterion, by quantifying each predictor's ability to discriminate injury from non-injury (Kent et al., 2003b). Kruskal's Gamma is calculated by first classifying each pair of observations with different outcomes (injury and non-injury) as either concordant or discordant. A pair is concordant if the model predicts a higher probability of injury for the injurious case than for the non-injurious case. A pair is discordant if the model predicts a lower probability of injury for the injurious case than for the non-injurious case. Gamma is defined by the number of concordant and discordant pairs in the dataset:

$$\gamma = \frac{N_c - N_d}{N_c + N_d} \qquad [1]$$

where N_c is the number of concordant pairs and N_d is the number of discordant pairs (ties are neglected). The closer the gamma value is to one, the greater the predictive ability of the model, where a gamma value of zero indicates the model has no predictive ability and a gamma of one indicates perfect prediction.

RESULTS

The porcine responses shown are for three exemplar cases, and are consistent with human volunteer tests. The force vs. displacement curves for the three porcine subjects fall within the upper and lower bounds of the human child volunteer data published by Chamouard et al. in 1996 which demonstrates the structural and biomechanical similarities of our selected porcine model and a human child to abdominal seatbelt loading.

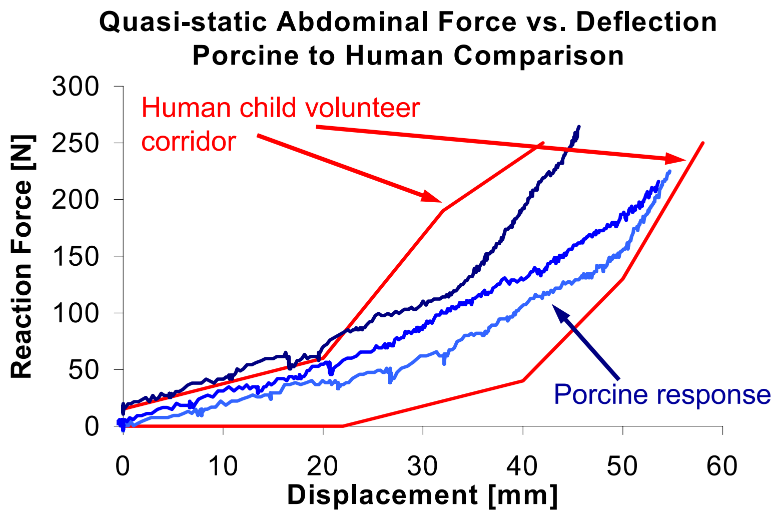

Figure 3: Comparison of quasi-static abdominal response for porcine subjects and human child volunteers.

Considering data from all of the tests, the most common AIS2+ injuries in our test series were to the spleen (24%), kidneys (37%) and the small intestine (42%). The kidneys most often showed a narrow, linear contusion along the border of either the upper or lower pole, but not both. Splenic injuries varied from small capsular lacerations and contusions to major, stellate lacerations and complete transections (typically at the attachment of the hilum). Intestinal injuries ranged from minor hematomas and petechial hemorrhage to large perforations (most commonly seen at the cecum) with significant vessel involvement.

Since there is little relative motion between the seatbelt and the anterior surface of the abdomen, and the belt began at rest, V_{max} is adequately described as the maximum relative velocity between the anterior and posterior surfaces of the abdomen (measured at the midline) during the loading phase of each dynamic test. C_{max} is the maximum abdominal compression as a normalized percentage of the initial abdominal depth. $V_{max}*C_{max}$ is the product of V_{max} and C_{max} and $(V*C)_{max}$ is the viscous criterion, equal to the maximum value of belt velocity multiplied by normalized abdominal compression. $F_{max}*C_{max}$ is the maximum posterior reaction force multiplied by C_{max}, and belt force is the maximum belt tension during the test.

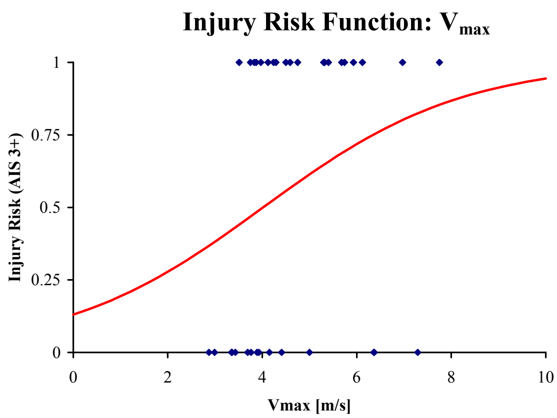

Figure 4: Injury risk function for V_{max}.

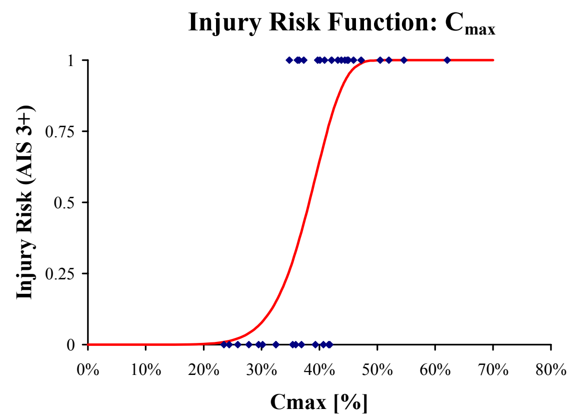

Figure 5: Injury risk function for C_{max}.

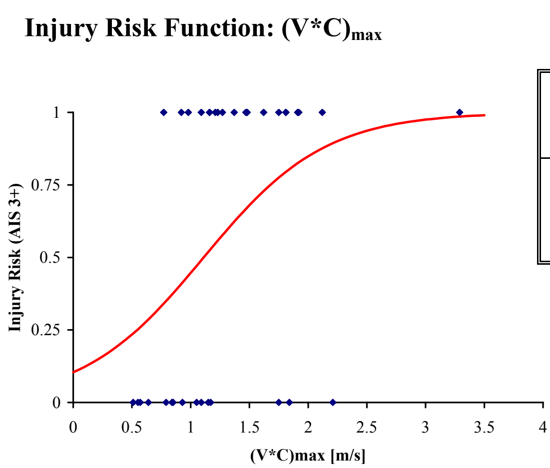

Figure 6: Injury risk function for $(V*C)_{max}$.

Table 1. Results of logistic regression analyses.

Predictor	Intercept α	Coeff. β	Standard Error in β	p
V_{max}	-2.154	1.939	0.8867	0.029
C_{max}	-14.209	35.826	12.734	0.005
$(V*C)_{max}$	-2.218	0.2797	0.1262	0.027

Kruskal's Gamma was computed for each injury criterion, and the results are summarized in Figure 7. Of the seven metrics used as injury predictors, the three with the lowest predictive ability (lowest Kruskal's Gamma) included belt velocity in some manner. The other four predictors (all with Gamma values above 0.80) did not take belt velocity into account.

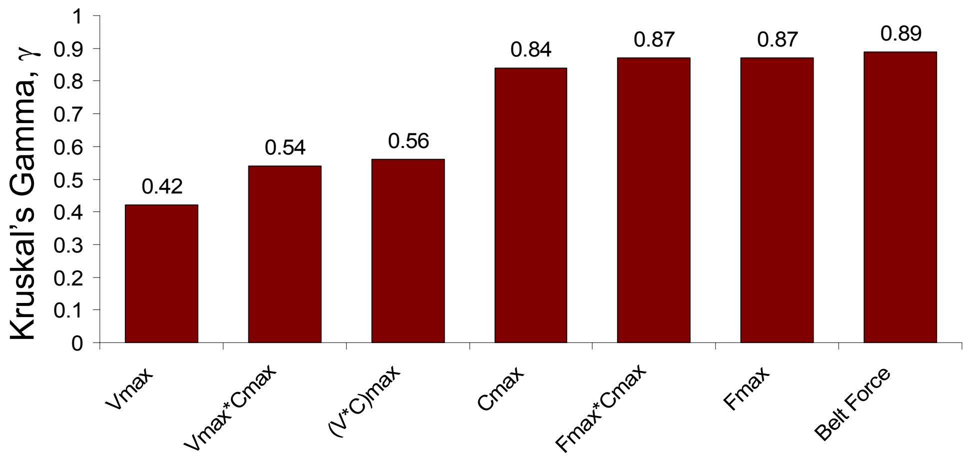

Figure 7: Comparison of predictive ability for several injury criteria.

DISCUSSION

The corridors developed from data published by Chamouard et al. in 1996 (Figure 3), agree well with our data for quasi-static abdominal compression which indicates that the porcine abdomen is a reasonable structural model for the pediatric human abdomen. The belt loading scenario was similar for both cases: the rate of loading was slow (able to be tolerated by child volunteers) and the belt force was limited to 250N to prevent injury. The two major differences in the test series was Chamouard et al. used human child volunteers sitting upright against a rigid posterior surface, where our tests utilized a porcine subject lying supine against a rigid posterior surface. Despite these experimental and postural discrepancies, the quasi-static data from our tests shows very good agreement with the human child volunteer data, supporting the selection of our specific porcine subject as a model for the pediatric human abdomen.

Cooper et al. (1994) compiled and tabulated data over a six-year period (1985-1991) for blunt and penetrating thoracic and abdominal injuries to children. It is interesting to note that the most common blunt injuries in their report were to the liver (28%), spleen (30%) and kidneys (28%). As stated previously, the most frequent injuries we observed were to the spleen (24%), kidneys (37%) and small intestine (42%). Our test matrix was not intended to be representative of the range of crash severities which occur in the field, however the injuries sustained in our tests show correlation to published data of clinically relevant injuries both in terms of type of injury and frequency. Furthermore, despite the geometric differences between the human and porcine abdominal anatomy (particularly the spleen and large intestine), the incidence and severity of abdominal injury seen in our tests closely matches those seen following real world automobile crashes (Cooper at al., 1994), suggesting that the structural differences of the porcine abdomen do not preclude it from being used as a useful model for human abdominal injury and tolerance.

Through our development of a porcine model of the pediatric abdomen in conjunction with a repeatable, controlled, experimental methodology, we were able to simulate abdominal lap belt loading to a child occupant in the laboratory setting. To better understand their contribution to abdominal injury, five parameters were independently controlled during each test: belt location, compression magnitude, compression rate, loading waveform, and muscle tensing. Our upper abdomen tests caused more frequent injuries to the liver, kidneys, and spleen, while intestinal injuries were more frequent when the subject's lower abdomen was loaded. We found that force-based and compression-based injury risk criteria - C_{max}, F_{max}, $F_{max}C_{max}$, and maximum belt force - were the best predictors of AIS 3+ injury in our test series. For our tests, we found that the inclusion of a velocity-based term (such as $(V*C)_{max}$), did not improve our ability to predict injury over a compression- or force-based criterion. This may be due to the relatively narrow range of velocities we measured (2.87 m/s to 7.75 m/s); Lau and Viano's (1985) original viscous criteria development was based on deformation velocities ranging from 3 m/s to 30 m/s using a blunt impactor.

CONCLUSIONS

This paper presents a preliminary report on the development of a porcine model and experimental methodology to study pediatric abdominal injuries. The selected porcine model was shown to have similar biomechanical characteristics to a 6 year old human subjected to seatbelt loading on the anterior abdomen, which indicates that the porcine abdomen is a reasonable structural model for the pediatric abdomen. 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. It is expected that further analysis of the completed test series will expand upon and clarify the mechanism and tolerance of seatbelt-induced abdominal injuries. This, in turn, will be used to guide the development of a reusable abdominal insert for the Hybrid III 6-year-old dummy, and ultimately lead to safer seatbelt designs for children.

ACKNOWLEDGEMENTS

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REFERENCES

ALDMAN, B., ANDERSON, A., and SAXMARK, O. (1974). Possible effects of air bag inflation on a standing child. Proceedings of the International Research Council on the Biomechanics of Impact. Meeting on Biomechanics of Trauma in Children, Lyon, France; September 1974.

ARBOGAST, K., MONG, D., MARIGOWDA, S., KENT, R., STACEY, S., MATTICE, J., TANJI, H., HIGUCHI, K., and ROUHANA, S. (2005). Evaluating pediatric abdominal injuries. Proc. of 19th International ESV Conference, Paper Number 05-0046.

CHAMOUARD, F., TARRIERE, C., and BAUDRIT, P. (1996). Protection of children on board vehicles – influence of pelvis design and thigh and abdomen stiffness on the submarining risk for dummies installed on a booster. Proc. of 15th International ESV Conference, pp. 1063-1075.

COOPER, A., BARLOW, B., DiSCALA, C., and STRING D. (1994). Mortality and truncal injury: the pediatric perspective. J of Pediatric Surgery, 29(1): 33-38.

DURBIN, D. R., ARBOGAST, K. B., and MOLL, E. K. (2001). Seatbelt syndrome in children: a case report and review of the literature. Pediatr Emerg Care, 17(6): 474-477.

GOGLER, E., BEST, A., BRAESS, H., BURST, H., and LASCHET, G. (1977). Biomechanical experiments with animals on abdominal tolerance levels. SAE Paper 770931.

KENT, R., BASS, C., WOODS, W., SHERWOOD, C., MADELEY, N., SALZAR, R., and KITAGAWA, Y. (2003a). Muscle tetanus and loading condition effects on the elastic and viscous characteristics of the thorax. Traffic Injury Prevention, 4(4), 297-314.

KENT, R., PATRIE, J., and BENSON, N. (2003b). The Hybrid III dummy as a discriminator of injurious and non-injurious restraint loading. Proc. of the 2003 AAAM Conference.

KENT, R., BASS, C., WOODS, W., SALZAR, R., and MELVIN, J. (2004). The Role of Muscle Tensing on the Force-Deflection Response of the Thorax and a Reassessment of Frontal Impact Thoracic Biofidelity Corridors. Proc. 2004 IRCOBI.

LAU, I. V. and VIANO, D. C. (1986). The viscous criterion—bases and applications of an injury severity index for soft tissues. Proc. 30th Stapp Car Crash Journal, SAE Paper 861882, 123-142.

MILLER, M. A. (1989). The biomechanical response of the abdomen to belt restraint loading. J. Trauma, 29(11), 1571-1584.

NANCE, M. L., LUTZ, N., ARBOGAST, K. B., CORNEJO, R. A., KALLAN, M. J., WINSTON, F. K., and DURBIN, D. R. (2004). Optimal restraint reduces the risk of abdominal injury in children involved in motor vehicle crashes. Annals of Surgery 2004, 239(1): 127-131.

OWINGS, C., CHATTIN, D., SNYDER, R., and NORCUTT, R. (1975). Strength characteristics of U.S. children for product safety design. University of Michigan Final Report, #FDA-73-32, Prepared for CPSC, Bethesda, MD.

PRASAD, P. and DANIEL, R. (1984). A biomechanical analysis of head, neck, and torso injuries to child surrogates due to sudden torso acceleration. Proc. of the 28th Stapp Car Crash Conference, SAE Paper 841656.

ROUHANA, S. W., ELHAGEDIAB, A. M., WALBRIDGE, A., HARDY, W. N., and SCHNEIDER, L. W. (2001). Development of a reusable, rate-sensitive abdomen for the Hybrid III Family of dummies. Proc. Of the 45th Stapp Car Crash Conference, paper no. 2001-22-0002.

ROUHANA, S. W. (2002). Biomechanics of Abdominal Trauma, in: *Accidental Injury: Biomechanics & Prevention*, A. M. Nahum and J. W. Melvin (eds.). Springer-Verlag: New York, pp. 405-453.

SNYDER, R., SCHNEIDER, L., OWINGS, C., REYNOLDS, H., GOLOMB, D., and SCKORK, M. (1977). Anthropometry of infants, children, and youths to age 18 for product safety design. UM-HSRI-77-17, CPSC, Bethesda, MD.

STALNAKER, R., MCELHANEY, J., and ROBERTS, V. (1973). Human torso response to blunt trauma, in: *Human Impact Response Measurement and Simulation*, Plenum Press: New York, pp. 181-199

STOCKER, J. and DEHNER, L. (2002). *Pediatric Pathology*. Lippincott Williams & Wilkins: Philadelphia.

TROLLOPE, M. L., STALNAKER, R. L., MCELHANEY, J. H., and FREY, C. F. (1973). The mechanism of injury in blunt abdominal trauma. J Trauma, 13(11), 962-970.

VIANO, D. C., and LAU, I. V. (1985). Thoracic impact: a viscous tolerance criteria. Proc. 10th International ESV Conf.

WOODS, W., KENT, R., ULLMAN, E., and BASS, C. (2002). Effect of multiple exhalation ports in a simulation of trantracheal ventilation with a porcine model of an obstructed airway. AAP 2002 National Conference and Exhibition, Boston, Massachusetts.

DISCUSSION

PAPER: **A Preliminary Analysis of the Biomechanical Response of the Pediatric Abdomen to Seatbelt Loading Using a Porcine Surrogate**

PRESENTER: ***Steve Stacey, University of Virginia Center for Applied Biomechanics***

QUESTION: *Stephan Duma, Virginia Tech*
Steve, really nice talk.

ANSWER: Thanks.

Q: I think the methodology here presents some great biomechanical data. The one question I had, and we saw a little glimpse at the ESV. I think Chris, you presented some of this. When you have a step-and-hold test and you end up with an injury from a test that's maybe 2 minutes long, how does that relate to, you know, a car crash? Or, how do you—How can you eliminate the possibility that this isn't from holding deflection for a minute or a minute and a half?

A: Right. The question is, you know, if you load the liver and you get a small laceration and as you hold it over two minutes if that laceration propagates from AIS-2 to an AIS-4 injury is a very good question. For our injury, when we developed our injury risk functions, we only looked at our ramp release test. We didn't use ramp hold tests for the development of our injury risk functions. We did the ramp holds to see more of the time-dependent, force relaxation characterization of the abdomen.

Q: It's perfect and you've got a large test matrix. So each of those—So you basically did one test per animal.

A: Yes. Sorry I didn't mention that. One test per animal and there's a matched—There's one matched test per test conditions. So 42 total tests, 21 individual test conditions if that makes any sense.

Q: Very excellent work. Thank you.

A: Thanks.

Q: *Patricia Fyhrie, General Motors*
One question that I have: Right in the very beginning, you said for children of, you know, I would assume an adolescent (and you didn't discuss that) I'd like, future work on using the 10 year-old hybrid 3. Wouldn't that be a better use of an occupant, you know, a choice to use?

A: We focused on a six year-old primarily because that's one of my first slides showed, I think it was, four to eight year-olds were at the highest risk for abdominal injury from adult belts. So that's where we came up with four to seven—that's where we came up with the six year-old.

Q: Okay.

A: I'm not all that familiar with the development of the 10 year-old hybrid 3, I must admit, but it will definitely be something we could consider in the future.

Q: Okay. Thank you.

Q: *Guy Nusholtz, Daimler Chrysler*
What point are you measuring?

A: For belt deflection?

Q: For belt deflection and velocity.

A: We are measuring—Come on. So, the string pot, as you can see. Or, probably not. So that's the string pot. It's attached to the belt accelerometer at the exact same point. So that's the point that we're measuring, which is the anterior surface of the belt and then in the high-speed video, I can actually track the boundary between the accelerometer and the belt. That's the point that I track when I do point tracking from the video.

Q: Are you getting the displacement from film or from the string pot?

A: Our gold standard is to displace them from the film. There was some significant overshoot from our string pot that we found after our first couple tests. Luckily we had video. I'm confident on our capability to get displacement from video, as well as velocity.

Q: Just a short: Did you try and check the accelerometer against your--?

A: Yes. I did. I don't know—It might be one of my back-up slides. Yes. So as you can see the bottom left—I'm sorry, upper left is a midline belt displacement. So there's a red, a green and a blue line there. The red line is the video and this one you'll see that the string potentiometer, which is the blue line, slightly over predicts the displacement. So in turn when you differentiate that to get velocity, you'll find that the blue line also over predicts the velocity. But we did have, in the majority of our tests, we did have a good agreement between our red line, which was video, and our green line, which was the double integrated—or, singly integrated for velocity of belt accelerometer.

Q: Okay. So, that's a reasonably good check. You're belt correlation is with belt force.

A: It is with belt force.

Q: It is with belt force. Typically in abdominal injuries, either you're looking at some of the solid organs like livers, spleens, and then you're looking at other organs in which you basically have to—and this even happens with the liver: You have to move it far enough so that you tear something. But typically with the solid organs, you do see a rate dependence. And here, what your best correlation is with force. What is the mechanism of injury if force is your best correlant and not displacement or velocity?

A: That's a very good question. An interesting thing that I'd like to do in the future is develop independent risk functions for upper and lower abdomen, and the Kruskal's gamma that I showed were all tested together, upper and lower. I haven't yet been able to look at the effects of individual injury criteria for lower abdomen versus upper abdomen and that's something that I would—that I'm going to do and hopefully that will—Maybe belt force will be the best predictor in lower abdominal injuries. I'm not sure. But I guess to more explicitly answer your question, I've thought a lot about what the injury mechanism could be if force is the best predictor and that's part of the reason that I'm here. Maybe we can give you some good ideas or helpful hints.

Q: It's an interesting phenomenon. Maybe a surrogate for something else and maybe a surrogate for a different type of displacement. All that data has to be sorted out. You've got the data. We don't. Thank you.

A: Thanks.

Q: *Stewart Wang, University of Michigan*
Very nice work. In the clinical scenario when we, that we typically see in these crashes, there's a fair amount of flexion involved. I mean, you're testing in a fully extended position with support along the back. Have you considered, you know, changing your geometry a bit so that there is some flexion? I believe that's gonna contribute significantly to the loading against the spine and again, also contributes to the mobility issue, which—

A: Right.

Q: Contributes to the tearing of the intestines.

A: As well as, you know, chance fractures of the lumbar spine, which are also associates with this type of injury. I hadn't explicitly considered—You mean like making like a V-shaped? So rather than have the pig lie flat, having it lie in a cup or something?

Q: Yeah because I think, you know, in the clinical scenario that can cause a fair amount of focusing but also has a pretty significant determinant given that most of the vascular attachments are superior …and what we oftentimes see is sort of a pulling down.

A: Right.

Q: That would contribute.

A: And the vascular attachments for a human, as we're bipedal, are a little bit different than a pig who, you know, spends most of its life horizontal. It's heart in a slightly different orientation and such; but again like I mentioned, our test fixture was the trade off between being able to accurately and reproducibly input forces and displacements and, you know, what actually happens in a crash, but it's definitely a valid point that we aren't capturing the vast majority of the kinematics that happen during the crash as well as the inertial contributions of force, you know, the organs moving forward into the belt. But, definitely an interesting thing to consider.

Q: Thanks.

Q: *Steve Rouhana, Ford Motor Company*
Just a quick response, too: Ian Lau did some tests with pigs that are upright back in the 80's and found that they didn't tolerate it very well because they're used to spending their life this way, so that's another reason not to test them that way.

A: Okay. Thank you.

Q: *Jean-Pierre Lassau, INRETS, France*
Have you considered measuring the outward pressure within the abdominal aorta? For instance in very old work on pigs, we have found very good correlation with injuries, between injuries and old pressure in the aorta.

A: We have found—Or, we did run some of our tests with pressure transducers not only—We did one in the aortic arch, one in the proximal aorta and one in the abdominal aorta. We also put one in the bladder and the trachea, so we've run some tests with a total of five transducers and we've never been able to correlate that. We've only run five of those tests, so it's tough to get any sort of significance with only five tests. We've never seen any vascular injuries in these pigs, interestingly enough. We've never seen an aortic rupture or anything like that and definitely not in the ones that we've had pressure transducers in, but that is one of the things that we are looking at as kind of a little sub-study. Hopefully on the next five pigs we'll be able to run those with pressure transducers and hopefully get a good number of tests with pressure transducers that we can possibly correlate to injury, as you mentioned.

Q: Thank you.