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Finite Element Model Simulation of Seat Belt-Thorax Interaction

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ABSTRACT

A computational study of thorax interaction with seat belt restraint was performed using a human thorax finite element model. The model consists of the rib cage, chest wall, internal organs, diaphragm and upper abdomen. Simulation was performed to validate the model against selected cadaver data from seat belt loading tests conducted by Cesari and Bouquet (1990) with good chest deformation data agreement. A series of crash simulations was performed for a 3-point belt system with reasonable results obtained showing increase of chest deformation and internal organ stress with crash velocity. The results show that the model is capable of undergoing long-duration deformation engagement with seat belt loads. Limitations of the model are recognized with recommendations given.

INTRODUCTION

The seat belt is acknowledged to be the most important safety feature today with proven safety record. The National Highway Traffic Safety Administration (NHTSA) reported that lap/shoulder safety belts, when used, reduce the risk of fatal injury to front-seat passenger car occupants by 45 percent and the risk of moderate-to-critical injury by 50 percent (NHTSA, 2004). It may be possible that the effectiveness of seat belt systems can be further enhanced. For example, the drivers in Indy-car racing, when protected by 5- or 6-point 3-inch-wide double shoulder belts, observed no chest injuries with some decelerations exceeding 100 g and crash velocities exceeding 72 mph (Melvin et al., 1998). A recent study by Rouhana et al. (2003) suggested that, compared to a 3-point seat belt system, a certain harness style shoulder 4-point belt system was able to shift the load to the clavicles and pelvis to reduce chest deflection by a factor of two, which has a potential to significantly reduce thoracic injury risk, but much more test and model validation work is still needed to generalize the findings.

It is known that the seat belt itself can cause injuries in some cases under crash conditions. Since the 1970s, force-limiting belts have been used to mitigate injuries from belt loading (Foret-Bruno et al., 1978). Recent studies have confirmed that thoracic injuries can be reduced using a force-limiting system (Foret-Bruno et al., 2001; Kent et al., 2001; Petitjean et al., 2002). In addition, given that the driving

population is aging and aging population's tolerance to seat belt loads is greatly reduced (Zhou et al., 1996), continuous research effort is needed to advance the safety and effectiveness of the seat belt system.

In parallel with dummy and cadaver testing, human finite element model (FEM) simulation has great potential to fulfill the need for analytical understanding of the seat belt-occupant interaction dynamics with prediction of rib fracture and tissue-level injuries, but such efforts are still very limited due to the difficulties of modeling long-term soft tissue deformation in car crash conditions. Recent work included that by Ruan et al. (2003) with validation carried out against selected benchmark laboratory data but not for real crash simulations with seat belt effects. Work by Iwamoto et al. (2002) primarily demonstrated the kinematic response of the whole body model meeting standard corridors. While there is often simulation studies using whole body dummy models, such as MADYMO (Rouhana et al., 2003), there is a great need for more advances in whole body simulations using human finite element models. Needless to say, many challenges and difficulties of developing and using human finite element models for crash studies are still not overcome.

The objective of this work is to validate an existing human thorax finite element model against benchmark laboratory data for the study of seat belt load on the thorax and evaluate the model's capability to simulate the thorax response for prediction of rib fractures and internal organ injuries under crash conditions. The thorax finite element model was previously developed for military applications, such as for behind armor blunt trauma, blast, and non-lethal weapon injury studies. Because of the need for predicting tissue-level injuries, extensive effort was carried out to accurately model the ribs, the muscle layers, and the interfaces between solid organs. The model has been validated against frontal and lateral cadaver impact test data for chest deformations (Niu et al., 2003). Results from a computational study using the thorax FEM against seat belt loads are also hereby presented.

METHODS

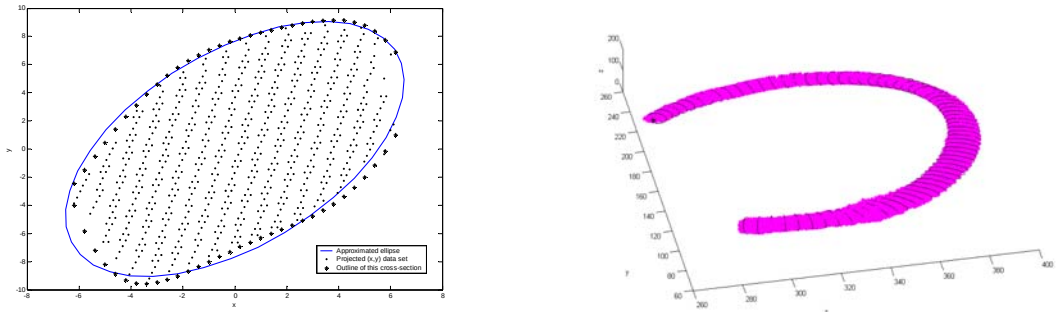
The thorax human finite element model was constructed with anatomical details of the rib cage, chest wall and internal organs using computed tomography (CT) scan medical image data of the torso with the emphasis of predicting tissue-level injuries (Niu, et al., 2003) (Figure 1). The thorax model is slightly shorter than the full torso. The model includes the ribcage, muscle and skin, lungs, heart, diaphragm, stomach, liver, and spleen. The clavicles and scapula, lower abdomen, pelvis, head, neck, and extremities were not included, because they were not needed for the military applications for which the model was originally developed (Figure 1).

(a) Full model (b) Internal organs

Figure 1: Human Thorax Finite Element Model.

One significant feature of the FEM is the use of inhomogeneous beam formulation to accurately model the rib curvature and properties with high computational efficiency and accuracy (Figure 2). The cross section of each rib is modeled as an ellipse described by its long and short principal axes and orientation (Figure 2a). The principal axes, orientation, and material properties of the rib cross section vary along the rib as prescribed by the CT image data (Figure 2b). It has been shown that this inhomogeneous

beam formulation for the ribs has much higher computational efficiency than three-dimensional brick elements, yet yielding better accuracies of rib deformation and internal stresses (Niu et al., 2003).



(a) Elliptical rib cross-section (b) Inhomogeneous rib

Figure 2: Rib Beam FEM Formulation.

Stress/strain-based injury criteria for rib fracture and internal organs have been developed for use with the human thorax FEM based on correlations with swine test data (Niu et al, 2003). To accomplish that, a swine thorax model was constructed using the same methodology as the human thorax FEM. An extensive series of blunt trauma tests using swine subjects undergoing a wide range of well-defined impact loading conditions were carried out. Rib fracture and internal organ injuries were documented using CT scan immediately after loading insult before necropsy. Pathological outcomes were also documented afterwards. Swine FEM simulations were performed for all tests conducted. Stress/strain-based injury criteria were then developed by statistical correlation of the FEM results with the observed injury outcomes for the ribs, lungs, and internal organs.

To model the seat belt, the HyperMesh software was used to fit the seat belt on the thorax similar to a shoulder belt configuration. The belt was modeled to be positioned diagonally across the mid-sternum on the chest with one end crossing over the shoulder and the other end at the waist over the abdomen (Figure 3). All finite element model simulations were performed using Version 9.70 of LS-DYNA3D software (LSTC, 2003). The seat belt-thorax model was validated against benchmark data collected by Cesari and Bouquet (1990). A series of simulations for an idealized 3-point belt system was carried out over a range of crash velocities to evaluate the model response.

Figure 3: Thorax-Shoulder Belt Model.

RESULTS

Model Validation

The human thorax model was first validated against frontal and lateral cadaver impact test data (Niu et al., 2003). The impact tests were conducted by Kroell et al. (1972) and Viano et al. (1989). The measured chest deformations were compared against simulation results. The FEM simulation set up for frontal impact tests is shown in Figure 4. The thorax model was scaled to match a standard 5th and 50th percentile male with 60.1 kg and 73.4 kg mass, respectively, to simulate a wide range of test cases. The summary of the data comparison for peak chest deformation is shown in Table 1 (Niu et al., 2003). The maximum chest deformation is strongly dependent on the subject size. The FEM-predicted rib deformations compare favorably with data. The predicted number of fractured ribs is also consistent with the reported test outcomes (Table 1).

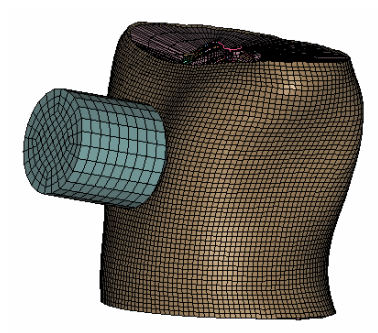


Figure 4: FEM of Hub Frontal Impact on Thorax.

The FEM thorax model was also validated against the laboratory cadaver data measuring chest deflection due to controlled application of the seat belt load (Cesari and Bouquet, 1990). In Cesari and Bouquet's tests, to simulate a driver's shoulder belt, a seat belt strap was placed across the torso of a supine cadaver. The two ends of the belt were routed from the sides of the cadaver down below the test table, where they were attached to a horizontal rigid bar. The bar was pulled down by an impact device that produced dynamic force. Tension forces in the belt were measured at the two ends. These forces were used as belt loads in the FEM model to simulate the belt loading history on the thorax (Figure 5).

The simulated deformation of the ribs and internal organs are shown in Figure 6 with the skin muscle removed. The elongation and deformation of the thorax can be seen in these time-sequence plots when compared to the fixed-length test table (Figure 6). The ribcage and internal organs are compressed by the belt load and reach the maximum deflection at about 55 ms (Figure 6c). The chest deflection results in the simulation were monitored at the mid-sternum, lower sternum, and the right front positions of the 7th and 9th ribs, respectively. These correspond to the reported deflection-time histories for cadaver test #17 (Cesari and Bouquet, 1990). The predicted deflection-time histories show favorable comparison with the reported data (Figure 7). The data comparison for the 9th rib is not as good as the others probably due to the model being too short with the full effect of the lower abdomen not captured (Figure 7d). Excellent data comparisons for the other three locations are observed (Figure 7a-c).

Table 1. Comparison of FEM Prediction With Hub Impact Cadaver Data.

	Test/FEM	Subject Mass (kg)	Impact Mass (kg)	Impact Velocity (m/s)	Peak Deformation (cm)	Rib Fractures
Frontal Impact on Thorax	Test	50.0 ± 18.0	23.34 ± 0.36	6.96 ± 0.30	9.03 ± 1.61	14.1 ± 7.1
	FEM	60.1	23.34	6.96	9.27	15
	FEM	73.4	23.34	6.96	8.76	8
	Test	50.0 ± 18.0	1.75 ± 0.16	12.83 ± 2.34	5.08 ± 0.72	0
	FEM	60.1	1.75	12.83	4.96	0
	FEM	73.4	1.75	12.83	4.77	0
Side Impact on Thorax	Test	67.2 ± 16.2	23.4	4.42 ± 0.86	8.40 ± 1.30	0.4 ± 0.9
	FEM	60.1	23.4	4.42	8.33	1
	FEM	73.4	23.4	4.42	7.37	1
	Test	67.2 ± 16.2	23.4	6.52 ± 0.32	11.20 ± 1.35	5.2 ± 1.5
	FEM	60.1	23.4	6.52	11.15	7
	FEM	73.4	23.4	6.52	10.11	5
	Test	67.2 ± 16.2	23.4	9.33 ± 0.71	14.18 ± 1.79	12.7 ± 4.5
	FEM	60.1	23.4	9.33	14.48	13
	FEM	73.4	23.4	9.33	13.75	13
Side Impact on Abdomen	Test	67.2 ± 16.2	23.4	4.79 ± 0.77	10.83 ± 2.30	0.8 ± 1.6
	FEM	60.1	23.4	4.79	11.41	3
	FEM	73.4	23.4	4.79	10.29	2
	Test	67.2 ± 16.2	23.4	6.83 ± 0.15	11.43 ± 0.76	3.3 ± 3.0
	FEM	60.1	23.4	6.83	12.43	4
	FEM	73.4	23.4	6.83	11.67	6
	Test	67.2 ± 16.2	23.4	9.40 ± 0.87	14.60 ± 2.36	3.8 ± 4.5
	FEM	60.1	23.4	9.40	14.85	5
	FEM	73.4	23.4	9.40	14.38	8

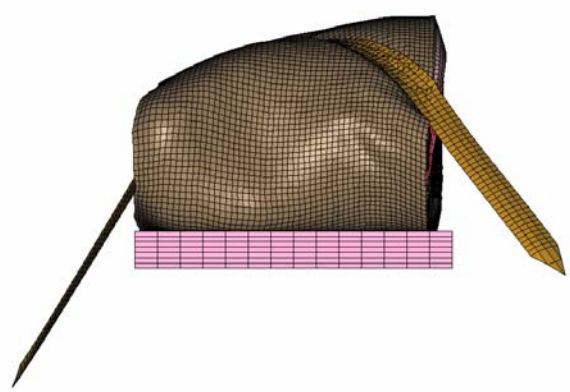
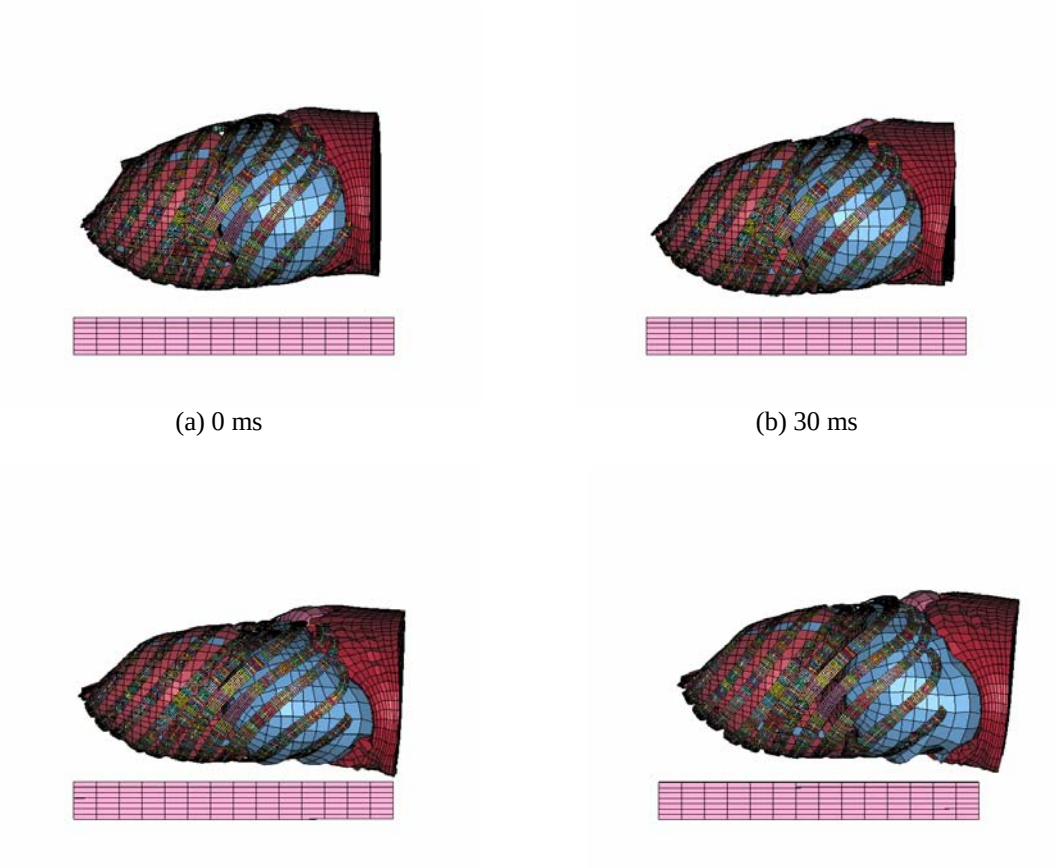


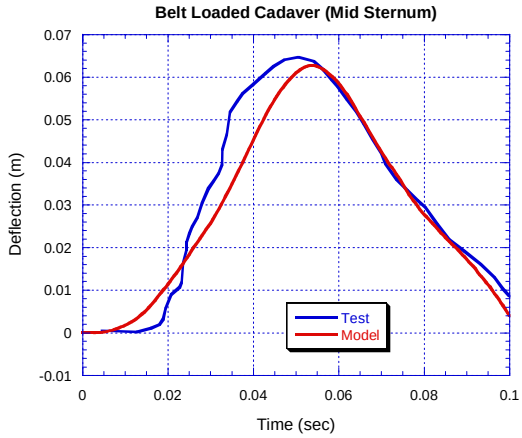
Figure 5: FEM of Seat-Belt Loading Test Setup.



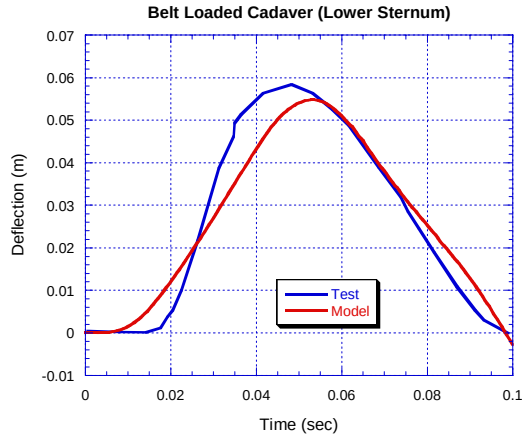
(a) 0 ms (b) 30 ms

(c) 55 ms (d) 80 ms

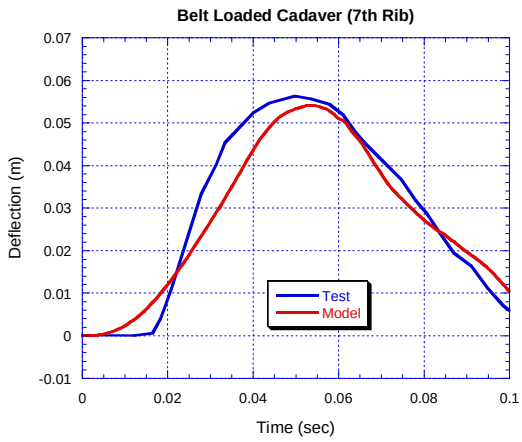
Figure 6: FEM Simulations of Rib and Organ Deformation.



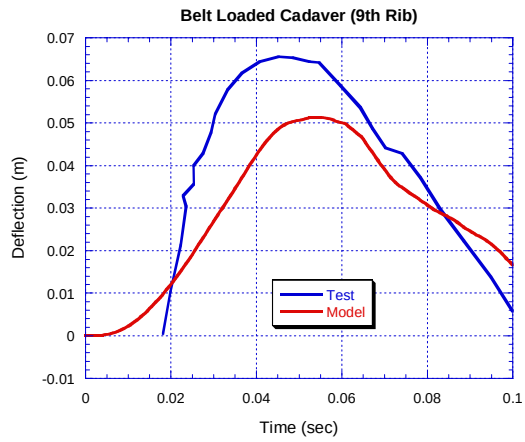
(a) Mid sternum



(b) Lower sternum



(c) 7th rib



(d) 9th rib

Figure 7: Validation of Human Thorax Model Against Cadaver Seat Belt Test Data.

Crash Test Simulation

Crash simulations were performed for an idealized 3-point shoulder/lap belt system with simplified lower abdomen routing contact (Figure 8). This simplified lap belt routing over the abdomen was adopted because the model did not have the lower abdomen, and this routing would probably exert more load on the abdomen than a typical 3-point belt system. The crash condition was simulated with the thorax moving at an initial (relative) velocity against the belts with fixed ends. Simulations were performed for initial velocities of 1.5, 3, and 4.5 m/s, respectively. These relative velocities were estimated based on analysis of vehicle and dummy chest accelerometer data from NCAP tests for typical passenger vehicles under 25-30 mph crash conditions (NHTSA, 1999).

Figure 9 shows the simulated thorax and internal organ deformations for the case with initial velocity of 4.5 m/s. At about 28 ms, the thorax came to a stop and the ribcage and internal organs were significantly deformed by the belt loading with the chest reaching the maximum deflection (Figure 9b). The effects of increasing initial velocity on chest deflection and internal organ pressures are shown in Figures 10 and 11. The thorax with higher initial velocity will push against the belt harder. Consequently, the peak thorax deformation increases from 3 to 16 mm when the velocity increases from 1.5 to 4.5 m/s (Figure 10). Higher initial velocity also results in higher internal organ pressure as larger chest deflection deforms the

internal organs more (Figure 11). The peak pressures in the internal organs at an initial velocity of 4.5 m/s are about 2-3 times higher than those at 1.5 m/s (Figure 11). None of the cases simulated resulted in rib fractures or internal organ injuries.

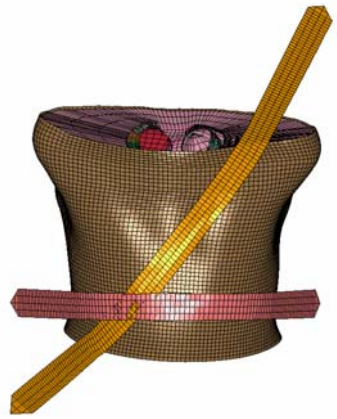


Figure 8: FEM of 3-Point Belt and Thorax Model.

(a) Whole thorax (b) Rib and internal organs

Figure 9: FEM Simulation Results at 28 ms with 4.5-m/s Initial Velocity.

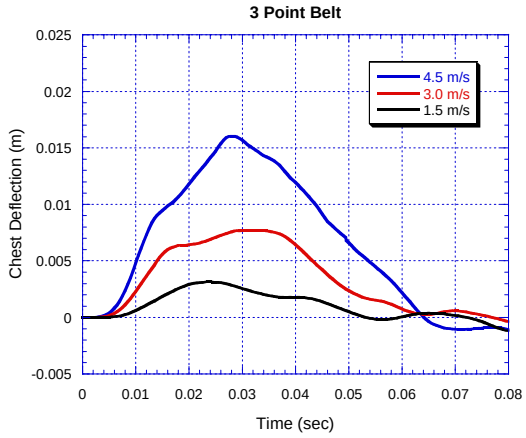


Figure 10: Simulated Effects of Crash Velocity on Chest Deformation.

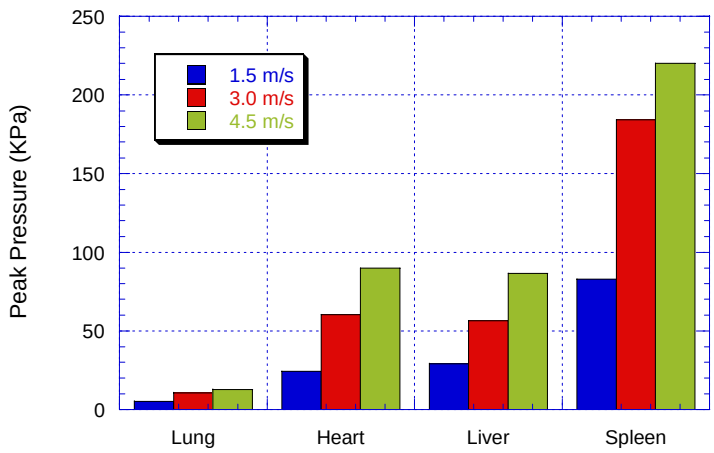


Figure 11: Simulated Effects of Crash Velocity on Internal Organ Pressure.

DISCUSSION

The long-duration whole body engagement with large deformation during crash likely remains a challenge to finite element model simulation of seat belt interaction with the occupant, and more research is still needed to advance the modeling capability. Most studies published emphasize model validation of whole body kinematics and gross deformation response similar to validating a dummy model but that is far from what is needed. The power of finite element modeling should be for the prediction of tissue-level injuries.

The results presented show that the human thorax FEM used is able to simulate the selected benchmark seat belt test reasonably well, and the model is capable of undergoing a long-duration deformation process under simulated seat belt loading condition during crash. The idealized 3-point belt system adopted actually put more load on the abdomen than a typical 3-point belt system. Limitation is recognized that only a very limited range of seat belt load conditions has been studied. The model should be extended to include the full torso geometry so that the full effects of the clavicle and pelvis engagements are captured.

CONCLUSIONS

An existing human thorax finite element model developed for military applications has been used to simulate seat belt load effects on the thorax. Simulation results compare well with selected benchmark cadaver test data. Crash simulations were also performed for an idealized 3-point belt system. The results show that the thorax model is capable of undergoing long-duration engagement with the seat belt loading, predicting gross deformation as well as rib and internal organ tissue stresses. It is recognized that the model should be extended to full torso geometry for seat belt studies in the future.

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DISCUSSION

PAPER: **Finite Element Model Simulation of Seat Belt-Thorax Interaction**

PRESENTER: ***Phil Chan, L-3 Communications/Jaycor***

QUESTION: *Steve Rouhana, Ford Motor Company*
My legs are longer than Guy's. Phil, most of the people I know have clavicles and pelvis and that boundary condition is quite important. So, I'm really surprised that you didn't try and simulate those before you ran your simulations. The other thing is: Our belt system was load-limited, which I don't know that you simulated. Did you simulate a load limit?

ANSWER: Simulate a what?

Q: A load limit in the shoulder belt?

A: No, we didn't. You mean in the seat belt interaction problem? We just fixed the end and then we moved the thorax into the belt.

Q: Okay. Can you comment on what difference you think it would make if you had the clavicle and pelvis as the boundaries?

A: Definitely, like what your paper pointed out, the clavicle and the pelvis would be able to pick up a lot of the loads and we would expect that would make a difference. And, I have pointed out: Yes, we took the model as is and we didn't do any expensive modification of the model. So essentially, our model is kind of like your configuration type of simulation by avoiding a load distribution to the clavicles more concentrated to the chest area. And so given that, even if you change the seat belt contact pattern, you would still see significant differences in the internal organ response and chest deformation. So we realize that the shoulder and the pelvis are not there, as I mentioned up front, but our exercise here is to use the model that had been developed with quite a lot of investment in it and see how it responded. And, we were very encouraged from the favorable comparison to Cesari's data.

Q: Okay. But you did get the wrong result because in our cadaver tests, we showed—

A: I appreciate your comment.

Q: We saw an elimination of just as much.

A: Definitely, we are very aware of what you have found and studied and we intend—This gives us like a confidence to go a few more steps.

Q: Okay. Thank you.

Q: *Guy Nusholtz, Daimler Chrysler*
And this is more continuing along the lines of not having an anatomical structure: There's always a load path that goes from the head and neck and from the pelvis through the lumbar into the thorax. Did you—You just looked at the belt loads for comparison purposes? But did you look at the dynamics of your model?

A: For the comparison to Cesari's data, we looked at the reported deformation of the chest and we applied the belt load as reported in the paper. And for the actual crash simulation, we just fixed the belt endpoints and then we modeled the belt—this belt elasticity we applied based on what we could, based on our best estimate. We just moved the thorax into the belt.

Q: But, the problem is there's a load path from the pelvis.

A: Right.

Q: Coming up to the thorax, so there's an additional—You've only got one load path. You've got the belt loads going into the chest.

A: Right.

Q: And, then you don't have a load path coming up from the pelvis. So I would think that the motion and the angles that thorax makes during the impact, along with the type of loads that you're going to get into the belts, are going to change dynamically, even if you get not only the same forces, you're going to have different type of stress comparison with that. So, it's hard to imagine that you're going to get exactly the same type of kinematic response in that thing. The other thing: One would expect that with two belts, you're going to get higher forces.

A: Right.

Q: One belt—Two belts is twice what one belt is. So you'd expect higher forces and you'd expect higher stress, and you'd expect to stop the thorax faster. So in order to evaluate, then, the difference between a 3-point and a 4-point, you have to adjust the loads and then look at what is the stress distribution across your finite element model. Otherwise, you're sort of predicting what we'd expect: You put two belts on, you get double the force or some increase in the force. You expect to stop the thorax earlier because you—Basically, all you have is a free mass. And then, you expect to see higher stresses in the various organs.

A: Yeah. Very good comments. We are aware of those issues and this is conducted as, like, a challenge to the system program or the model, and we appreciate those issues that you have pointed out. And definitely, those should be accounted for for more further study.

Q: Okay. Thank you.

Q: *Joel Stitzel, Wake Forest University*
I noticed, looking at your simulation, that there was a lot of deformation at the upper abdomen region and not necessarily normal to the abdomen but also kind of out-of-plane deformation. I guess, in the spirit of the workshop, you acknowledge not having the belt load, not having the shoulder load. One way to deal with that deformation: It didn't appear, from deformation, that you had symmetry plane there kind of pushing back on that region and to put a symmetry boundary condition where that model is cut off might actually help your response quite a bit. It's kind of a step in the right direction, but maybe not the final solution.

A: Yeah. That's a good suggestion. Yeah, we just have a free-end boundary condition right now. Yes. Yes. Good. Thanks.