THE LOAD PATH FROM UPPER LEGS TO CHEST IN THE HYBRID III DUMMY; EXPERIMENTS AND SIMULATIONS

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ABSTRACT

It has been shown that the design of the Hybrid III dummy's hip joint can cause abnormally high spikes in the chest accelerations. These spikes are generated when the pelvis rotation is suddenly stopped by the bottoming out of the hip joint. This creates large lumbar shear and tension forces which act to resist forward movement of the dummy's chest. This problem has partly been resolved by the introduction of "modified femurs". However, even with modified femurs, high peaks have still been observed in chest accelerations of some front barrier crashes.

In order to analyze the load path from upper legs to chest, dynamic experiments have been performed on the hip joint (with modified femurs and with standard femurs), on the isolated lumbar spine and on a partial dummy consisting of upper legs, torso, neck and head. These tests have been used to significantly improve an existing model of the dummy.

In the hip joint, a considerable rate-dependency was found and the adjustment of hip friction was found to be an important factor. For different lumbar spines from the same manufacturer, major differences in response were found. These differences between dummy parts are a concern for reproducibility of full dummy tests.

A sensitivity analysis showed that such dummy related factors lead to variations in the order of 2-8% for peak chest acceleration and chest deflection, but lead to much larger variations in lumbar loads.

INTRODUCTION

Uncharacteristic high thoracic spine accelerations have been noted by several vehicle manufacturers when driver and passenger air bag restraints were used in combination with the unbelted 50^{th} percentile Hybrid III dummy. These peaks were shown to be caused by bottoming out of the hip joint. This problem is mostly referred to as "hip lock" or hip joint interference. As this response of the dummy was not considered biofidelic, modified femurs have been designed. Several evaluations showed that this modification reduced the signs of hip lock (Abramoski et al., 1994; Abramoski et al., 1995; SAE 950660; Nusholtz et al., 1995; Klinich et al., 1995; Kanno et al., 1996). Well validated mathematical models of the Hybrid III dummy failed to reproduce these peaks in the thoracic spine acceleration. Furthermore the lumbar spine loads were not accurately predicted. Apparently the load path from upper legs to chest was not well captured. In order to clarify the mechanics of this load path the following series of experiments was performed.

1) Dynamic tests on the isolated lumbar spine
2) Dynamic hip flexion tests
3) Dynamic tests on a partial dummy including hip joints, upper torso, neck and head

These tests have been used to improve an existing model of the dummy. With this improved Hybrid III model, full scale tests have been analyzed in order to further clarify the role of different dummy components in the load path from legs to spine. Major variations in resistance were found between different lumbar spines, and it was found that hip friction is generally not set as prescribed. To evaluate these sources of variability on injury numbers a sensitivity analysis was performed on realistic applications of the Hybrid III dummy.

LUMBAR SPINE TESTING & MODELLING

In a previous test programme the relations between rotations (bending & torsion) and displacements (compression & elongation and shear), and moments and forces were determined in a quasi-static manner (MADYMO, 1994). Damping coefficients were then estimated on the basis of pendulum test results. Below more extensive dynamic tests on the lumbar spine are described.

Lumbar spine bending and shear

Dynamic lumbar spine tests were performed where the lumbar spine was rigidly connected to a sled. A sled acceleration with the appropriate pulse shape was induced with a Monterey setup. On the lumbar spine a Hybrid III spine box was attached including a spine load cell. Ribs and neck bracket were removed. The spine box yields an inertial loading to the lumbar spine. By attaching one or more load masses to the spine box this inertial loading was manipulated. Three loading assemblies were used:

- shear assembly; an assembly with a CG below the spine box was designed such that mainly shear was induced in the lumbar spine.
- bending assembly; a mass was attached on top of the spine box which resulted in strong bending and limited shear in the lumbar spine.
- torsion assembly; a mass was attached lateral (to the right) of the spine box such that a combination of torsion, bending and shear was induced in the lumbar spine.

For all conditions, two or three loading severities were applied with one lumbar spine which further will be referred to as *specimen 1*. The shear and bending tests with the highest loading severities have been repeated with another lumbar spine (*specimen 2*).

Repeatability and reproducibility - For repeated tests with specimen 1 only minor differences were observed. Film results showed a good repeatability for displacements and rotations. The response of specimen 2 differed notably from specimen 1. Furthermore the repeated tests with specimen 2 showed rather large differences. Both for the shear and the bending condition, specimen 2 gave reduced spine deformations as compared to specimen 1. These results indicate that the reproducibility of different spines is poor. Repeatability is good for specimen 1 with which all test conditions were studied and repeatability is not good for specimen 2. Given the differences found it seemed worthwhile to test more specimens. Additional tests on three lumbar spines were conducted using the most severe loading conditions of the bending and shear test configurations. Test specimens were chosen to have spines at the low, mid and high range of the part specification tolerance of 75 to 85 durometer. The spines tested had measured durometers of 77, 81, and 84. As part of this study the spine cable tension was also varied +/- 15% from the nominal specification of 12 in-lbs. Since there is no performance specification for the Hybrid III dummy lumbar spine, these two variables essentially are the only controlled parameters that could affect its performance (the spine geometry is also defined but because it is made out of polyacrylate it does not completely return to its initial geometry after it has been deformed). Each test configuration was done two times resulting in a total number of 36 tests. The results show that the spine stiffness can vary significantly, even among certified lumbar spines. A strong relationship was found between the lumbar spine durometer and the bending stiffness as is evident by looking at Figure 1. The higher the durometer of the lumbar spine, the higher the measured loads and the lower the spine box rotation. There was no relationship evident between the spine cable tension and the spine stiffness. It should also be mentioned that the measurements of the spine durometer were done by a person familiar with taking this measurement from First Technology Safety Systems. The same lumbar spines were remeasured by the same person 3 months later and were found to be within +/- 1 durometer from the original measurement. But when other people measured the durometer, they got significantly different values (by as much as 6). This indicates that specifying a durometer specification may not be appropriate since it appears to be very user dependent. This is primarily due to the lumbar spine not having a flat surface from which the durometer could be measured.

Lumbar spine axial compression

The response of the lumbar spine in axial loading is particularly relevant for aircraft seat testing. Therefore dynamic tests were performed for axial compression. These were performed with a Zwick-Rel dynamic testing machine at various loading rates.

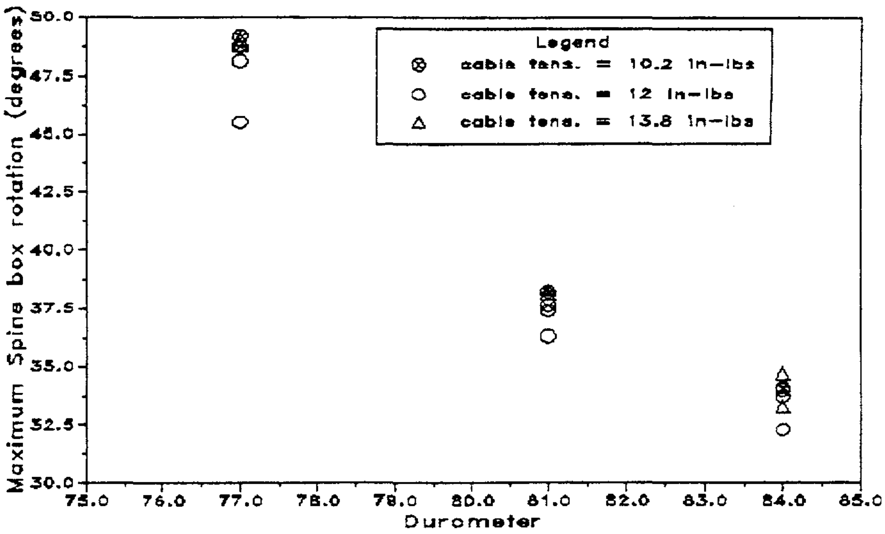

Figure 1: Spine box rotation vs. spine durometer from the lumbar spine bending tests.

Lumbar spine modelling

The dynamic lumbar spine tests described above have been used to improve the existing lumbar spine model (MADYMO, 1994). The model was optimized, such that the differences between the simulations and the experimental results were minimized. Some parts of the optimization were performed manually. The most complicated parts of the optimization were performed automatically (MADYMO, 1997). The automated optimization has the following advantages:

- several parameters can be optimized together; the program deals with the interaction between the parameters and finds an optimal set of parameters,
- different signals from one test, and even from different tests can be combined into one objective function

In several steps the model was optimized. The existing model was based on extensive quasi-static tests and a limited set of dynamic experiments. Therefore it was expected that mainly the damping parameters of the model should be adapted. Some information on the rate dependent behaviour of the Hybrid III lumbar spine can be found in Begeman et al. (1994). Here it was concluded that the shear resistance of the lumbar spine is strongly rate dependent. These effects were already apparent at the applied maximum loading rates of 50 mm/s whereas shear rates up to 2.8 m/s were found in the component tests now used for modelling. In the existing model the spine cables were modelled as a very high joint stiffness for elongation. An improved prediction was obtained by modelling the cables separately as a Kelvin element. As stated above the results for specimen 2 differed strongly from specimen 1. The model was optimized for specimen 1. Finally the model was optimized also for specimen 2. This optimization had the following results. The bending resistance of specimen 2 was found to be 1.4 times that of specimen 1. The shear resistance was estimated to be practically the same for both specimens. This confirms that specimen 2 has a much higher resistance in bending as compared to specimen 1.

HIP JOINT TESTING AND MODELLING

Dynamic component tests have been performed on the hip joint. The goal of these experiments was to determine the dynamic resistance of the hip joint. The modified femurs are designed to prevent the occurrence of interference in the hip joints. Hip lock occurs when the dummy's upper femur bottoms out and makes metal to metal contact with the pelvis bone. A distinction is made between "hard" hip lock and "cushioned hip lock" (Klinich et al., 1995). Hard hip lock can occur in the standard femur and cushioned hip lock can occur in the modified femurs. Two pelvis/femur assemblies have been tested. The first assembly is a standard 50^{th} percentile Hybrid III pelvis with standard femurs. The second assembly is a new pelvis with modified femurs. These two assemblies are respectively abbreviated as "standard femurs" and "modified femurs".

The pelvis was held "rigidly" at the lumbar spine attachment with the upper legs directed upwards and the lower legs removed. Both legs were tested at the same time in order to study differences in behaviour between the left and the right hip. It is known that the left and right hips behave differently because the pelvis is not left/right symmetric (Klinich et al., 1995; Abramoski et al., 1994).

The modified femurs were tested at two loading severities and at different loading rates. The highest severity tested induced "cushioned hip lock". The low severity tests were also repeated with standard femurs and did not result in hip lock. Accelerations in x, y and z directions were recorded at the sled, the pelvis and the knees. The hip flexion angles were calculated from string pot measurement. The flexion angles calculated from the string pots were verified using high speed video. It was concluded that the string pots were able to follow the dynamics of the knees accurately.

Hip friction

Calibration procedures describe that "Limb joints are set at 1 G, barely restraining the weight of the limb when it is extended horizontally" (CFR part 572 subpart E). Calculations assuming a horizontal upper and lower leg were performed. Thus the static hip joint friction was estimated to be about 56.1 Nm. However, in our experience a friction much below 56.1 Nm is often applied in real dummies. Simulations of various full dummy tests indicated a hip friction in the order of 12.8 Nm which was further adopted for the model. This will be treated further in the discussion.

Friction has been implemented for the hip joints with the COULOMB FRICTION model. Both a constant friction and an additional load dependent friction were specified.

Hip stiffness and damping

Hip flexion requirements were specified by the SAE Large Male and Small Female Dummy Task Group (SAE 950660). Calibration specification tests were carried out for the modified femurs which were used in the dynamic hip flexion tests. It was found that the left femur did meet the specifications and that the right femur was right on the limit of 46 degrees at 340 Nm. The torque-angle result for the left leg calibration was applied in the model. Additional damping was implemented as being dependent on the rotation angle. The dynamic component tests were used to optimize the model. Optimization methods were used to systematically determine parameters providing a best fit for several output variables of different experiments. The hip joint model was optimized using test data of the left joint. It was found that the model based on the left hip is sufficiently accurate for the right hip. The hip model was updated using test data of the modified femurs. Moderate loading of standard femurs was also simulated with the new hip model. From these simulations it was concluded that the new model provides a reasonable prediction for moderate loading of standard femurs.

DYNAMIC TESTS ON HIP JOINTS, TORSO, NECK AND HEAD

Tests with a partial dummy have been executed to study the behaviour of the combined hip/lumbar spine section of the dummy. In such tests, lumbar spine deformation is not only resisted by the lumbar spine itself but also by contact interactions of the rib cage, the abdomen and the lower torso (Heinz, 1993). In all tests the arms were removed. Head and neck were included in these experiments. In all forward loading tests, the lower legs were removed and the knees were mounted on the sled. The pelvis was supported by a rigid horizontal plane. The knee-slider, and knee rotation mechanism were included in the tests. So, the dummy could move forward slightly, and rotate freely around the knee axis. In the rearward loading tests, the upper legs were replaced by rigid supports, and the pelvis was also supported at the back.

Four forward loading tests, were performed with a belt restraining the upper torso. The belt was attached to the base of the neck. The belt was chosen such that it approximates the restraining effect of an airbag. Before performing the experiments, several simulations were run to select the appropriate belt characteristics. This belt limits the rotation of the torso, but the films show that even with this belt, the ribs came close to the upper legs.

Two experiments were also performed without the abdomen. The abdomen reduces the recorded lumbar spine bending torques (MY) by around 10%. This is logical since there is load sharing between lumbar spine and contacts; part of the total bending torque is generated by the abdomen. In other signals smaller effects were found. Even in the lumbar My the differences observed are not very large. However, also when the abdomen is removed, contact interactions still occur between ribs, jacket, pelvis and legs. The experiments do not show directly how large this influence is. Only tests with the abdomen present have been used below for model validation and improvement.

Optimization of the model

The experiments were used to optimize a model representing contacts between ribs, jacket, abdomen and pelvis. These contacts will further be mentioned as "abdomen contacts". A notable effect of these contacts was found for tests with large forward bending of the torso. With these contacts present in the model an improved prediction was obtained of the lumbar spine loads. Upper and lower spine bending torques were reduced up to 35%. These contacts also affected chest kinematics and accelerations.

FULL DUMMY TESTS

Several tests with the complete dummy were analyzed to validate the complete dummy model, and to analyze the load path from legs to femurs. A relatively simple sled test with a rigid seat and with separate shoulder and lap belt has been used for validation. Test and simulation are described as an example in the MADYMO database manual (MADYMO, 1997). Results obtained with the new model results were almost identical to those presented in the manuals for the existing model. The actual tests were performed with standard femurs. However, the updated model based on modified femurs was applied. For this condition, only minor hip rotations were observed. This illustrates that for these conditions, the updated model is also suitable to simulate tests with standard femurs.

Barrier tests with a driver airbag, unbelted were simulated. Kinematics are shown in Fig. 2 and phasing of important experimental signals is given in Fig. 3. Around 40 ms knee bolster contact induces axial femur loads and pelvis acceleration (Fig 3, upper). Around 70 ms maximal lumbar forces and chest accelerations are observed (Fig 3, middle). Validation results are shown in Fig. 4.

SENSITIVITY ANALYSIS

Above concerns were raised about factors in the Hybrid III dummy which will negatively affect the reproducibility and repeatibility of tests. A sensitivity analysis was performed to quantify these effects (Table 1). This analysis was performed for the following configurations

1. *driver airbag unbelted*: this test with validation results is described above.
2. *depowered driver airbag*: The driver airbag unbelted model was modified to simulate the recently adopted AAMA proposal for FMVSS 208. This includes changing the barrier crash pulse to a half sine (17.2g-125ms) pulse and using a less forceful inflatable restraint.
3. *aircraft drop test 30 deg nose down 16 m/s.* This test was performed in accordance with MIL-S-58095 and the body was effectively restrained by a five-point harnass belt.

The following variations of the Hybrid III model were analysed:

1. *softer hip lock* was simulated by applying a hip flexion stiffness based on modified femurs but with a reduced stiffness beyond 340 Nm. Instead of the highly non-linear bottoming out function now a linear stiffness was taken. It should be noted that this model also matches the calibration specs for 340 Nm.
2. *hip friction of 56.1 Nm*: the friction of 12.8 Nm from the standard database was increased to 56.1 Nm.
3. *double lumbar resistance*: the resistance of the lumbar spine model was doubled to simulate a variation comparable to the maximal expected component variations.
4. *no spine cable*: the KELVIN element representing the spine cable was removed from the model. This variation was performed mainly to assess the effect of the spine cable on lumbar tensile forces.
5. *rib-pelvis contacts removed*: this variation was performed to assess the contribution of the contact interactions between ribs, abdomen and pelvis.

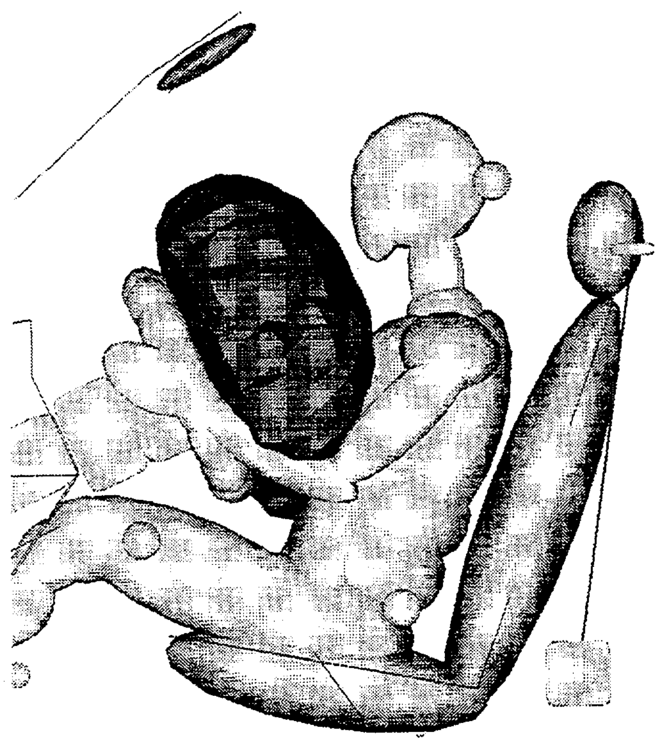

Figure 2. Kinematics of model driver airbag unbelted at 40 ms.

Table 1.
Sensitivity analysis; each block represents a loading condition. For each condition first results are given for the experiment and for the improved dummy model. Then the relative effect of several model variations is given with respect to the standard model. Effects below 1% are omitted (-)

condition model variation	peak chest acc (3 ms) [m/s2]	peak chest deflection [m]	peak lower lumbar Fz [N]	peak lower lumbar My [Nm]	peak femur axial force Fz [N]
driver airbag unbelted					
experiment	720	0.048	-6600 tension	-190	8400
standard model	707.7	.05342	-5630	-382	9529
softer hip lock	-5.6%	-	-4.8%	-12%	-
hip friction 56.1 Nm	-5.7%	-	-8.5%	-7.2%	-9.7%
lumbar resistance*2	-5.9%	-3.7%	-11.7%	+30%	-
no spine cable	-5.8%	+1.2%	-8.5%	-10%	-
no abdomen contact	-	-	-	+0.5%	-
depowered driver airbag unbelted					
experiment	-	-	-	-	-
standard model	357.5	0.03232	+2588 compressive	-210.7	8351
softer hip lock	-	-	-	-	-
hip friction 56.1 Nm	-6.7%	-7.8%	-	+6.6%	-2.4%
lumbar resistance*2	-3.1%	+2.1%	+2.1%	+20%	-1.4%
no spine cable	-	-	+3.6%	+1.3%	-
no abdomen contact	-	-	-	+1.8%	-
aircraft drop test 30 deg nose down					
experiment					
standard model	336.6	0.0178	+6637 compressive	-452	2553
softer hip lock	-	-	-	-	-
hip friction 56.1 Nm	-5.9%	-6.2%	+2.1%	-13%	+8.5%
lumbar resistance*2	+37%	+2.6%	+110%	+59%	-2.0%
no spine cable	-	-	-	-	-
no abdomen contact	-3.9%	-1.5%	-28%	-16%	-

DISCUSSION

Dynamic tests have been performed on lumbar spines, on hip joints with standard and modified femurs and on partial dummies. These tests have been used to improve an existing model.

In the hip joint, a considerable rate-dependency was found. For component tests below hip lock level, limited differences were observed between the old and the new femurs. This is in agreement with full dummy evaluations described in the literature (Klinich et al., 1995; Nusholtz et al., 1995). The current specification for hip resistance surely reduces variability induced by the condition of the dummy (SAE950660). Some variation may still be found above the calibration level. Table 1 indicates that such a variation could affect chest G's in the order of 6% for conditions with cushioned hip lock. The standard calibration requirement for hip friction results in a value of about 56.1 Nm. Only 12.8 Nm was implemented in the model since in our experience a friction much below 56.1 Nm is often applied in real dummies. The sensitivity analysis indicated a considerable influence of hip friction on the dummy response. Given the influence of hip friction it is recommended that this variable is well controlled in experiments. This would improve reproducibility of tests and would facilitate modelling.

Major differences in response were found for different lumbar spines. These were shown to relate to durometer testing (see Fig. 1). However lumbar durometer measurements are found to be very user dependent. Alternatively a dynamic bending calibration could be specified for the lumbar spine. This would help in reducing test variability. In component tests and in the sensitivity analysis only minor effects of the spine cable were found. Tests and simulations on the partial dummy showed significant effects of contacts between ribs, abdomen and pelvis. These contacts add to the bending resistance of the lumbar spine, and thereby affect the lumbar spine loads. The sensitivity analysis showed that this effect was particularly relevant in the aircraft test.

CONCLUSIONS

Testing on hips, lumbar spines and partial dummies provided insight in the load path from legs to upper torso and was used to improve an existing model of the dummy. Major variations in bending resistance were observed for the lumbar spine, and concerns were raised about the adjustment of hip friction. The sensitivity analysis showed that such dummy related factors lead to variations in the order of 2-8% for peak chest acceleration and chest deflection, but lead to much larger variations in lumbar loads.

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Figure 3. Experimental signals of test driver airbag unbelted

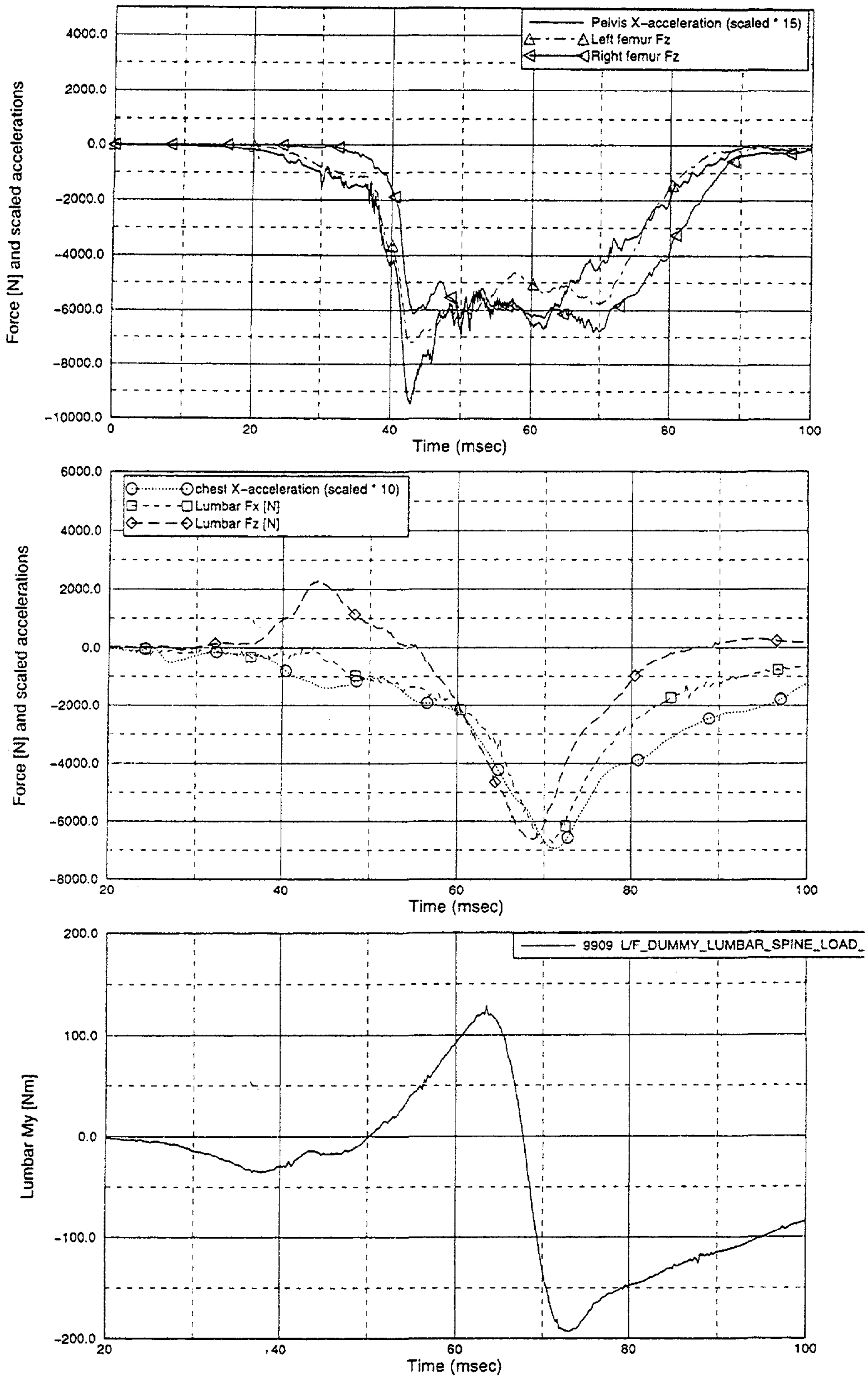

Figure 4. Validation results of test driver airbag unbelted