Validation Of A Seat-Dummy Simulation Model For Rear-Impact

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ABSTRACT

Seat test standard protocols have been established by insurance research institutes and consumer test organisations are developing similar test procedures to assess the performance of seats under rear-impact crash conditions.

With several numerical simulation models of the BioRID II being commercially available this study is intended to validate a multi-body rear-impact dummy model in a neutral seat environment for a range of seating postures and impact severities. This enables the systematic investigation of those parameters of the seat which influence the biomechanical loading on the dummy.

For this purpose, five dynamic tests were conducted on a newly developed test device that employs a stationary carriage with the seat and dummy and is accelerated from the rear by a sled-on-carriage impactor system. The BioRID dummy was placed on the so-called Chalmers seat which was utilized in earlier EU co-funded research projects and provides several adjustment possibilities to represent different seat shapes and characteristics. Starting with a medium severity crash pulse four additional validation tests were carried out with lower and higher crash severity as well as different seat and seating positions to cover a broader range of conditions.

Modelling involved both the detailed measuring and computational representation of the Chalmers seat as a multi-body model with facet surfaces as well as careful documentation of the placement of the BioRID model on the seat.

Based on the comparison of the model response with the kinematics and biomechanical measurements from the basic test an acceptable conformity between numerical model and validation test could be found for most body regions. However, some shortcomings in the dummy model were identified.

INTRODUCTION

Soft tissue neck injuries, often termed whiplash-associated disorders, are among the most frequently reported injuries of car occupants in many countries with a high level of motorization [1]. Although assessed as minor injures and usually healing without further bodily impairment they represent a large burden to society due to the large number of incidents, mostly among occupants of passenger cars struck in the rear by another vehicle. A comparative study conducted by the European insurance industry in 2004 reported a large difference in the situation of minor cervical trauma claims among European countries, both in number of claims and in cost of injury [2]. For instance, in Germany approximately 47% of all bodily injury claims are linked only to minor neck injuries, for France this rate is only 3% and for the United Kingdom it is approximately 76%. The annual costs of these injuries range from an estimated 1.5 million Euros in Finland to 500 million in Germany to more than 2.9 billion Euros in Italy. Many research studies have investigated into the injury mechanisms and have tried to establish biomechanical threshold values for these types of injuries, but a common understanding does not exist until today. The International Insurance Whiplash Prevention Group (IIWPG), a group of insurance research institutes from North America, Australia and Europe developed a geometric and dynamic test and assessment protocol for car seats [3]. This formed the basis for test programs of new vehicle seats which have been published for three years in a row to inform consumers and to increase the awareness in the automotive industry about the safety performance of seats under rear-impact conditions. The assessment criteria employed in this protocol have their foundation in surveys of the number of neck injury claims in connection with particular passenger car models [4]. For two vehicle models, the number of claims for the model generation equipped with standard seats was compared with the claim figures for the subsequent generation which featured seat designs specifically engineered for better rear-impact neck protection. Employing the BioRID II anthropometric test device, laboratory tests with these seats under rear-impact conditions displayed significant differences in some biomechnical variables depending on the design strategy against neck injury. While all of them aim at reducing the upper neck forces (shear and axial) the methods differ. Specifically energy-absorbing seat backs reduce the longitudinal acceleration in the T1 vertebra of the BioRID II dummy whereas re-active head restraints seek to minimize the time to close the gap to the back of the head. These physical values serve as seat evalution criteria in the IIWPG test procedure. While a considerable number of passenger car front seats has been evaluated according to this protocol it still requires physical testing. Some dummy responses have shown to be sensitive to slight variations in seating position and seat variability as it may occur in serial production. Moreover, the interaction between dummy and seat is often difficult to observe since the view can be obstructed by the dummy's arms, clothing or the seat upholstery.

This was the major motivation to develop a validated numerical simulation model which includes a commercially available BioRID II model in a neutral seat environment. Such a model would enable to study the influence of a variety of seat and impact parameters on the dummy loadings with special focus on the cervical and thoracic spine. Numerical simulation can also serve as a cost-efficient and time-saving alternative to testing when fundamental effects of other seat types, e.g., for light commercial vehicles or rear seat benches, need to be investigated. Hence, great store was set by the prognostic capabilites of the numerical model for a range of possible dummy seating positions and impact severities.

VALIDATION TESTING

Development of Dynamic Test Device

Experience from other laboratory experiments shows that the initial dummy position has large influence on the result of a dynamic rear-impact test. Therefore, the objective was to design a test set-up with a stationary carriage which is accelerated when the crash pulse sets in. A Hyge sled was not available for testing. Instead, a system was developed where the seat is mounted on a stationary carriage which is accelerated from the rear by an impactor device with a defined crash pulse. Sets of coil springs are used to transfer the impact forces which enhances the repeatibility of the force characteristics between tests and reduces test costs.

The concept was worked out using a MADYMO simulation model of the test set-up. The first evaluations of a system which employed a single impacting mass revealed that this would permit only the realisation of harmonic crash pulses. In order to simulate crash characteristics of non-harmonic shape like the IIWPG crash pulse an additional mass is necessary to interact with the seat carriage at a later point in time during the impact. A similar effect was achieved with the development of a sled-on-carriage system, incorporating the impactor carriage and an additional sliding mass. By adjusting the amount of additional mass, the time that it interacts with the seat carriage and the stiffness of the coil springs the crash pulse can be tuned to the desired shape. Different options of setting-up the additional masses on the impactor carriage were investigated. A simple lumped mass model with coil springs was used to determine a suitable combination of mass and spring stiffnesses. However, no masses were assigned to the springs in the model. The schematic test set-up is shown in Figure 1.

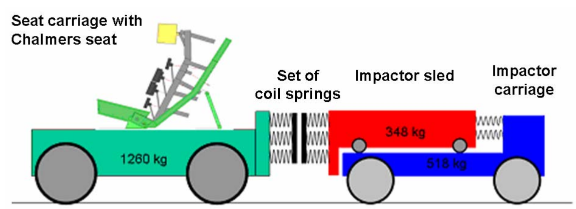


Figure 1. Schematic test set-up.

During the first pre-tests with a physical test device it became clear that the natural oscillation of the spring coils has significant influence on the resulting crash pulse. The MADYMO model was therefore refined in the area of the spring simulation. Each coil spring was modelled as a system consisting of ten bodies with linear stiffnesses in between. The large number of springs resulted in a complex calculation model, but with good prognostic capability. By distributing the springs among the impactor sled and the seat carriage their effect on the crash pulse could be greatly reduced.

Figure 2. Sets of coil springs between impactor sled and seat carriage.

Figure 3 shows the target crash pulse, the IIWPG curve in this case, in comparison with the calculated pulse and the pulse measured in the experiment. The corridor defined by IIWPG is met except for a slight deviation in the time frame after 87 milliseconds.

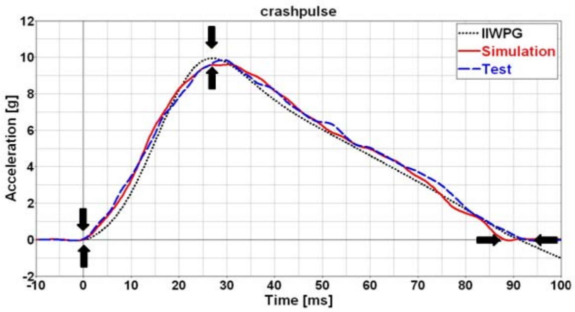


Figure 3. Comparison of IIWPG target pulse, calculated and test crash pulse.

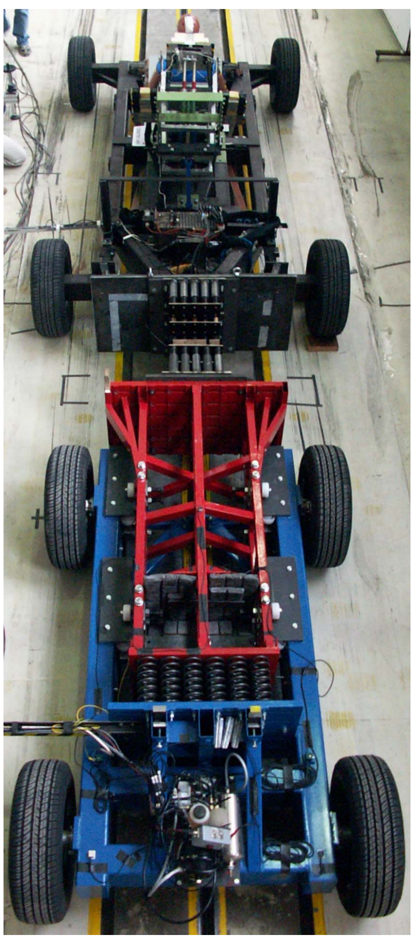


Figure 4. Complete test set-up with impactor (foreground) and seat carriage (background).

Test Seat

For the primarily intended purpose of a parameter study the use of production car seats entails some disadvantages. The observation of the interaction between dummy and the seat is limited due to upholstery, especially on the seat cushion and seat back sides. Purposeful variation of seat parameters requires extensive modifications of the seat construction and may alter its performance unintentionally. For this reason, the so-called Chalmers seat was used in the dynamic validation tests. It was developed in the course of the „Whiplash“ project co-funded by the EC [5] and was employed in several research projects focussing on rear-impact neck injury and protection. This seat displays a generic design with a number of separate and adjustable seat elements which allows a detailed investigation of the interaction between dummy and seat and the variation of isolated seat parameters. The seat features a rigid seating surface and an adjustable seat back frame with an articulated sub-frame which carries four movable seat back elements and a movable head restraint (see Figure 5 and 6). All of the seat back elements and the head restraint can be individually adjusted. The sub-frame is connected to the seat back frame by means of a deformation element that works according to the principle of a sheet metal bending brake.

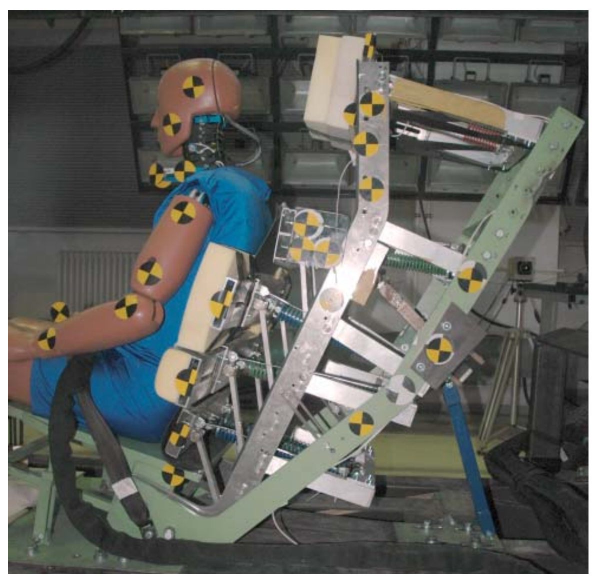


Figure 5. Chalmers seat.

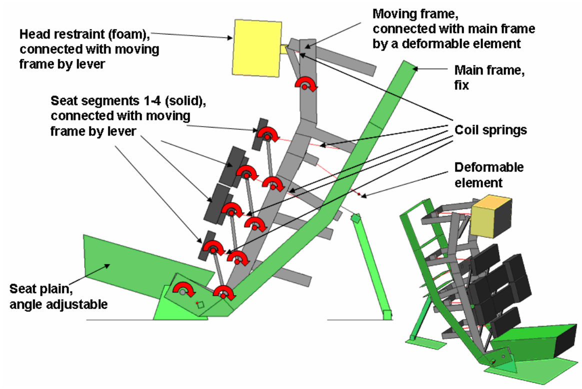


Figure 6. Schematic representation of Chalmers Seat.

Validation Tests

A large number of characteristic points on the dummy and on the seat were measured before the tests to obtain detailed data for the following set-up of the numerical simulation model and to ensure a repeatable dummy position for all tests (see Figure 7). The tests were documented with three high-speed video cameras. On the seat carriage, the longitudinal acceleration was measured and signals on all standard measurement locations of the BioRID II dummy were recorded and processed according to SAE standards (see Figure 8).
The test dummy was freshly calibrated before the test series and conditions of the test environment met IIWPG requirements. This provided an extensive amount of information and measurement data from the validation tests for the development of a numerical simulation model.
Altogether, five validation tests were conducted with the described test set-up and using the BioRID II anthropometric test device: a basic test applying the IIWPG crash pulse, one test each with reduced and increased crash severity, one test with increased backset between dummy head and head restraint and one test with increased seat back angle.

Test No.	Description
V01	Increased backset
V02	Basic test conditions
V03	Reduced crash severity
V04	Increased seat back angle
V05	Increased crash severity

Starting from the basic test conditions, the purpose was to vary the crash pulse severity in two tests and maintain the remaining parameters, and to vary the seat geometry in two tests while maintaining all other parameters. The initial positioning of the test dummy was carried out according to IIWPG requirements as far as the Chalmers seat design allowed; the seating position was then replicated for the tests with variation of the crash pulse. In the tests with increased backset and with reclined seat back, the dummy posture had to be changed slightly to achieve the desired seat adjustments.

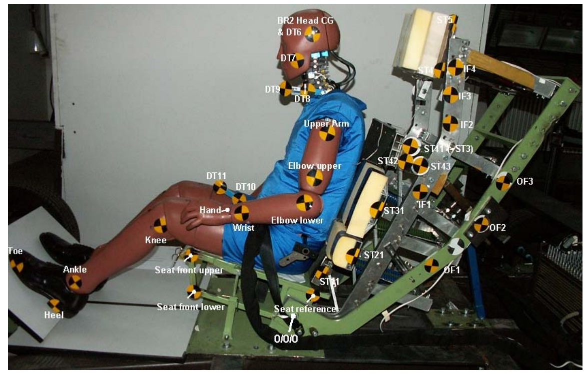


Figure 7. Measurement points on dummy and test device.

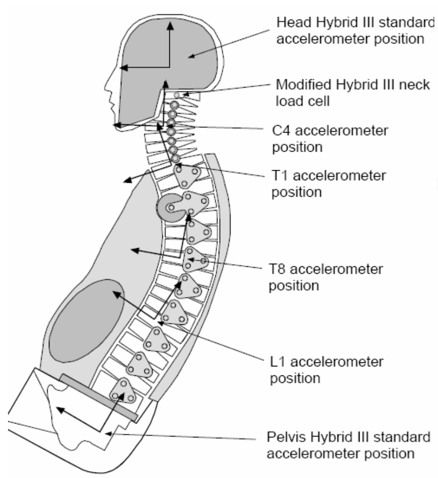


Figure 8. Measurement locations on BioRID II dummy.

The test results were assessed largely in accordance with IIWPG guidelines. The maximum shear and axial forces measured on the upper neck are represented in a rating chart as used by IIWPG to classify the combined neck force values according to one of the three categories of “low”, “moderate” and “high” neck forces (see Figure 10). Similarly, the maximum longitudinal acceleration measured on the T1 thoracic vertebra is assessed when rating the performance of seats under rear-impact conditions.

Effect of Crash Severity

In three of the validation tests, the crash severity was varied whereas all other variables remained unchanged. Starting from the crash pulse in base test

V02, representing the IIWPG condition, a pulse of lower severity with a peak acceleration of 8 g's and a pulse of higher severity with almost 13 g's were applied. Their corresponding delta v's total 13.0 kph and 17.6 kph, respectively.

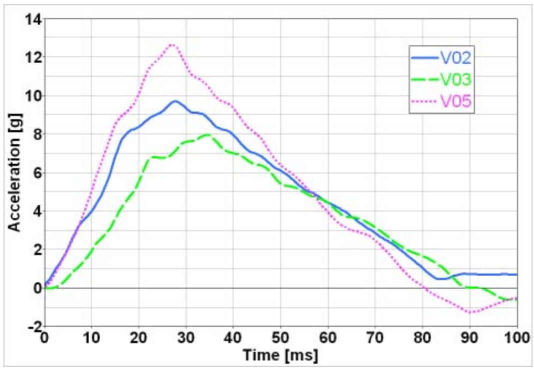
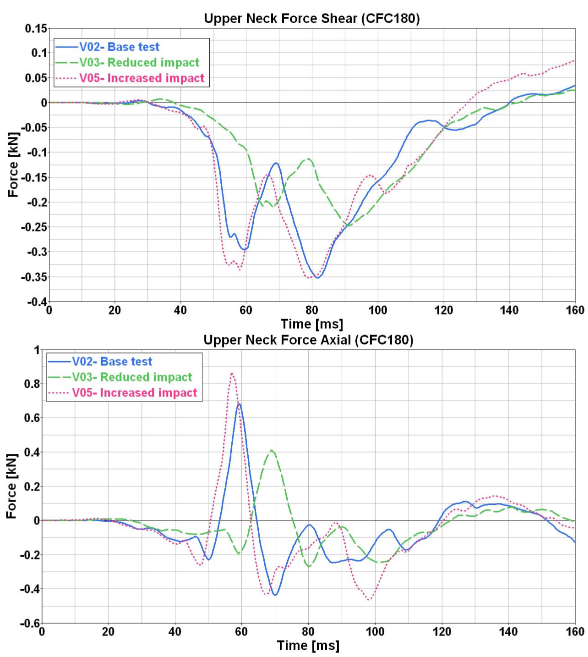


Figure 9. Different crash pulses used for validation tests.

The following figures show the effect of crash severity on dummy loadings according to IIWPG. However, the present study takes into consideration also negative shear values for illustration in the rating chart and assessment of the validation quality.



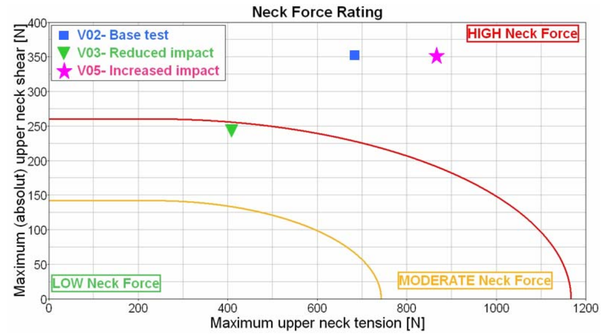


Figure 10. Effect of crash severity on upper neck forces.

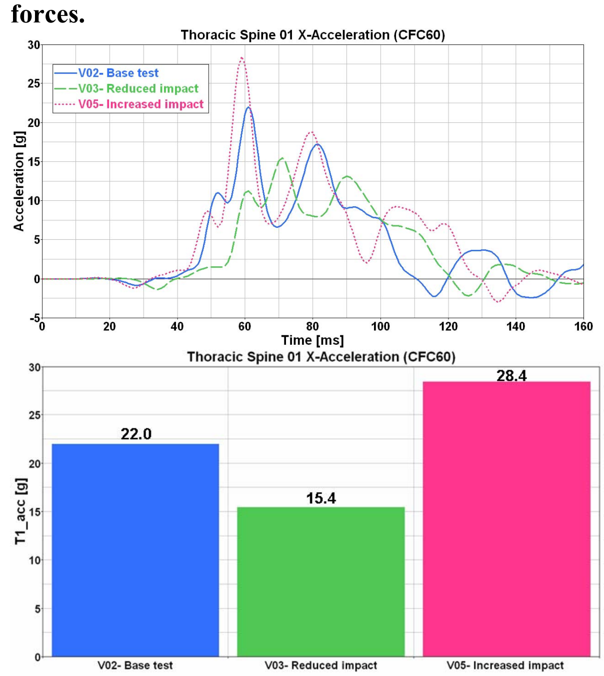


Figure 11. Effect of crash severity on T1 longitudinal acceleration.

Effect of Backset

Backset is the initial horizontal distance between the back of the dummy's head and the front of the head restraint. While the base test V02 featured a small backset, this distance was considerably increased in test V01, partially by changing the head restraint position relative to the seat back and by slightly modifiying the dummy posture. The initial positions are shown in Figure 12.

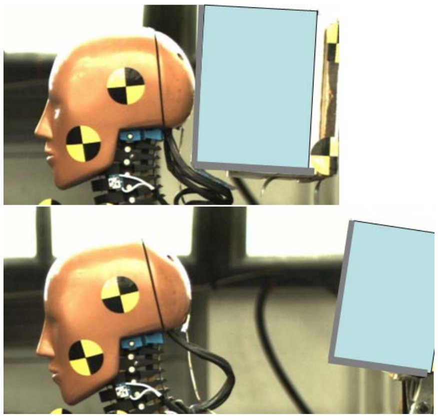


Figure 12. Initial dummy and head restraint position in test V02 (top) and V01 (bottom)

Changing the backset leads to large differences in neck loadings (see Figure 13). With increasing backset the tendency for positive shear in the neck, i.e., the head moving rearward in relation to the thorax, rises. The T1 acceleration was reduced in the test with larger backset, but the difference in dummy position probably contributed to this effect.

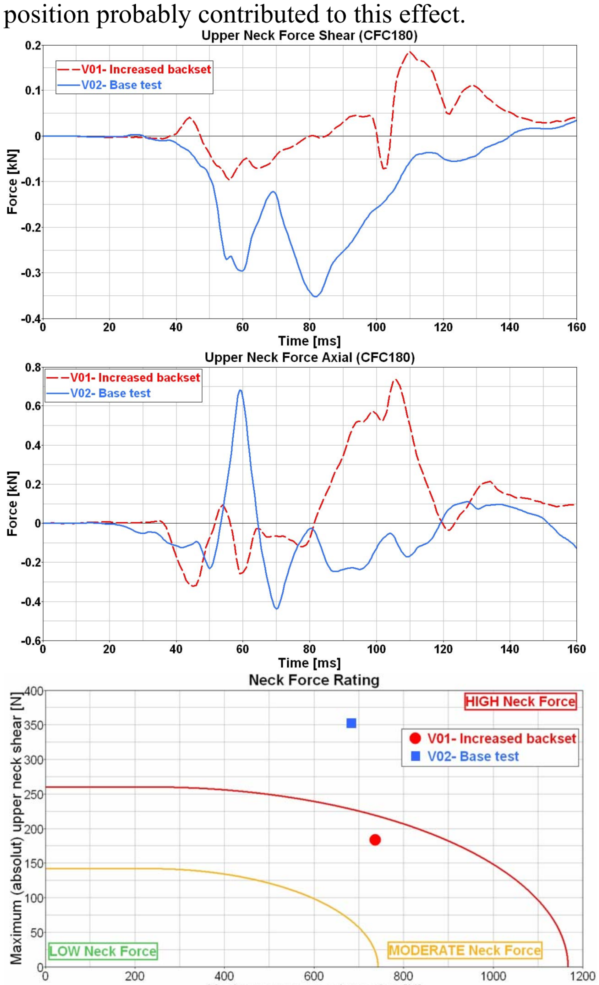


Figure 13. Effect of backset on upper neck forces.

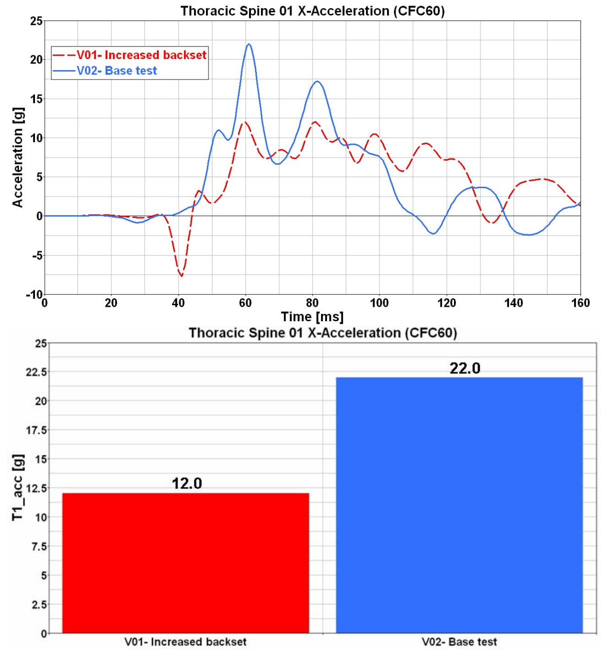


Figure 14. Effect of backset on T1 longitudinal accleration.

Effect of Seat Back Angle

When the seat back angle was increased in test V04 it was necessary to adjust also the dummy's posture. The initial situations for this and the base test V02 are compared in Figure 15.

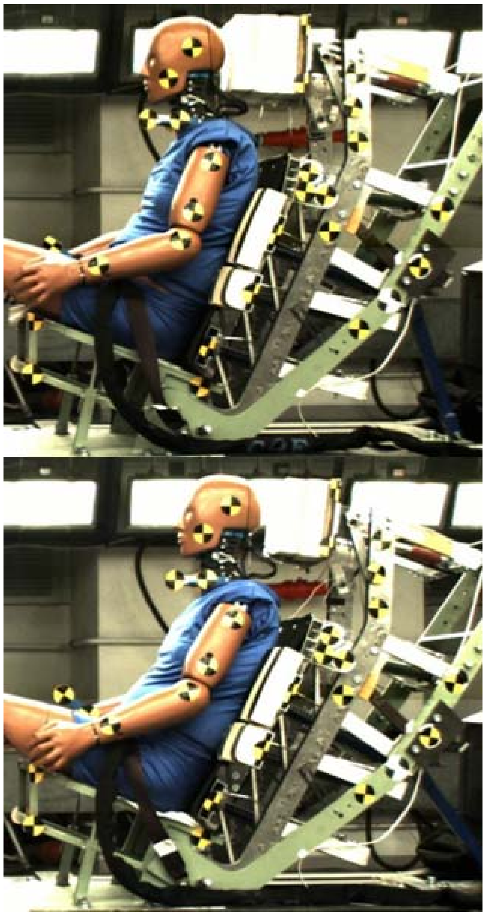


Figure 15. Initial dummy and seat position in test V02 (top) and V04 (bottom).

With stronger inclination of the seat back a reduction of the tension and shear force peak values as well as T1 acceleration can be observed. Nevertheless, this should not serve as a simple measure to reduce the biomechanical loadings in a rear-impact as it increases the risk of the occupant sliding over the seat back.

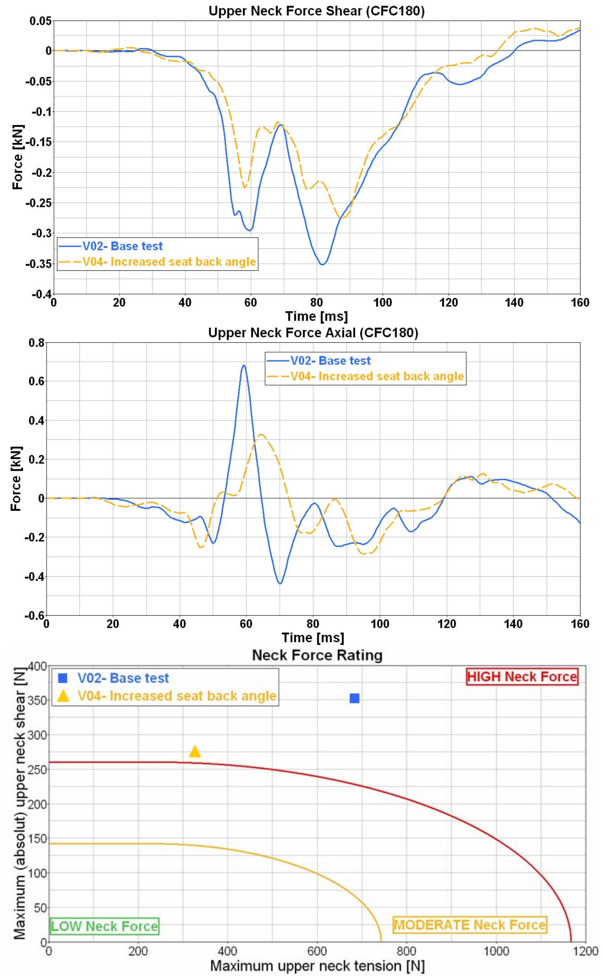


Figure 16. Effect of seat back angle on neck force.

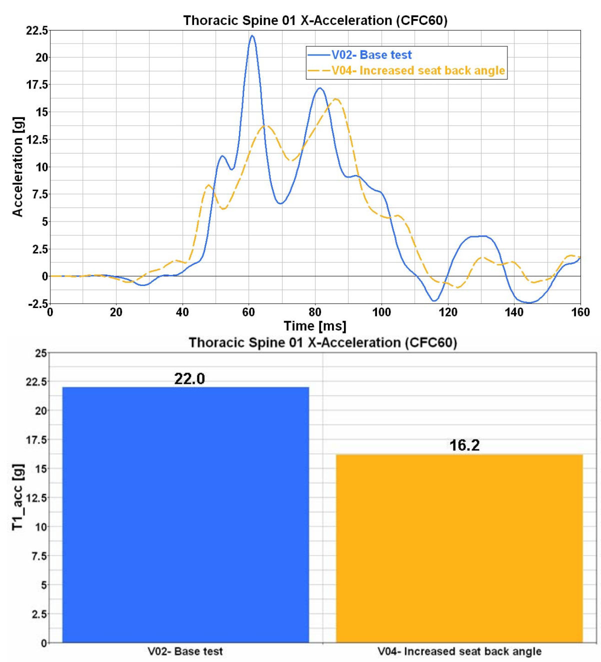


Figure 17. Effect of seat back angle on T1 accel.

Numerical Simulation Model

The simulation model was set up mostly as a multi-body model in MADYMO 6.3.2. The seat was built up completely new. For this purpose, the geometry of the Chalmers seat was digitized with a coordinate measuring device and modelled with the Pro/ENGINEER CAD software system. The mass inertias of inidividual seat parts were calculated on the basis of weight and geometry measurements. Spring characteristics were calculated based on the measurement of their specific geometries. The seat model was developed as a multi-body system with discrete masses and joints. In order to replicate the seat geometry as closely as possible the measured data were transferred into a facet model and superposed on the multi-body model.
A PAMCRASH finite element model was used to determine the characteristics of the sheet metal bending brake located between the seat back frame and the sub-frame since extensive experience regarding model set-up and material properties existed already. The obtained force-deformation characteristics were then transferred to the MADYMO model. The foam of the head restraint on the Chalmers seat was modelled in finite element code with solid elements.

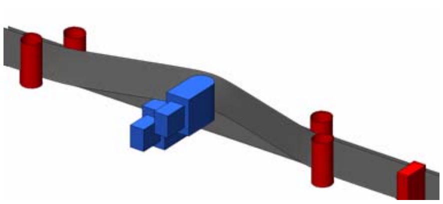


Figure 18. Numerical model of deformation element.

For the numerical representation of the BioRID II the current MADYMO dummy was used which includes facet surfaces (MADYMO 6.3.2, facet dummy version 2.1) (see Figure 19). The numerical model of the dummy was updated twice over the duration of the project. The initially employed model (MADYMO 6.2.2 with BioRID Version 1.1) showed geometric incompatibilities in the neck area due to an unrealistic representation of the geometry of the end stops which led to unnatural joint characteristics. The calculations were repeated when a BioRID II model update was available and provided a stable and satisfying result.

A sensitivity analysis was carried out to investigate the robusteness of the complete model which showed that the calculation requires a time step of one microsecond to avoid large numerical scatter in the results.

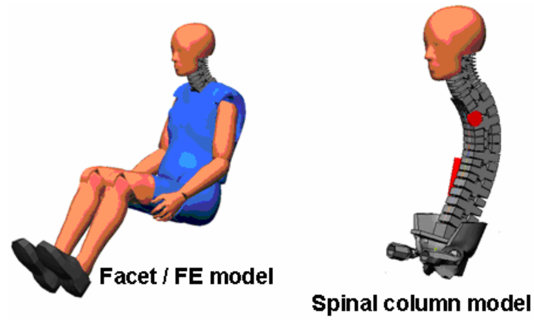


Figure 19. MADYMO BioRID II model.

VALIDATION OF SIMULATION MODEL

In a first step, the crash pulses measured in the validation tests were included in the MADYMO model. Dummy positioning coordinates from the basic test were processed in finite element code and superposed on the MADYMO model to allow accurate positioning of the numerical dummy model. H-point, legs and arms as well as characteristic points on the spine and head were positioned to coincide as much as possible with the measured pre-test dummy coordinates (see Figure 20). However, it proved difficult to adjust the model's spine in such a manner that the balanced position of the dummy model matched the initial seating posture of the real BioRID II dummy. When the numerical model was positioned accurately it produced high accelerations in the vertebrae when the calculation was started, yielding useless calculation results. Hence, the calculation was started with a relaxation phase of 50 milliseconds to eliminate the initial loading effect and allow the model to assume a balanced state. Any small movements of the dummy model during this relaxation were anticipated in the initial position so that the relaxed posture matched the seating position in the validation tests. Only then the complete model was subjected to the crash pulse and the actual simulation phase started. The positions of the joint coordinate systems were not adjusted as they can be expected to reflect the real geometry.

Figure 20. Positioning of dummy model.

The contacts between the dummy and the seating surface and the four seat back elements were defined such that only the characteristics of the dummy model were adapted; the seat surfaces were assumed to be undeformable. Because the foam in the hip area proved too soft an adjusted force-deflection characteristic was applied to produce a realistic behavior of this body region. The dummy could remain unchanged for proper definition of the contacts with the seat back elements. However, the choice of the friction model has considerable influence on the calculation result.
The foam used for the head restraint in the tests was relatively soft and required the use of a solid-foam model to reflect the non-linear effects of the geometry properly. Since no separate component tests could be conducted the foam properties were defined in the context of the complete simulation model.
A specific problem arises – at least in conjuction with the Chalmers seat design – from the fact that the facets of the dummy jacket are attached to the spine segments in the numerical model. In case of relative rotational movement between two adjacent spine segments the size of the facets on the back of the jacket changes. This effect can produce large deformations so that the jacket geometry becomes incorrect. This problem could be solved by adapting the contact characteristics.
The entire model was validated exclusively on the foundation of the base test V02. The validation quality of the numerical model can be judged by the comparison of both the kinematics and the major loading curves between numerical simulation and experiment. The motion of the dummy model matches that of the real dummy very well when their silhouettes are overlaid (see Figure 21).

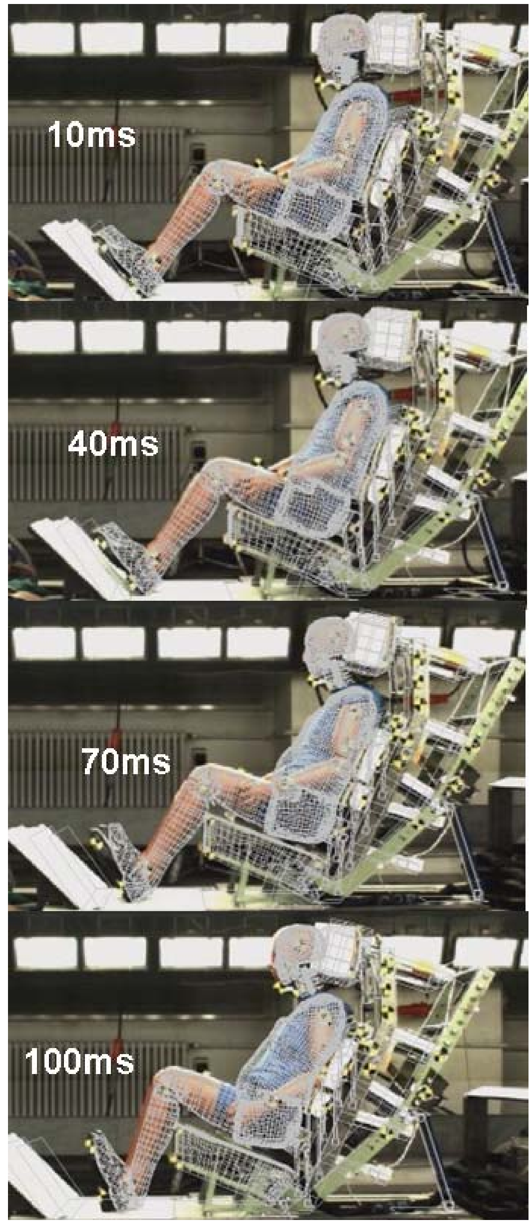


Figure 21. Comparison of kinematics in test and simulation for V02.

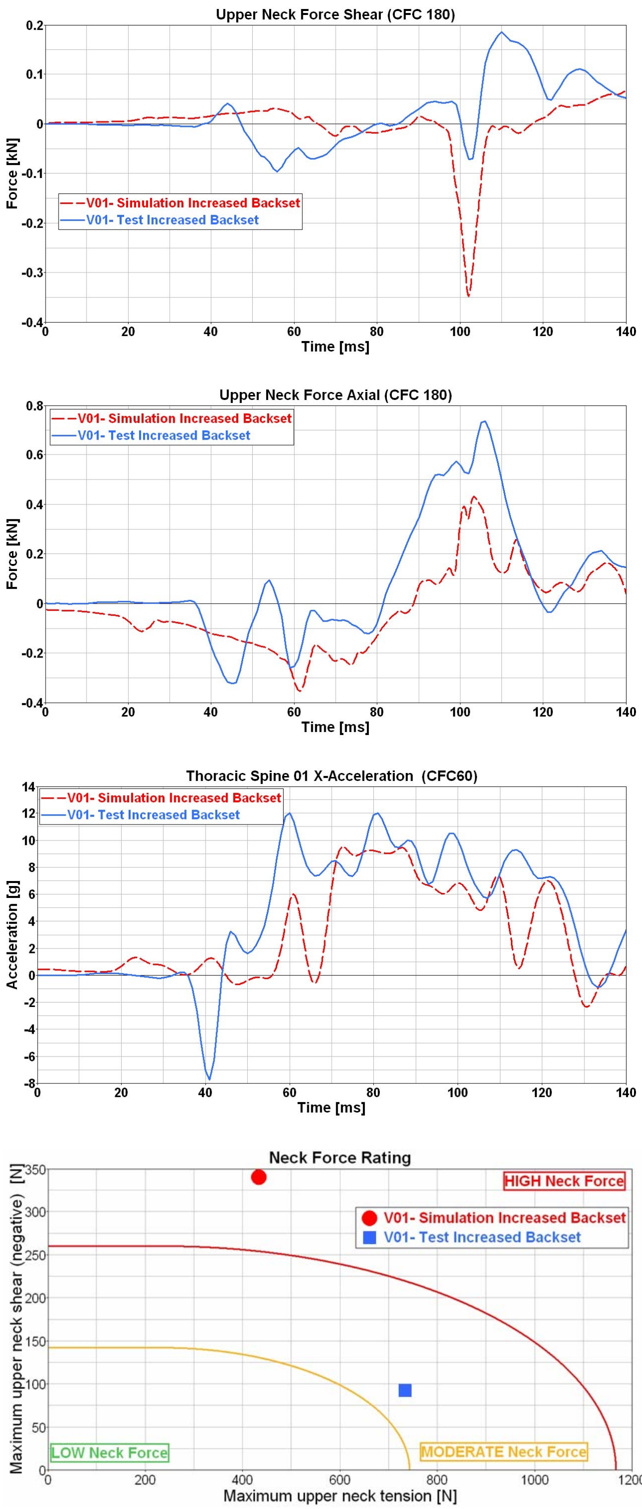


Figure 22. Validation result for test V01.

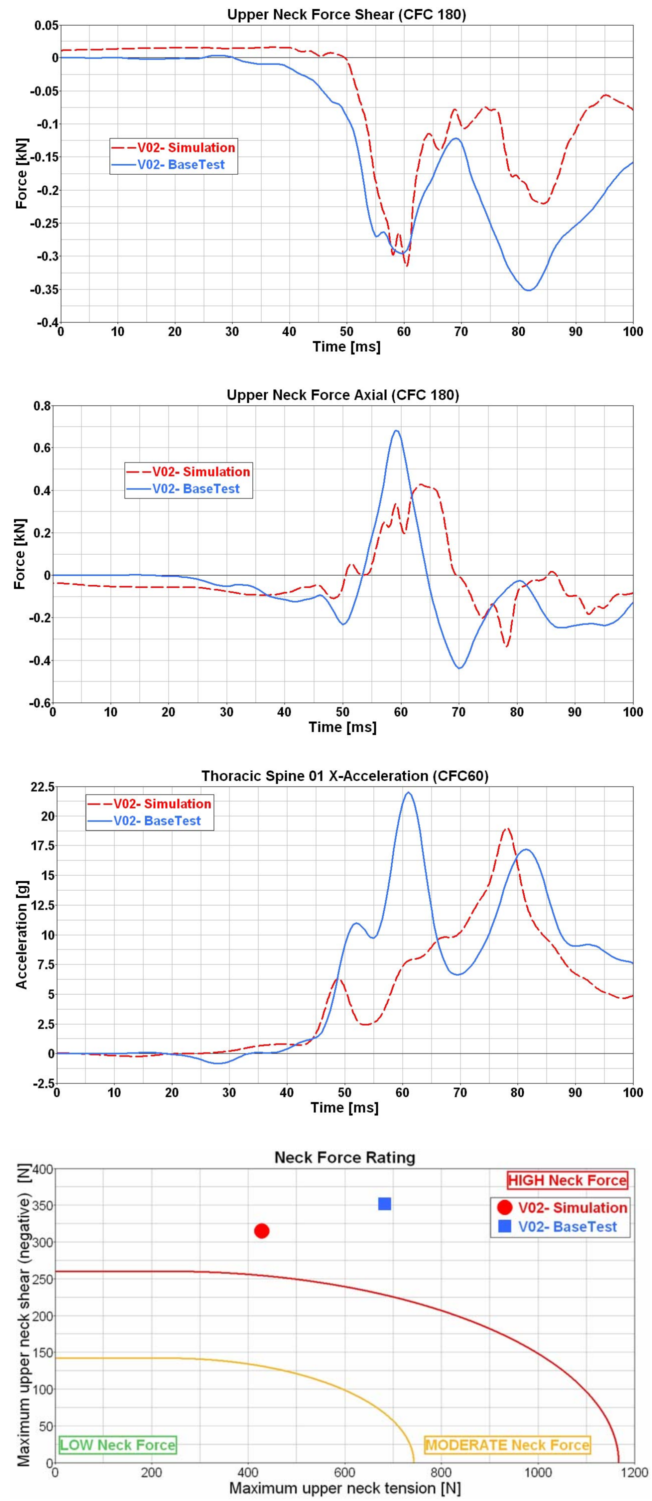


Figure 23. Validation result for test V02.

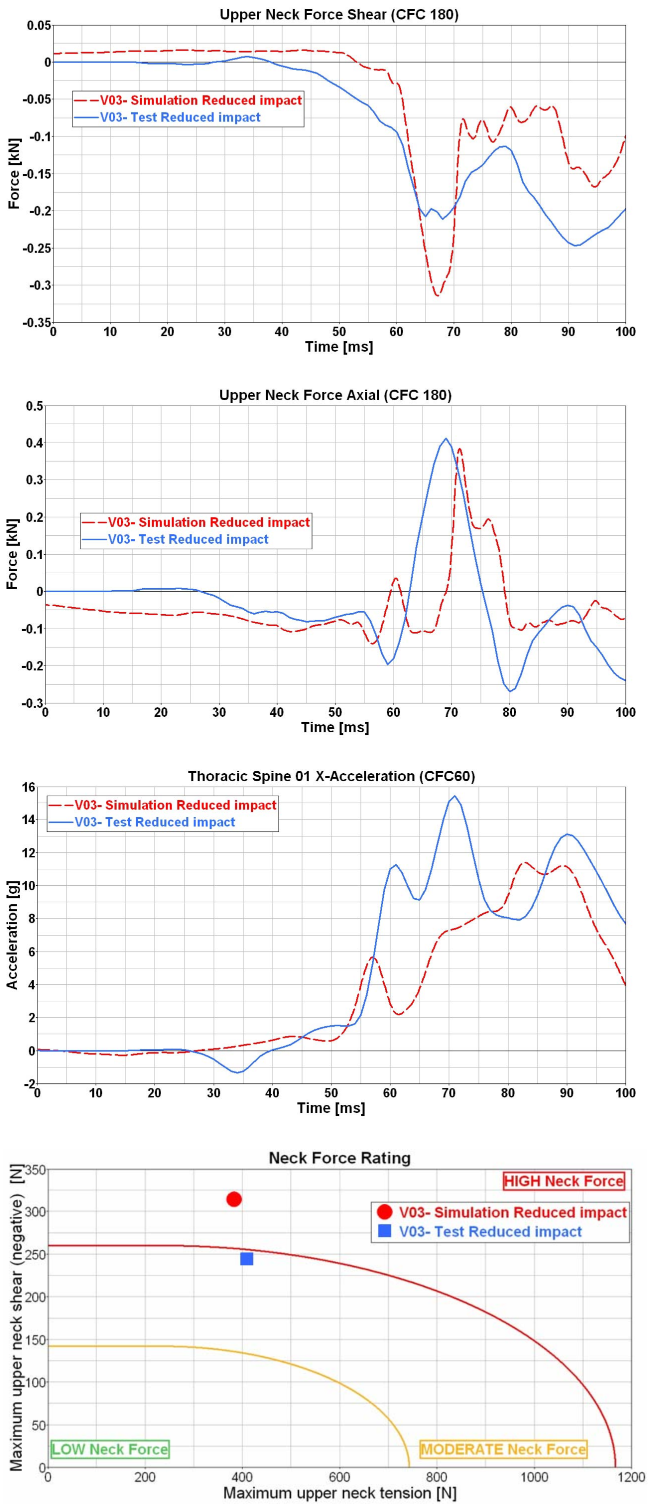


Figure 24. Validation result for test V03.

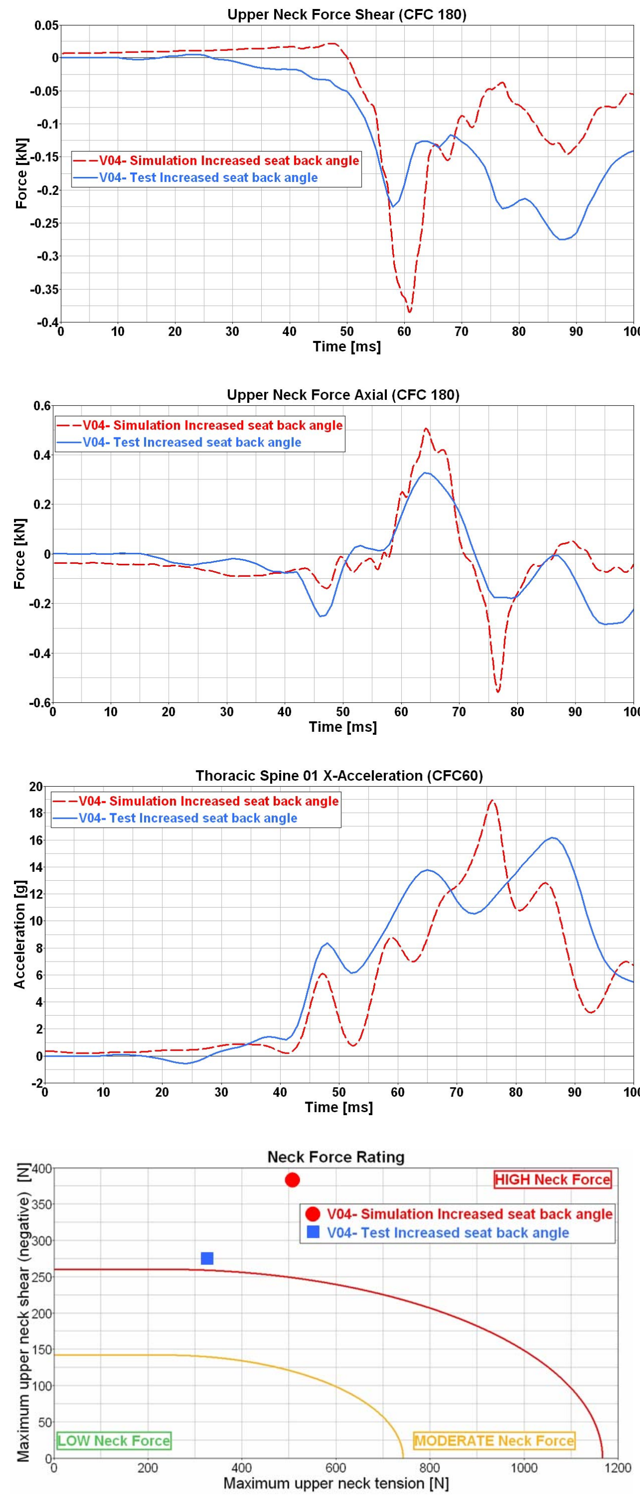


Figure 25. Validation result for test V04.

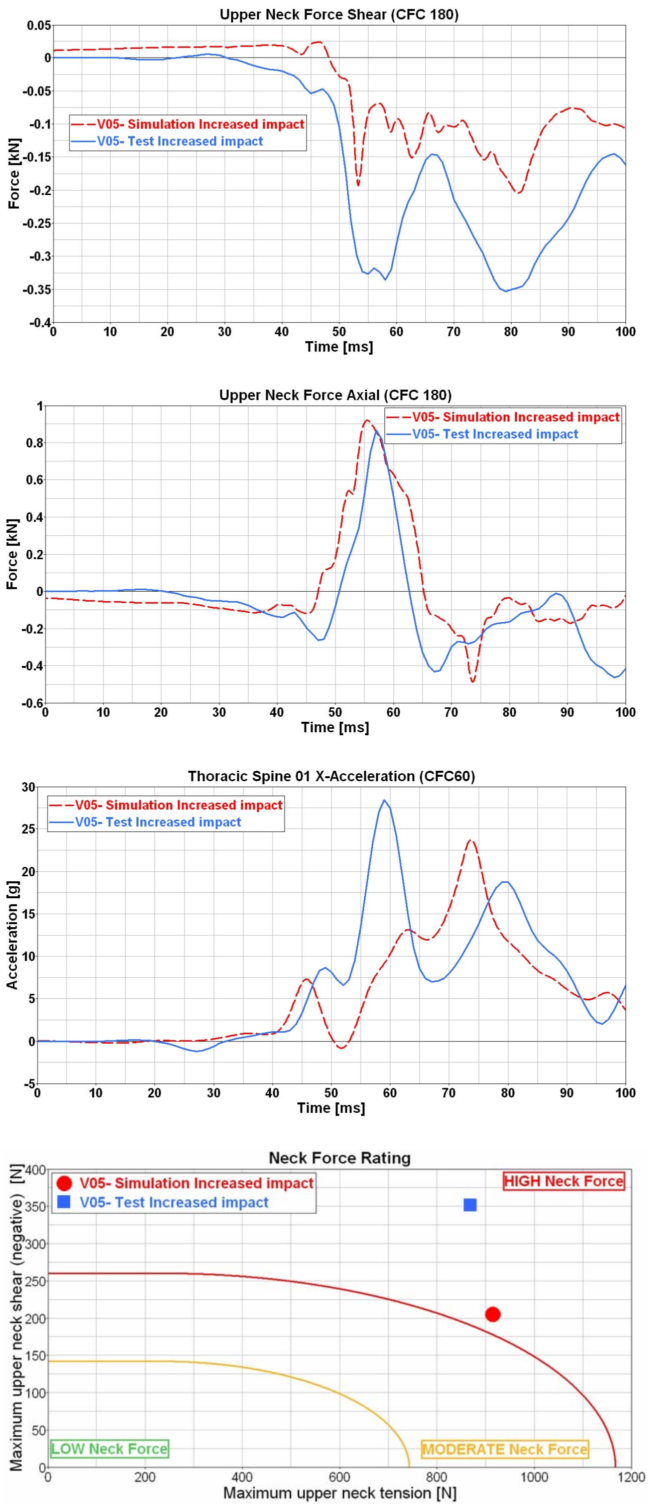


Figure 26. Validation result for test V05

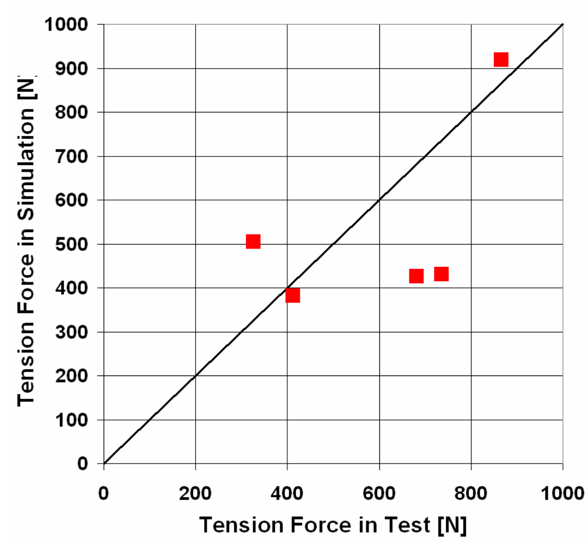


Figure 28. Upper neck shear forces.

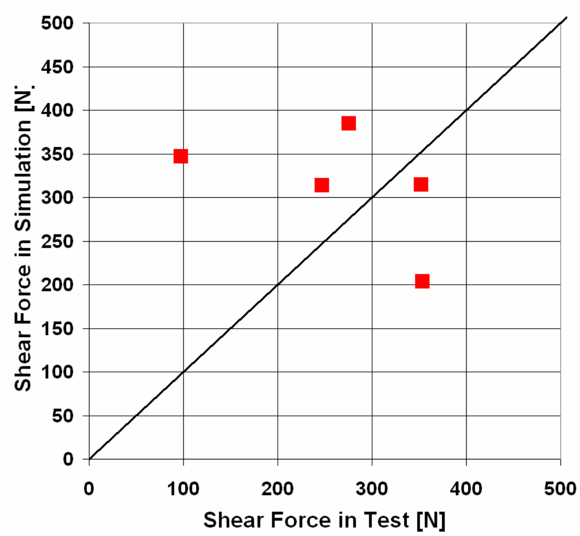


Figure 29. Upper neck tension forces.

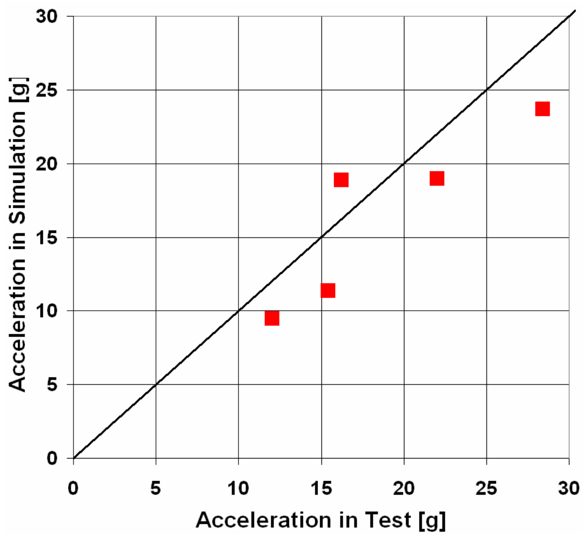


Figure 27. Thorax T1 accelerations.

	Upper Neck Shear Force	Upper Neck Tension Force	T1 Acceleration
V01	358%	59%	79%
V02	89%	63%	86%
V03	127%	93%	74%
V04	140%	155%	117%
V05	58%	106%	83%

Figure 30. IIWPG-relevant criteria from simulation in relation to test results (100%)

When comparing the characteristics of the dummy loading curves obtained from testing and calculated with the numerical model the criteria measured in the pelvic region corresponded well. The same holds true for the head accelerations. However, for the body regions which are relevant for IIWPG assessment the loading characteristics show less agreement. An overview of the extrema from the simulation model in relation to those measured in the tests is given in Figures 27 to 29 and Figure 30. Data from the rebound phase after the head contact with the head restraint had ended were not considered for this evaluation.

The IIWPG protocol assesses only positive shear force values and axial tension forces, i.e., a rearward relative motion of the head relative to the torso. The BioRID II dummy typically displays this behavior in rear-impact tests with production car seats. However, also negative shear forces occurred in some of the validation tests during the relevant phase of head-to-head restraint contact. This can be attributed to the particular configuration chosen for the Chalmers seat where the head restraint is closer to the head than usually found in production seats. Therefore and in contrast to IIWPG practice, the present study takes into consideration also negative shear values when illustrating simulation results in the rating chart and to assess the validation quality.

The axial forces are represented quite well by the simulation model, both in general character of the graph and in magnitude of the peak values, except for the test condition with increased backset V01 (see Figure 22 to 26). The peak values of the shear force differ significantly from those measured in the tests in some cases, but the times that the peaks occur correspond between simulation and experiment. This supports the visual ascertainment that the overall motion of the dummy model coincides with that of the test dummy. Again, the largest deviations between simulation and test curves are found in the situation with increased backset.

When analysing the longitudinal acceleration graphs of T1measurements most tests demonstrate a pronounced first peak at the beginning of the dummy loading phase. This peak is possibly caused when the head restraint accelerates the head and this signal is transferred through the cervical spine to the thorarcic spine. It is not prominent in the graphs obtained with the simulation model. Apart from this peculiarity, the T1 longitudinal accelerations are well represented by the numerical model in general character and in magnitude.

The complete simulation model which was validated only on the basis of test V02 predicts also the biomechanical loadings measured in test V03 (reduced crash severity) quite well. The conformity with the test results from V05 (increased seat back angle) and V06 (increased crash severity) is limited mostly due to the deviations in neck shear force. The test scenario which included a very large backset and modified initial dummy position (V01) is reflected only roughly in simulation although characteristic points in the kinematics between test dummy and model match.

DISCUSSION AND CONCLUSIONS

For the build-up of a numerical simulation model to study the influence of seat design factors on occupant loadings under rear-impact a number of validation tests were conducted. The chosen test set-up allows to adjust the crash pulse by means of a simple and cost-efficient method. Using a system which accelerates a stationary carriage with the seat and dummy enables to accurately measure the dummy position as it is present when the impact occurs. This is a pre-requisite for the subsequent simulation model validation. Five tests were conducted with the aim to cover a range of major influential factors on the occupant kinematics and biomechanical loadings. Accordingly, the crash severity and the seat geometry were varied in the tests. This allows to evaluate the prognostic capabilites of the simulation model for a range of test conditions.

The generic seat design of the Chalmers seat employed in the dynamic experiments facilitates the adjustment of seat parameters independently of each other. The structure and geometry of this test device differs from those of many production car seats, e.g., in the area of the seat cushion and the connection between between seat back and head restraint. The dynamic behavior of the Chalmers seat may therefore deviate in detail from that of a production seat or at least require extensive pre-testing to adjust its characteristics. Particularly, it was found that the seat configuration used for some of the validation tests differed in behavior from most car front seats. The head restraint caught the dummy's head very early which resulted in uncommon neck shear force characteristics.

The tests provided input data to describe the experimental set-up in a numerical simulation model, mainly in multi-body code MADYMO. Particular elements of the seat model were realized in FE code and major surfaces which come in contact with the dummy model were carried out as facet surfaces to allow realistic interaction with the body parts. The model of the employed BioRID II anthropometric test device is based on multi-body code, but features a facet representation of the body surfaces. However, a considerable amount of validation work on certain body regions of the original dummy model was necessary to replicate adequately the interaction with the seat as observed in the experiments. The finite element version of the dummy model which was also available at the time of the study did not improve the simulation results significantly and was therefore abandoned. FE model updates that have entered the market in the meantime demonstrate more promising quality. Investigations of the model robustness revealed that the time step size for the calculation should not fall short of one microsecond. Future efforts to optimize the model should focus on improving the dummy model and its interaction with the seat surface. It is also possible that the difficulties encountered with the inconsistencies in the dummy back in conjuction with the Chalmers seat are less pronounced when production seats with a softer and continious seat back surface are used.

A recent MADYMO model version (MADYMO 6.3.2, facet dummy version 2.1) of the BioRID II with a water-filled abdominal cavity and jacket in FE method was implemented in the existing model. It indicates that the described difficulties in the dummy back region are largely overcome, but would have required a completely new validation of the thoracic and cervical spine area. Hence, this approach was not further pursued in the course of the present study.

Another alternative is the utilization of a complete finite element representation of the BioRID II dummy which is also available on the market. Using the experience from the MADYMO-based simulation model it is planned to set up also a finite element version in LS-DYNA code. Future research in this field should therefore include a systematic evaluation of the different model approaches.

The calculated biomechanical loadings from the current simulation model correspond quite well with the characteristics of the measured accelerations and forces on the test dummy, especially for the head and the pelvic area. However, some significant deviations are evident in the neck and upper thoracic area which currently prohibit to use the model to forecast IIWPG-relevant loadings. Nevertheless, the model can be used to predict tendencies when seat design parameters or crash severity are varied. It can therefore be considered a suitable research tool to study the influence of fundamental rear-impact factors. However, it should be utilized only within the range of the tested conditions and should regard the limitations of the test set-up and the anthropometric test device. Possible subjects of application are the investigation of different crash pulse characteristics, seat back or head restraint positions on the biomechnical loadings of the BioRID II. A number of parameter variations with regards to the seat position has been conducted already which indicate positive effects on the seat performance.

The application of the model in a vehicle environment including a particular car seat could not be realized during this study. Provided that sufficiently detailed MADYMO seat models are available this offers possibilities for enhancing the safety of present or future seat concepts.

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