EFFECT OF HEAD RESTRAINT SYSTEM ON RISK FOR WHIPLASH INJURY IN LOW SPEED REAR IMPACT AND CORRELATION OF BIORID II TEST RESULTS USING LS-DYNA/MADYMO COUPLING

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ABSTRACT

Seat and especially head restraint design have a significant influence on risk for whiplash injury in low speed rear impact. For seat optimization, especially during early product development, a proper simulation of dummy behavior using finite element analysis (FEA) is important. For development and evaluation of the FEA methodology, several seats were equipped with occupant activated (re-)active head restraint riACT™, and tests were performed on a HyGe acceleration sled using both the IIWPG 16 km/h triangular pulse and IIWPG test protocol. A correlation between BioRID II dummy behavior in sled test, and LS-DYNA coupled Madymo BioRID II facet dummy behavior is demonstrated. After FEA correlation results interpretation, a study of active head restraint robustness is illustrated.

Keywords: REAR IMPACT, HEAD RESTRAINTS, WHIPLASH, NECK, SHEAR, and FINITE ELEMENT ANALYSIS

THIS ARTICLE IS A CONTRIBUTION to the ongoing activity to determine the most effective options to prevent low speed whiplash injury. Head restraint (HR) need to be assessed in order to better understand the pros and cons of the best possible options for effectiveness and robustness in reducing costly whiplash injuries for passive HR systems and active anti-whiplash HR systems. Passive solution for anti-whiplash most often means having a well-tuned seat system with relatively low backseat, i.e. head to HR gap. Active solution means having an active HR system that deploys to reduce head to HR gap during a rear impact crash event, allowing higher initial backset.

In this article, two tests were executed according to International Insurance Whiplash Prevention Group (IIWPG) protocol for developmental purposes and assessment of a passive solution versus an active solution. The first test is without active HR and the second test is with active HR. Occupant kinematics and BioRID II dummy neck forces, along with different occupant injury criteria, were used for occupant injury assessment in the two tests, see Figure 1 and Table 1.

Furthermore, an additional test with an active HR was performed to assist in development of FEA methodology. LS-DYNA/MADYMO Coupled model with MADYMO BioRID II facet dummy was tuned to the actual test. Some aspects of FEA correlation are shown for purpose of illustration. Upon completion of FEA correlation, an active HR robustness study for backset parameter change was performed to confirm that up to a reasonable backset the active HR is very effective.

PASSIVE VERSUS ACTIVE SYSTEM, AND ROBUSTNESS OF ACTIVE SYSTEM

The percentage improvement in low speed rear impact test evaluation criteria with and without active HR system is shown in Table 1. From a similar test Figure 1 illustrates the difference in neck deformation and effectiveness of the active head restraint. The apparent difference in neck curvature, Figure 1, which is shown as qualitative assessment evaluation criteria, and Table 1, support the thesis of active HR effectiveness in reducing risk of whiplash. The passive and active HR samples were tested in the same seat structure to ensure comparability between tests, test setup conditions were the same for both types of HRs

Table 1 - Active v. Standard HR per IIWPG Protocol (Source: Johnson Controls GmbH)

Evaluation Criteria	Reduction [%] Active vs. Passive
Backset BioRID [mm]	-
NIC [m²/s²]	-51%
Upper Neck F_x [N]	-100%
Upper Neck F_z [N]	-45%
Upper Neck M_y [Nm]	-59%
Nkm criterion	-31%
LNL criterion	-59%

for all key parameters, including backset and HR height. For the seat without active HR, backset was 52 mm and for the seat with active HR 54 mm. From Table 1, neck injury criteria NIC for active HR decreased 51% compared to passive HR system. The upper neck shear force F_x decreased almost to zero, i.e. 100%, which is certainly promising for many occupants concerned about whiplash injury. The upper neck axial force reduces by approximately 59% for active system. Other injury criteria like upper neck Moment M_y, Nkm, and LNL follow the same trend, see Table 1. The high effectiveness of active HR is due to the fact that the activation is in the lower seat, with a higher mass actuation of the system (pelvis and leg) and not influencing the upper torso region (no torso acceleration / neck force) for actuation.

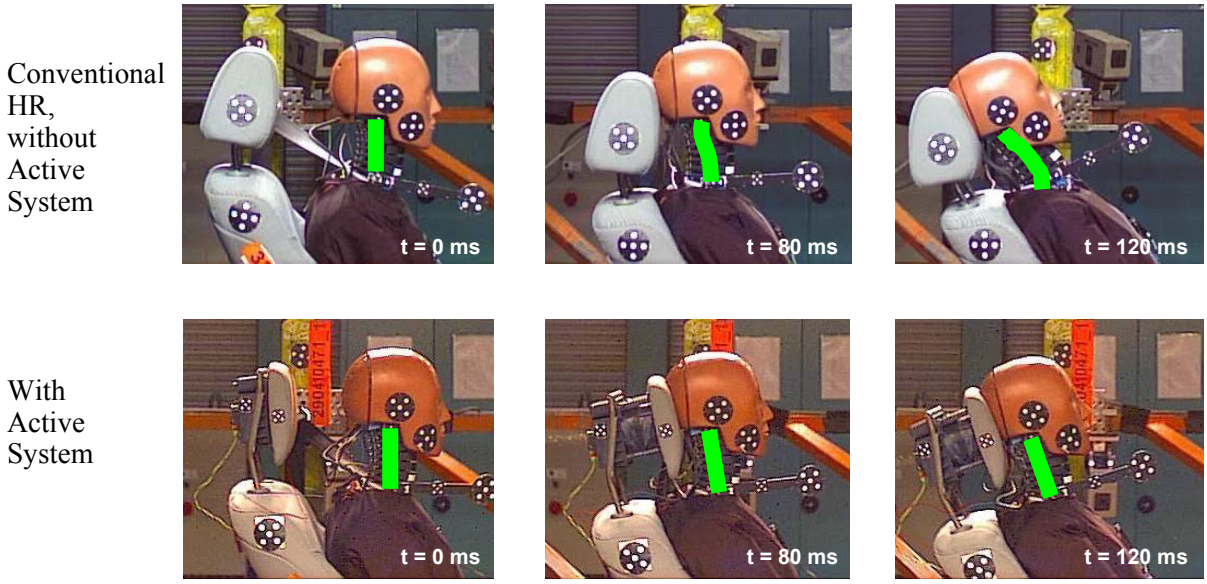

Fig. 1 – Neck Deformation in Low Speed Rear Impact with and without Active H/R, (Source: Johnson Controls GmbH)

FEA Modeling of Low Speed Rear Impact is the next step in assessing the active system robustness to a reasonable backset value. Figure 2 illustrates modeling of IIWPG low speed rear impact using LS-DYNA and MADYMO coupled code with BioRID II facet dummy. For purpose of correlation the third test was performed with a different seat structure and active HR. Test setup parameters were recorded and used as an input to the FEA solver. Included in Figure 2 is the image of the set-up for the IIWPG sled test.

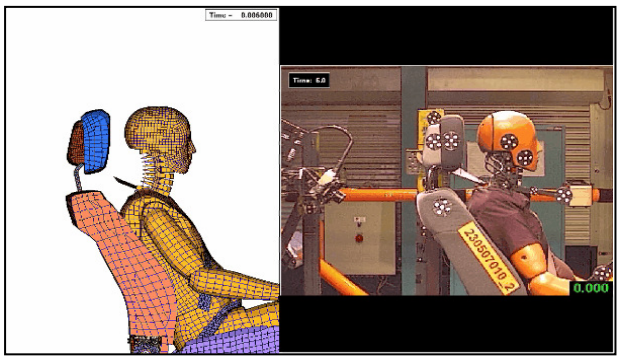

Fig. 2 – LS-DYNA/MADYMO Model and Test (Source: Johnson Controls Inc.)

Table 2 shows some of the key parameters achieved in correlation to test with 44 mm backset. For example, achieved contact time is very close, within 1.2 milliseconds, while upper neck axial force shows largest difference. Figure 3 illustrates some of the parameters considered in correlation activity. It must be noted that even two or more tests, on the same samples and with controlled test setup, may produce similar although not the same results. The complete understanding of FEA modeling limitations is the key for the results interpretation. Some of the limitations are just true representation of looseness in the test sample. The looseness primarily comes from joints and interfaces. For example, looseness in HR rods, HR sleeves to upper cross member interface, and free play in recliner, seat lifter mechanism, and tracks. Moreover, there are limitations in foam stiffness and foam strain rates modeling. The above mentioned FEA modeling limitations can cause differences in both response to impact trends and response to impact magnitudes. In general, the FEA model reacts sooner to the occupant loading than actual test, due to limitations in free play modeling, as shown in the head x-acceleration, T1 x- acceleration, upper neck F_x and upper neck F_z in

Figure 3. In more detail, the difference in T1 acceleration may be attributed to free play in recliner, seat lifter mechanism, tracks and in fastener joints.

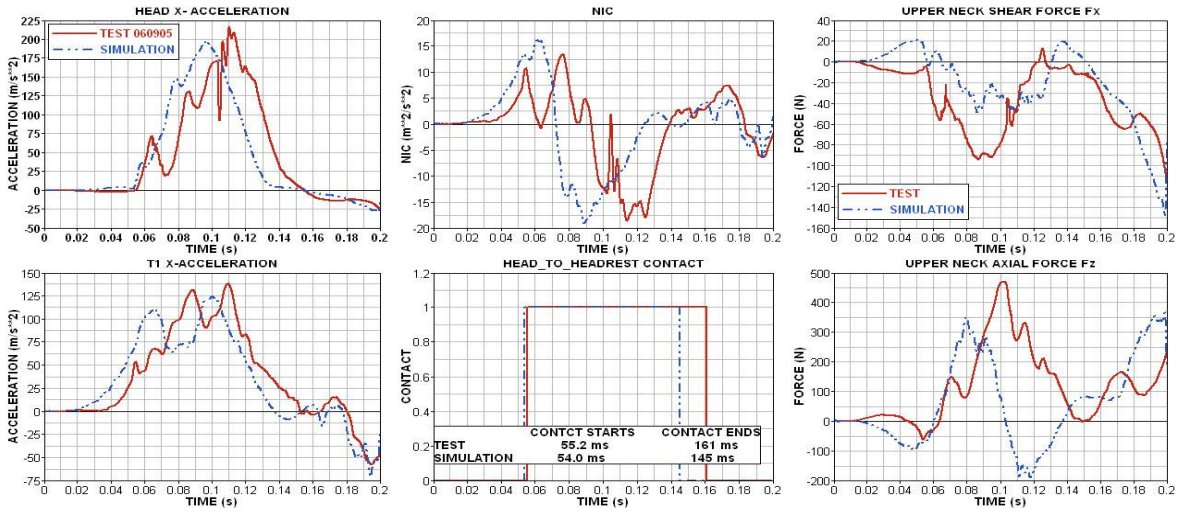

Fig. 3 - Test v. FEA Simulation Comparison (Source: Johnson Controls Inc.)

Moreover, responses to impact in simulation tend to have less noise compared with actual test, which is another consideration in the results interpretation. Considering all the assumptions built in the FEA model and complex nature of crash event, magnitudes of the response to impacts are very close, with exception of upper neck F_x and upper neck F_z. The upper neck F_x shows initial different trend then quickly reverses the trend to come in phase with actual test. This can be attributed to already discussed limitations in modeling, and possibly in FEA BioRID II dummy calibration. The upper neck F_z shows initially similar trend compared with test, but after 80 milliseconds shows both issues with magnitude and trend. Additional data point is used to ensure validity of FEA modeling, see Figure 4. Cervical Spine C4 accelerometer is the closest to upper neck accelerometer and x-acceleration shows both good trends and good magnitude compared to test. Good agreement of C4 acceleration in Figure 4 and Head x-accelerations in Figure 3 indicates possible issue with FEA dummy calibration for forces, only. The fact that FEA is a controlled environment, and knowing that response to impacts should behave similarly in the future runs, upper neck F_x and upper neck F_z can be used for directional prediction only.

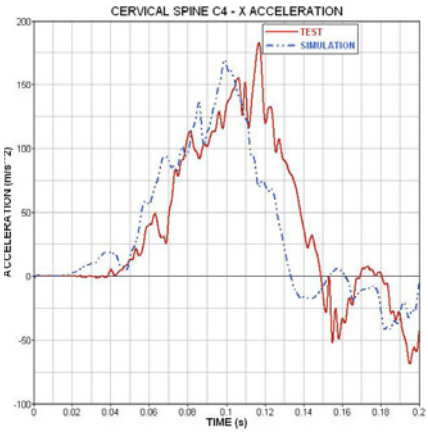

Fig. 4 - Cervical Spine C4 X-acceleration, Source: Johnson Controls Inc.

ACTIVE HR ROBUSTNESS ASSESSMENT VIA FEA MODEL is a logical next step. Table 2 shows the robust performance for the backset in range from 30 millimeters to 90 millimeters. This backset is achieved by adjusting HR rods geometry, changing only backset and keeping the stiffness of upper cross member and head restraint rods constant.

Overview of the performed test and FEA simulations is given in Figure 5 in terms of IIWPG evaluation chart. Marker #1 shows the test results without active HR for backset of 52 mm, i.e. just conventional/standard HR, marker #2 shows the test with active system which had almost no shear force on the upper neck, marker #3 shows the test that was used for FEA correlation. Markers #2 and #3 depict the test results for 2 different seats, both seats with active system. Good results are accomplished with active system independent of the seat structure, which indicates the robustness of the active system. Marker #4 shows the results for FEA iteration with backset of 30 mm, 44 mm, 50 mm, and 70 mm. Marker #5 shows the active system with results for backset of 90 mm, still well within the good zone. The low variation of the biomechanical values for backsets between 30 to

70 mm results from the design of the active HR system, which has variable deployment for only the necessary travel to achieve contact with the head. The deployment velocity depends on impact speed, but deployment velocity for a given impact speed is independent of backset.

Table 2 - Test and FEA Correlation Results; and Performance for Different Backsets (Source: Johnson Controls Inc.)

	Backset (mm)	Contact Time (msec)	T1 Acceleration (g)	Up. Neck Shear F_x (N)	Up. Neck Ax. F_z (N)
FEA	30	50	12.6	20.7	299
TEST	**44**	**55**	**14.1**	**12.7**	**468**
FEA	**44**	**54**	**12.6**	**19.5**	**348**
FEA	55	58	12.4	21.0	349
FEA	70	61	12.7	26.1	337
FEA	90	66	12.2	60.8	404

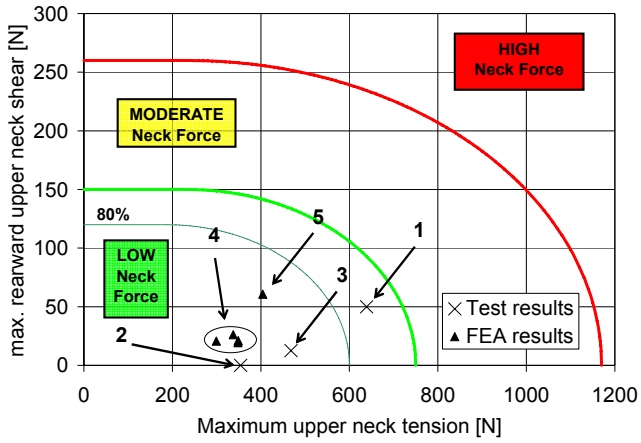

Fig. 5 - Neck Forces Scale per IIWPG (Source: Johnson Controls Inc.)

From Table 2, the T1 acceleration in FEA does not change much regardless of backset since T1 is primarily function of seatback stiffness. Seatback stiffness, including recliner characteristics, is represented in the detailed FEA model with already discussed limitation of free play modeling. However, other values like contact time, upper neck F_x and upper neck F_z show very reasonable trends. Those trends have a very small gradient up to 70 mm of backset and a larger gradient just above 70 mm of backset. The larger gradients above 70 mm of backset indicate the limitation of the active HR effectiveness at high head to HR distances. Considering that upper neck shear force F_x and upper neck axial force F_z are used for directional prediction only, small percentage of F_x and F_z change for a change of backset indicate active HR robustness, which is likely well within LOW Neck Force zone of Figure 5.

CONCLUSION Both testing and FEA show that upper neck shear force F_x can be significantly reduced and even eliminated with a proper active HR device and axial force F_z can be controlled with a proper active HR position. A clear understanding of limitations in FEA modeling methodology is important in the interpretation of the FEA results. Both successful FEA correlation and understanding of IIWPG low speed rear impact is achievable. This FEA methodology can reduce the number of required sled tests, and reduces seat development cost and time by allowing tuning of design through FEA iterations. This in turn yields robust design for whiplash prevention, whereas tuning in sled test is only feasible in a limited way as test repeatability may make design improvements difficult.

Current calibration of Madymo BioRID II Facet dummy is sufficient for good understanding of the rear impact event and to drive appropriate design changes. Nevertheless, improvements are desirable. Namely, TNO/MADYMO facet BioRID II dummy calibration enhancements in terms of magnitude of neck forces, neck moments and pelvis acceleration are deemed necessary to obtain even better results.

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