Simulation Method for Whiplash Injury Prediction Using an Active Human Model

Jeroen Broos[1], Riske Meijer[1]

Abstract Despite improvements in seat design over the last decade for rear-impact crash scenarios, the burden for insurance companies seemed to be so high that in several EU countries the legislation concerning claims for whiplash injury has recently been made more stringent. Many studies have been performed in an attempt to understand the injury mechanisms involved. This resulted in scientific publications of several different theories about the injury mechanisms and injury criteria.

In this study a simulation method was developed for whiplash injury prediction that takes into account injury criteria covering several whiplash injury mechanisms by using an active human model. Two passenger car seats were modelled and validated on component level as well as against BioRID II sled tests under various loading conditions. In order to test the sensitivity of the developed simulation method rear-impact simulations were performed with the active human model in conditions where parameters known to affect the whiplash injury risk were varied. Here, for each condition simulated the injury criterion showing the highest injury risk was assumed to give the overall injury risk on WAD2+.

The developed simulation method showed similar effects on the risk of whiplash as known factors found in literature. Therefore, it was concluded that this simulation method can be used to further optimise whiplash assessment procedures and passenger seat designs for a more robust protection in the large range of real world rear-impact accidents.

Keywords Whiplash, neck muscles, rear impact, human modelling, injury criteria.

I. INTRODUCTION

It is estimated that 800 000 EU citizens suffer whiplash injuries annually. Of these injuries 40 000 result in long term suffering with an associated socio-economic impact of approximately 10 billion EUROS per year and account for approximately 70% of all injuries leading to disability sustained in vehicle crashes [1]. Real world data analysis revealed that existing whiplash protection concepts have a 50% reduction in long-term whiplash risk [2]. Despite the improvements in seat design in order to reduce the risk to sustain a whiplash in a rear-end impact, the burden for insurance companies seemed to be so high that in several EU countries the legislation concerning claims for whiplash injury has recently been made more stringent. The reason for the high number of whiplash claims is said to be partly due to the fact that medical science is unable to properly classify whiplash injuries in patients concerning the soft tissues (severe neck injuries, such as fracture or dislocation of vertebrae is visible on X-rays or MRI), and as such can only rely on the patients' complaints, creating a situation in which it is possible for inappropriate claims for pain and impairment to be presented to the motor insurers [3]. However, there are strong indications that chronic neck pain is caused by psychological muscle reaction resulting from high initial pain rather than the initial injury [4][5]. Therefore, further research and developments to prevent whiplash as well as to treat whiplash stays important.

Whiplash is a term used to describe the motion of the head-neck complex during a rear-end car impact, which results in symptoms related to certain damage of neck tissues. Therefore, the Quebec Task Force (QTF) classified Whiplash Associated Disorders (WAD) based on signs and symptoms [6]. For the purpose of preventing as well as curing whiplash injuries, many studies have been performed in an attempt to understand the injury mechanisms involved, of which the most cited ones are summarised below.

Aldman [7] hypothesised that hydro-dynamic pressure phenomena in the central nervous system (CNS) could be a potential injury-causing factor to the cervical spinal ganglia. This was based on the observed symptoms, such as weakness or abnormal response in the parts of the body (mainly the neck, shoulders and

[1]) Jeroen Broos (jeroen.broos@tno.nl, Tel. +31 888663476) is a Research Scientist, and Riske Meijer is a Senior Research Scientist at TNO, Netherlands Organization for Applied Scientific Research.

upper back) that are connected to the CNS via the cervical nerve-roots, as well as vision disorders, dizziness, headaches, unconsciousness and neurological symptoms in the upper extremities. Svensson *et al*. [8]-[9] proved this hypothesis from pressure measurements in the spinal canal of an anaesthetised pig. Next, Boström *et al*. [10]-[11] developed the Neck Injury Criterion NIC, which is based on a correlation between the pressure in the spinal canal and the acceleration and velocity of the occipital condyles relative to the first thoracic vertebra.

Yang *et al*. [12] and Deng *et al*. [13] hypothesised that the axial compression together with the shear force in the first phase of the impact, which results in an extension-flexion motion or S-shape of the cervical spine, are responsible for the higher observed frequency of neck injuries in rear-end impacts versus frontal impacts of comparable severity. The axial compression first causes loosening of cervical ligaments, making it easier for shear-type soft tissue injuries to occur. Tests with C1-T1 specimens showed that shear stiffness values were reduced significantly with increased axial compressions [12], and that shear forces could be harmful to the facet joints, in particular in the upper neck region (occipital condyles) [13]. Based on this hypothesis, Schmitt *et al*. [14] developed the neck injury criterion N_{km}, based on the hypothesis that a neck injury in rear impact is caused by a linear combination of loads and moments at the upper cervical spine. Based on accident reconstructions with a BioRID II, Kullgren *et al*. [15] found a good correlation for NIC and N_{km} to neck injuries.

Yoganandan *et al*. [16]-[18] hypothesised that extension of the neck during rear-impact loading leads to stretch of the facet joint capsule. Based on that, Stemper *et al*. [19]-[20] performed tests with isolated head-neck post-mortem human subjects (PMHS), which were subjected to various rear-impact loading conditions. The results showed that the lower levels of the cervical spine are more likely to be injured at higher rear-impact input velocities than the middle and upper levels, and proposed neck injury criteria based on lower neck loads, anterior-posterior shear force, tension force and extension and flexion moments.

Ono *et al*. [21]-[22] hypothesised that excessive deflection between the cervical vertebrae cause neck injuries. Ono *et al*. [23] used human modelling and volunteer testing, including X-ray cineradiographic imaging, as well as accident reconstructions to develop strain-based injury criteria and limits for cervical segments C2/C3 to C6/C7. As it is difficult to adapt a rear-impact dummy such that principal strain, shear strains and strain rates between cervical vertebrae replicate that found in a human neck, Ono *et al*. [23] related the newly developed injury criteria to physical quantities. Comparing the synchronisation of the peak of the strain-based injury criteria to that of the selected physical quantities, NIC and the neck tension and backward shear forces and bending moments, showed a similar trend, i.e. timing of peak takes place during the interaction of head with the head restraint. For this reason these injury criteria were considered as appropriate neck injury evaluation parameters, and limits for 5% and 95% risk on WAD2+ were defined for these quantities.

The hypothesised injury mechanisms differ in vertebral level, tissue and the timing (impact phase, head-neck motion) during which the injury occurs. The ways of proving the hypotheses, e.g. PMHS tests, volunteer tests, human modelling, accident reconstruction, each have their own limitations for determining whiplash injury mechanism. The authors hypothesised that the injury mechanism depends on several factors, like severity of acceleration, seat design, occupant position and the build of the person, i.e. difference in neck length, muscle strength, ligament strength, head weight, etc. In other words, on tissue level the loads can differ per case, and therefore the authors hypothesise that all above-described injury mechanisms and derived injury criteria could be valid. On top of that, several factors seem to increase the whiplash injury likelihood and/or severity, such as female gender [4][23]-[29], higher impact severity [25][27][30]-[31], wrong head restraint position [32]-[34], lack of awareness [29], and rotated head positions [28]-[29]. Physical test methods, crash test dummies, and dummy models have the limitation that only a few of the mentioned variations can be covered.

The objective of this study was to develop a computer simulation method using an active human model that takes into account the injury criteria covering the above-mentioned whiplash injury mechanisms as well as muscle stress, and thereby enabling further enhancement of seat designs, anti-whiplash systems, as well as accident reconstructions. Earlier published simulation methods using a whole body computer human model [35]-[37] did not cover all above mentioned injury mechanism, nor muscle stress, but mainly had the purpose to get insight in possible injury mechanisms.

For the developed simulation method, seat models based on existing seats were used along with an active human model with detailed neck. Tests with a rear-impact dummy were used as verification for the developed seat models. All above described injury criteria, with proposed limits, were used. From injury criteria limits, the injury risk was calculated. The effect of various known whiplash risk affectimg parameters on the injury criteria output and the overall injury risk was studied.

II. METHODS

Simulation models

Two seat models were developed, a poor rated (seat 1), as well as a fair rated (seat 2), according to Euro NCAP whiplash score (<2 points versus >2 points, respectively) [38]. First, a mesh was made from the two seats using 3D scans. Next, numerical models were made from the seats in the multi-body and finite element software package MADYMO version 7.4.1 [39]. The mesh of each seat was divided into different components (seat base, seatback and head restraint). Each component was rigidly connected to bodies that were connected to each other using bending and/or translational joints. The contact forces of the seat cushions and joint characteristics were modelled by means of defining stress-strain characteristics. Also, a three-point belt system was modelled based on known belt characteristics. The various cushion components (seat base front, seat base mid, seat base rear, seatback low, seatback mid, seatback top, head restraint front, head restraint top, side wings) of the two seat models were characterised using quasi-static indentation tests. The seat cushion characteristics were validated using quasi-static tests with human shape impactors (head, back and buttocks), as well as dynamic impact tests with head and back shape components. The joint characteristics were characterised and validated using quasi-static H-point machine impactor tests with human-shaped components (head and back).

Furthermore, the whole rear-impact performance of both seats was validated using available Euro NCAP rear-impact sled tests with a BioRID II. In addition, for validation of the seat models' rear-impact performance with the head restraint in the lowest and highest position and occupant in bent forward position, sled tests with a BioRID II were performed with each of the seats with the head restraint in the lowest and highest position, and with the BioRID II in bent forward and nominal position. Table I provides an overview of the sled tests used for validation of the two seat models, and Fig. 1 and Fig. 2 show the initial position of the BioRID II in test 5 and test 10, respectively. The BioRID II was instrumented according to the Euro NCAP dynamic whiplash test protocol **Error! Reference source not found.**. Furthermore, for each test condition one of the three pulses from the Euro NCAP Rear Whiplash Test Protocol was applied (see Table I and Fig. 3).

TABLE I

REAR-IMPACT SLED TESTS USED TO VALIDATE THE SEAT MODELS

Test #	Seat #	ATD	ATD condition	Head restraint position	Acceleration
1	1	BioRID II	Nominal	Nominal	SRA-16 (Low)
2	1	BioRID II	Nominal	Nominal	IIWPG-16 (Mid)
3	1	BioRID II	Nominal	Nominal	SRA-24 (High)
4	1	BioRID II	Nominal	Highest	SRA-16 (Low)
5	1	BioRID II	Bent forward	Lowest	SRA-24 (High)
6	2	BioRID II	Nominal	Nominal	SRA-16 (Low)
7	2	BioRID II	Nominal	Nominal	IIWPG-16 (Mid)
8	2	BioRID II	Nominal	Nominal	SRA-24 (High)
9	2	BioRID II	Bent forward	Highest	SRA-16 (Low)
10	2	BioRID II	Nominal	Lowest	SRA-24 (High)

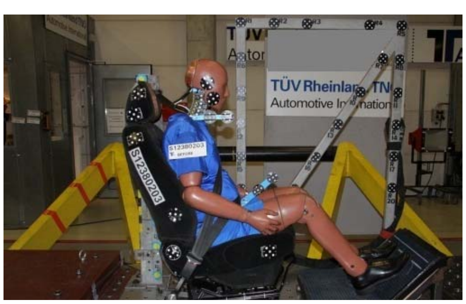


Fig. 1. Initial position of rear-impact test 5 with seat 1, lowest head restraint position and BioRID II bent forward.

Fig. 2. Initial position of rear-impact test 10 with seat 2, lowest head restraint position and BioRID II in nominal position.

To develop a simulation method that can take into account several plausible whiplash injury criteria and real-life safety rear-impact scenarios, the MADYMO active human model (version 1.1) [41] was used. This model was selected because it has a detailed neck, including musculature, that was originally developed to study the whiplash injury mechanism [42], and it is validated for multi-directional crash (PMHS tests) and pre-crash loading situations (living human subject tests) [44]. Also, this model contains posture maintenance [43] that made it possible to simulate various initial positions, including initial muscle stress in the neck, which could affect the neck kinematics and as such the output of the whiplash criteria.

Whiplash injury risk prediction

As mentioned in the Introduction, various plausible hypotheses are available about the cause of whiplash injuries. The injury criteria covering these different hypotheses, for which limits have also been defined, were taken into account in the simulation method for whiplash injury assessment. Table II summarises the selected whiplash injury criteria with injury limits related to 5% and 95% WAD2+ injury risk.

Injury risk curves were developed for all presented criteria by interpolating between 0% and the 5% and 95% injury risk limits, as shown in Fig. 4. The whiplash injury risk resulting from each individual criterion was determined using the injury risk curve and the peak response of each criterion during a simulation. The injury responses were all shifted in a way that the output was zero at the start of a simulation, with the active human

TABLE II

WHIPLASH INJURY CRITERIA WITH ACCOMPANYING INJURY LIMITS

Criterion	Limit	WAD2+ risk		Injury mechanism
Upper neck backward shear force	340 N 730 N	5% 95%	[23]	Facet joint capsule damage by excessive load or motion
Upper neck tension force	475 N 1130 N	5% 95%	[23]	Facet joint capsule damage by excessive load or motion
Upper neck flexion moment	12 Nm 40 Nm	5% 95%	[23]	Facet joint capsule damage by excessive load or motion
Upper neck extension moment	12 Nm 40 Nm	5% 95%	[23]	Facet joint capsule damage by excessive load or motion
Lower neck backward shear force	340 N 730 N	5% 95%	[23]	Facet joint capsule damage by excessive load or motion
Lower neck tension force	257 N 1480 N	5% 95%	[23]	Facet joint capsule damage by excessive load or motion
Lower neck flexion moment	12 Nm 40 Nm	5% 95%	[23]	Facet joint capsule damage by excessive load or motion
Lower neck extension moment	12 Nm 40 Nm	5% 95%	[23]	Facet joint capsule damage by excessive load or motion
NIC	8 m^2/s^2 30 m^2/s^2	5% 95%	[23]	Damage to spinal ganglia by sudden pressure pulses
N_{km}	0.5 1.75	5%* 95%*	[15]	Ligament damage at occipital condyles by excessive shear forces and moments acting on the upper cervical spine

*N_{km} injury limits are related to risk on whiplash with symptoms longer than one month.

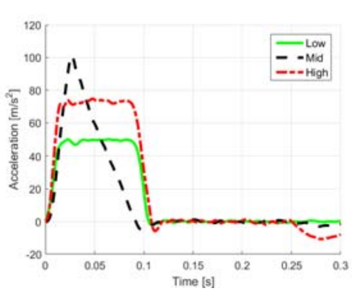


Fig. 3. Acceleration pulses used for the rear-impact sled tests and simulations.

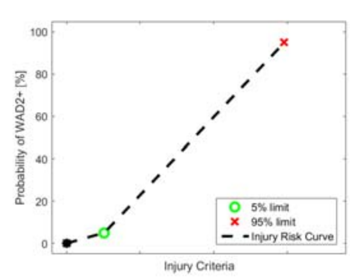


Fig. 4. Injury risk curve determination from 5% and 95% injury limits.

model in initial position. The peak response of each criterion was measured until the rebound phase of the simulation, which was determined by the moment when the head loses contact with the head restraint. The total whiplash injury risk resulting from a simulation was determined by the criteria from Table II, resulting in the highest whiplash injury risk. The risks of the criteria were not combined to determine the total whiplash injury risk as the different criteria were not always completely independent variables.

Simulations

The limits of nine of the 10 injury criteria presented in Table II were based on human volunteer and numerical human model simulations [23], and can therefore be used directly for the active human model used in this study. However, the N_{km} injury criterion was developed by accident analysis with a BioRID II model [15] and therefore cannot be used directly for the active human model. For this reason the BioRID II sled tests, which were used to validate the seat models (see Table I), were repeated with the active human model and the output of the N_{km} injury criterion was compared to that of the BioRID II model. For the comparison, the N_{km} injury criterion was normalised with respect to the 5% injury limit. Figures 5 and 6 show the set-up of the simulated sled tests 1 and 9, respectively, of Table I with the BioRID II model and the active human model.

In order to analyse the sensitivity of the selected injury criteria, rear-impact simulations were performed with the active human model with various known factors affecting the whiplash injury risk with each of the seat models: pulse severity, pulse direction, head restraint position, occupant position, seat characteristics, and neck muscle activation. In total, 30 simulations were performed, as summarised in Table III. Simulation 1 is further referred to as the reference simulation. Variations on this reference simulation are shown in bold in Table III.

For the rear-impact pulse severity the three acceleration pulses from the Euro NCAP rear-impact assessment test protocol were applied (see Fig. 3). A variation in direction of 10 degrees rotated among the vertical axis was applied to the high pulse.

Four different initial positions were simulated: position based on the BioRID II positioning in the Euro NCAP whiplash assessment sled tests (nominal); backward on a seat with 10 degree rotated seatback; forward bent position; and nominal with the head 30 degrees rotated around the vertical axis (yaw). The nominal position and three position variations are shown in Fig. 7.

Furthermore, the head restraint position was varied from its nominal position, as defined in the Euro NCAP rear-impact assessment test protocol, to the lowest and highest position possible for each seat. Also, the distance between head restraint and active human model head was varied by moving the head restraint in frontal direction 50 mm closer to the head, as well as in backward direction increasing the distance by 50 mm. Note that this variation is not possible with the hardware seats.

The characteristics of the recliner of the seat models were varied to investigate the effect of the amount of seatback rotation. The spring characteristic of the recliner was up- and down-scaled with a factor 1.5. Note that this is also not possible with the hardware seats.

The awareness settings of the active human model were varied by altering the co-contraction level of the neck muscles (bracing) and the reaction time. In the reference simulation the co-contraction level was 20% and reaction time was 0.1 s, while for the "aware" simulation the co-contraction was increased to 80% and the reaction time decreased to 0 s, i.e. occupant had time to react before the impact. Also, simulations were performed without any active muscle behaviour (passive). The used co-contraction levels and reaction times were based on the active human model validation study of Meijer *et al.* [43]-[44].

Fig. 5. Simulation set-up of rear-impact test 1: seat 1, with head restraint in nominal position and occupant in nominal position.

Fig. 6. Simulation set-up of rear impact test 9: seat 2, with head restraint in highest position and occupant in bent forward position.

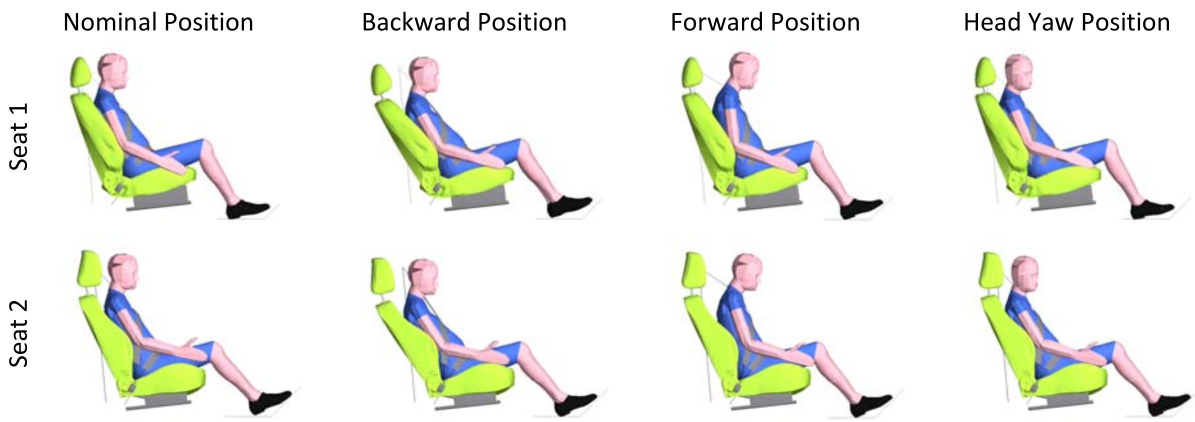


Fig. 7. Simulation set-ups with active human model on seat model 1 and 2 in nominal (simulation 1–6 and 10-17), backward (simulation 7), forward (simulation 8) and head yaw (simulation 9).

TABLE III
SIMULATION MATRIX: VARIATIONS ON THE REFERENCE SIMULATION (#1) ARE INDICATED BOLD

#	Pulse	Pulse direction	AHM Awareness	AHM Position	AHM Mass	AHM Gender	Head restraint position	Recliner position	Recliner stiffness
1	High	0°	Normal	Nominal	75 Kg	Male	Nominal	Nominal	Nominal
2	**Low**	0°	Normal	Nominal	75 Kg	Male	Nominal	Nominal	Nominal
3	**Mid**	0°	Normal	Nominal	75 Kg	Male	Nominal	Nominal	Nominal
4	High	**10°**	Normal	Nominal	75 Kg	Male	Nominal	Nominal	Nominal
5	High	0°	Normal	**Backward**	75 Kg	Male	Nominal	Nominal	Nominal
6	High	0°	Normal	**Forward**	75 Kg	Male	Nominal	**10° back**	Nominal
7	High	0°	Normal	**Head yaw**	75 Kg	Male	Nominal	Nominal	Nominal
8	High	0°	Normal	Nominal	75 Kg	Male	**High**	Nominal	Nominal
9	High	0°	Normal	Nominal	75 Kg	Male	**Low**	Nominal	Nominal
10	High	0°	Normal	Nominal	75 Kg	Male	**Front**	Nominal	Nominal
11	High	0°	Normal	Nominal	75 Kg	Male	**Rear**	Nominal	Nominal
12	High	0°	Normal	Nominal	75 Kg	Male	Nominal	Nominal	**Stiffer**
13	High	0°	Normal	Nominal	75 Kg	Male	Nominal	Nominal	**Less stiff**
14	High	0°	**Aware**	Nominal	75 Kg	Male	Nominal	Nominal	Nominal
15	High	0°	**Passive**	Nominal	75 Kg	Male	Nominal	Nominal	Nominal

III. RESULTS

Validation of the seat models

The rear-impact performance of the seat models was validated by simulating rear-impact sled tests and comparing the simulated kinematics and dynamics of the BioRID II model with the results of the tests. The acceleration and injury response time history signals of the BioRID II in rear-impact sled tests and corresponding simulations are presented in Appendix A and Appendix B, respectively. The resulting acceleration and injury response peak values of the simulations and sled tests are comparable for the performed tests and simulations with the head restraint in nominal and highest position, except the head x-acceleration and lower neck tension force. The head x-acceleration is generally higher in the sled tests compared to the simulations, while the lower neck tension force is generally lower in the sled tests compared to the simulations. In the tests and simulations with head restraint in lowest position, the acceleration outputs are similar, except the head x-acceleration of the BioRID II for seat model 1. The peak values of the upper and lower neck backwards shear and tension force, as well as $N_{km,}$ are generally higher in the sled test with seat model 1 compared to the simulations. On the other hand, peak injury responses are generally lower in the sled test with seat model 2 compared to the simulations.

BioRID II model versus Active Human Model

The rear-impact sled tests performed for the validation of the seat models were also simulated with the active human model. The acceleration and injury response time history signals of the active human model in the rear-impact simulations are presented in Appendix A and Appendix B, respectively. Table IV summarises the differences in peak N_{km} injury responses between the active human model and BioRID II model injury responses. The mean and range of the N_{km} injury responses from the active human model and BioRID II model simulations are similar.

TABLE IV

DIFFERENCES IN PEAK N_{KM} INJURY RESPONSES BETWEEN BIORID II MODEL AND ACTIVE HUMAN MODEL SIMULATIONS

Criterion	Active human model		BioRID II model		Mean difference with respect to 5% limit
	Mean	*Range*	*Mean*	*Range*	
N_{km} [-]	0.38	0.18–1.24	0.38	0.19–1.08	-0.01

Effect of various rear-impact scenarios

In total, 15 different rear-impact scenarios were simulated with each of the seat models. In Table V and Table VI the resulting whiplash injury risks are presented for all performed simulations with seat model 1 and 2, respectively. In the reference rear-impact simulations the whiplash injury risk is 44% and 28% for seat 1 and 2, respectively. The upper and lower neck tension forces are the dominant criteria for both seats, where the upper neck tension force results in the highest whiplash injury risk for seat 1, and the lower neck tension force for seat 2.

The simulations with the low and mid rear-impact pulses result in lower risk on whiplash injury compared to the simulations with high pulse. In particular, the lower and upper neck tension forces are lower. As in the simulation with the high pulse, for seat 2 the lower neck tension force is the dominant criterion, while for the seat 1 the NIC gives the highest risk of whiplash injury in the simulations with low and mid pulse. The simulations with the 10 degree rotated pulse results in similar whiplash injury risks, with only small differences observed for the injury criteria outputs.

Due to a more forward and backward initial position of the active human model, the whiplash injury risk increases. In particular, the upper neck tension forces increase compared to the reference simulation with nominal position, and are the dominant criteria for the backward simulations for both models. Furthermore, the NIC responses are increased and are the dominant criteria in the simulations with the bent forward initial position for both seats. For the seat 1 the upper and lower neck flexion moments are also increased by the bent forward position. The simulations with the initial head rotation (head yaw) result in lower whiplash injury risks for both seats, which is caused by decreased peak values for the upper and lower neck tension forces.

By increasing the head restraint height, the whiplash injury risk stays on a similar level, while lowering the head restraint position increases the whiplash injury risk. For both seats the upper neck tension force increases by lowering the head restraint position. For seat 2 this is also the dominant criteria. For seat 1 the lower neck flexion moment is the most dominant whiplash injury criteria in the simulation with the head restraint in lowest position. Note that the lower neck flexion moment does not increase by lowering the head restraint of seat 2. Changing the head restraint position in horizontal position affects the neck tension forces and NIC. In particular, the upper neck tension forces decrease by decreasing the initial head-to-head restraint distance. By increasing the head-to-head restraint distance, the NIC peak response increases and becomes the dominant whiplash injury criterion in the simulations with the seat 2.

By reducing and increasing the recliner stiffness, the seatback rotation is increased and decreased, respectively. This only slightly affects the whiplash injury risk for seat 1. However, by reducing the recliner stiffness of seat model 2 and with that increasing the seatback rotation, the NIC injury response increases and becomes the dominant whiplash injury criterion. Due to the increased NIC peak response, the whiplash injury risk increases from 28% to 40% by reducing the recliner stiffness of seat 2.

By increasing the awareness of the active human model, the whiplash injury risk is lowered due to decreased peak responses of the lower neck tension force, and to a lesser intent the upper neck tension force. For seat 2 the dominant criterion changes from the lower neck tension force to the upper neck tension force by increasing the awareness. On the other hand, decreasing the awareness by removing the active muscle behaviour of the active human model hardly affects the whiplash injury risk.

TABLE V

WHIPLASH INJURY RISK FOR VARIOUS REAR-IMPACT SIMULATIONS WITH SEAT 1 (INJURY CRITERION THAT GIVES HIGHEST INJURY RISK IS INDICATED BOLD FOR EACH SIMULATION)

Injury criteria	Reference*	Pulse			Human position			Head restraint position				Recliner		Awareness	
		Mid (0°)	*Low (0°)*	*High (10°)*	*Back-ward*	*For-ward*	*Head yaw*	*High*	*Low*	*Front*	*Rear*	*Stiffer*	*Less stiff*	*Aware*	*Passive*
Upper neck backward shear force	1%	2%	1%	1%	4%	5%	1%	2%	3%	3%	2%	1%	1%	2%	0%
Upper neck tension force	**44%**	4%	4%	**43%**	**71%**	60%	**28%**	**36%**	66%	20%	**57%**	**42%**	**35%**	**24%**	**43%**
Upper neck extension moment	2%	2%	2%	2%	4%	5%	2%	2%	2%	1%	2%	2%	1%	1%	2%
Upper neck flexion moment	16%	11%	4%	22%	55%	32%	11%	10%	60%	14%	5%	19%	15%	4%	3%
Lower neck backward shear force	4%	3%	2%	3%	4%	3%	3%	5%	2%	3%	14%	5%	3%	1%	4%
Lower neck tension force	35%	10%	4%	32%	49%	35%	23%	30%	42%	**27%**	37%	33%	30%	0%	34%
Lower neck extension moment	0%	0%	0%	0%	0%	1%	0%	0%	0%	0%	0%	0%	0%	1%	0%
Lower neck flexion moment	4%	2%	2%	16%	27%	58%	4%	4%	**76%**	3%	8%	3%	5%	4%	3%
NIC	5%	**12%**	**5%**	5%	45%	**80%**	5%	10%	11%	4%	30%	10%	5%	3%	5%
N_{km}	3%	3%	2%	10%	41%	22%	3%	3%	54%	3%	3%	3%	4%	2%	2%
Whiplash	**44%**	**12%**	**5%**	**43%**	**71%**	**80%**	**28%**	**36%**	**76%**	**27%**	**57%**	**42%**	**35%**	**24%**	**43%**

* REFERENCE SIMULATION: HIGH PULSE, NOMINAL INITIAL POSITION, NOMINAL HEAD RESTRAINT POSITION, VALIDATED RECLINER STIFFNESS, STANDARD AWARENESS (20% CO-CONTRACTION, 0.1S REACTION TIME)

TABLE VI

WHIPLASH INJURY RISK FOR VARIOUS REAR-IMPACT SIMULATIONS WITH SEAT 2 (INJURY CRITERION THAT GIVES HIGHEST INJURY RISK IS INDICATED BOLD FOR EACH SIMULATION)

Injury criteria	Reference*	Pulse			Human position			Head restraint position				Recliner		Awareness	
		Mid (0°)	*Low (0°)*	*High (10°)*	*Back-ward*	*For-ward*	*Head yaw*	*High*	*Low*	*Front*	*Rear*	*Stiffer*	*Less stiff*	*Aware*	*Passive*
Upper neck backward shear force	2%	3%	1%	2%	4%	37%	1%	2%	2%	2%	4%	2%	4%	3%	2%
Upper neck tension force	20%	4%	3%	28%	**61%**	58%	18%	16%	**34%**	4%	39%	24%	33%	**12%**	28%
Upper neck extension moment	1%	2%	1%	1%	2%	39%	2%	1%	1%	1%	2%	2%	2%	2%	1%
Upper neck flexion moment	10%	2%	2%	12%	3%	14%	5%	11%	12%	5%	26%	8%	9%	5%	3%
Lower neck backward shear force	2%	3%	2%	1%	3%	3%	1%	3%	0%	0%	4%	3%	2%	0%	2%
Lower neck tension force	**28%**	**15%**	**9%**	**31%**	43%	38%	**24%**	**26%**	32%	**23%**	32%	**28%**	33%	0%	**32%**
Lower neck extension moment	1%	1%	0%	0%	0%	2%	1%	0%	1%	1%	0%	0%	1%	1%	0%
Lower neck flexion moment	5%	2%	1%	5%	4%	4%	4%	4%	2%	9%	3%	4%	3%	2%	2%
NIC	7%	8%	4%	7%	41%	**≥95%**	7%	18%	19%	5%	**52%**	12%	**40%**	9%	12%
N_{km}	3%	3%	2%	4%	3%	27%	3%	3%	4%	3%	5%	3%	3%	3%	3%
Whiplash	**28%**	**15%**	**9%**	**31%**	**61%**	**≥95%**	**24%**	**26%**	**34%**	**23%**	**52%**	**28%**	**40%**	**12%**	**32%**

* REFERENCE SIMULATION: HIGH PULSE, NOMINAL INITIAL POSITION, NOMINAL HEAD RESTRAINT POSITION, VALIDATED RECLINER STIFFNESS, STANDARD AWARENESS (20% CO-CONTRACTION, 0.1S REACTION TIME)

IV. DISCUSSION

The two passenger seat models were validated by comparing the responses of a BioRID II model with a BioRID II in various rear-impacts. Overall the responses of the BioRID II model compared to the BioRID in the tests were similar, especially for the tests with the head restraint in nominal and high position. With the head restraint in lowest position, the peak injury responses of the upper and lower neck backwards shear and tension force, as well as the $N_{km,}$ were smaller in the simulation with seat model 1 than in the sled test. On the other hand, for seat model 2 with head restraint in lowest position, these peak injury responses were higher in the simulation than in the sled test.

The upper and lower neck shear, tension and bending loads as well as NIC were related to whiplash injury by analysing volunteer tests and accident reconstructions with a numerical human model [23]. Therefore, it is plausible to use these limits for the active human model, which is well validated against rear-impact tests with volunteers as well as PMHS [44]. However, the N_{km} criterion was linked to whiplash injury by accident reconstructions with the BioRID II model [15]. As rear-impact simulations with the BioRID II model and active human model showed similar N_{km} peak responses for both models, the N_{km} could be used as the whiplash injury predictor for the active human model. While non-linear injury risk curves, which are commonly used, were developed for all adopted whiplash injury criteria [15][23], the whiplash injury risk was investigated by using linear interpolation between the 5% and 95% limits of the whiplash injury criteria. This method was used as some of the adopted non-linear injury risk curves predicted a significant risk on injury for zero loads. Therefore, for these injury risk curves the 5% risk limits were based on non-injured volunteer tests [23]. Another limitation regarding the whiplash injury risk estimation was the relative wide confidence intervals of the injury risk curves from which the limits were determined.

Furthermore, various known factors affecting the whiplash injury risk were investigated. The developed simulation method was sensitive for different pulse severities. Lowering the pulse severity not only reduced the whiplash risk, it also changed the dominant injury criterion for the seat 1 simulations. The higher NIC peak responses for the mid severity pulse can be explained by the higher peak accelerations of this pulse. On the other hand, the simulation method was not sensitive for a pulse direction variation. All selected injury criteria, which are all measurements in the sagittal plane, showed similar responses, while the pulse direction was varied. However, peak injury responses in the lateral direction, which were not taken into account in the injury criteria, were increased. This also explains the decreased whiplash injury risk prediction in simulations with initial head yaw. In reality, a higher risk of whiplash injury is expected with rotated head or a slightly different angle of the pulse direction, since this will induce additional loads (in another direction) on the (cervical) spine.

A more forward as well as a more backward initial position of the active human model affected the calculated risk of a whiplash injury. A forward and backward initial position increased several injury responses. The NIC injury response was raised due to an increased head-to-head restraint distance, which increased the maximum acceleration and velocity of the head. Also, the tension forces and bending moments increased as the active human model upwards movement increased, which resulted in more rotation of the head and neck around the head restraint. Note that the BioRID II and active human model have not been validated for out-of-position in rear-impacts. However, the physical properties of both BioRID II and active human model are well validated and the active human model is extensively validated for various loading directions. Another limitation concerning the out-of-position simulations is that the spinal shape of the active human model has been designed for nominal car driving position [41]. Living humans attain a well-balanced and comfortable spinal alignment, which is not fully included in the out-of-position simulations. This could have affected the results of simulations with initial backwards position.

The effect of the head restraint position and recliner stiffness on the risk of a whiplash injury was also shown with the performed simulations. The head restraint height affected the amount of head and neck rotation around the head restraint, resulting in higher tension forces and bending moments in the neck for lower head restraint positions. Different head-to-head restraint distances affected the maximum head acceleration and velocity before the head to head restraint impact, resulting in higher NIC and neck tension forces by increasing the head-to-head restraint distance.

Furthermore, it was believed that active musculature behaviour can affect the whiplash injury outcome, which was confirmed by simulations with a more aware active human model. The braced musculature behaviour in the neck reduced the tension forces, which can be explained by higher initial compression forces in the neck caused by the muscle contraction. This was not only observed for the simulation with aware active human model but also for the simulated scenarios with initial head yaw. The additional neck muscle forces due to the non-nominal head position caused higher initial compression forces in the neck. While the aware behaviour caused lower whiplash injury risk, the scenarios with passive human behaviour did not show differences in whiplash risk compared to the simulations with nominal active human model settings. An explanation could be the reaction time in the nominal simulation of 100 ms.

The performed rear-impact simulations with the active human model showed that different injury criteria predicted the highest whiplash injury risk. The upper and lower neck tension force were the dominant whiplash injury predictor in most of the performed simulations. Also, the lower neck flexion moment and NIC predicted the highest whiplash injury risk in at least one simulated rear-impact scenario. This implies that several of the presented whiplash injury mechanism hypotheses could be valid, as the dominant whiplash injury predictors were found in the lower and upper neck, as well as different dominant loading mechanisms: tension forces, flexion moments or NIC. The upper and lower neck backwards shear forces and extension moments were not dominant in the rear-impact simulations and predicted relative low whiplash injury risks for most simulated scenarios. However, relative low whiplash injury risk estimations by, for instance, the upper neck backwards shear force, does not mean that there is no correlation between the criterion and whiplash injury, which was shown by Davidsson *et al.* [45].

There are strong indications that chronic neck pain is caused by psychological muscle reaction rather than the initial injury [4]. Therefore, a muscle criterion, like muscle elongation or elongation rate, could explain whiplash injuries caused by low severity pulses. The presented simulation method with the active human model makes it possible to add an injury predictor on musculature level. Also, muscle criteria could explain additional whiplash injury risk in rear-impact scenarios with lateral components.

The performed rear-impact simulations showed generally better performance of seat 2 compared to seat 1, which is in line with the Euro NCAP whiplash score (fair and poor rated, respectively). It has been shown that Euro NCAP whiplash score generally correlates with real world injury risk [46], which was supported by the results of this study. However, for some scenarios, as low pulse or bent forward initial position, the relative poor rated seat showed better performance than the fair rated seat.

Additional validated seat models which can be used in the developed simulation method of this study to reconstruct rear-impact accidents with and without whiplash outcome from an accident database, are needed for validation of this simulation method. However, this model and the way the whiplash injury risk was calculated showed similar effects (decreasing or increasing) on the risk of whiplash as found in literature. Therefore, the authors are confident that this simulation method is a good approach and could already be used in an early design process to further optimise seat designs and anti-whiplash concepts.

V. CONCLUSION

In this study a simulation method was developed for whiplash injury prediction that takes into account injury criteria covering several whiplash injury mechanisms by using an active human model. In order to test the sensitivity of the developed simulation method rear-impact simulations were performed with the active human model in conditions where parameters known to affect the whiplash injury risk were varied. The simulation method showed similar effects on the risk of whiplash as known factors found in literature. Therefore, it is concluded that this simulation method makes it possible to further optimise whiplash assessment procedures and passenger seat designs for a more robust protection in the large range of real world rear-impact accident scenarios. It is recommended to further verify the method developed here by simulating many real world rear-impact accidents from accident databases and developing additional whiplash injury predictors that could explain additional whiplash injury risk in rear-impact scenarios with lateral components.

VI. ACKNOWLEDGEMENT

The authors thank Paul van Hooijdonk of TASS International BV for the development of the seat models.

VII. REFERENCES

[1] Linder, A. *et al.* (2013) ADSEAT–Adaptive seat to reduce neck injuries for female and male occupants. *Accident Analysis & Prevention,* **60**: pp.334-343.

[2] Kullgren, A. *et al.* The effect of whiplash protection systems in real-life crashes and their correlation to consumer crash test programmes. *Proceedings of the 20th International Technical Conference on the Enhanced Safety of Vehicles (ESV)*, 2007, Lyon, France.

[3] Insurance Europe. "European motor insurance markets, November 2015" Internet: http://www.insuranceeurope.eu/european-motor-insurance-markets, Date updated 03-06-2016.

[4] Schmitt, M. A. (2010) Understanding functioning and health in patients with whiplash-associated disorders. *Utrecht University*, The Netherlands.

[5] Scholten-Peeters G. G., Verhagen A. P. *et al.* (2003) Prognostic factors of whiplash associated disorders: a systematic review of prospective cohort studies. *Pain,* **104**(1): pp.303-322.

[6] Spitzer, W. O., Skovron, M. L. *et al*. (1995) Scientific monograph of the Quebec Task Force on Whiplash-Associated Disorders: redefining "whiplash" and its management. *Spine*, **20**(8 Suppl.): pp.1S–73S.

[7] Aldman, B. An analytical approach to the impact biomechanics of head and neck. *30th Annual Proceedings Association for the Advancement of Automotive Medicine*, 1986, Barrington: pp.439–54.

[8] Svensson, M. Y., Örtengren, T., Aldman, B., Lövsund, P., Seeman, T. (1989) A Theoretical Model for and a Pilot Study Regarding Transient Pressure Changes in the Spinal Canal under Whip-lash Motion. *Chalmers University of Technology*, Göteborg, Sweden.

[9] Svensson, M. Y., Aldman, B. *et al*. Pressure Effects in the Spinal Canal during Whiplash Extension Motion: A Possible Cause of Injury to the Cervical Spinal Ganglia. *Proceedings of IRCOBI Conference*, 1993, Eindhoven, The Netherlands, pp.189–200.

[10] Boström, O., Svensson, M. Y. *et al*. A new neck injury criterion candidate - based on injury findings in the cervical spinal ganglia after experimental neck extension trauma. *Proceedings of IRCOBI Conference*, 1996, Dublin, Ireland, pp.123–36.

[11] Boström, O., Håland, Y. *et al.* (2000) Neck Injury Criterion (NIC) and its Relevance to Various Possible Neck Injury Mechanisms. *Frontiers in Whiplash Trauma*, IOS Press, Amsterdam, The Netherlands.

[12] Yang, K. H., Begeman, P. C., Muser, M. H., Niederer, P., Walz, F. On the role of cervical facet joints in rear end impact neck injury mechanism. *SAE Technical Paper*, 1997, pp.127–9.

[13] Deng, B., Luan, F. *et al*. (2000) Testing shear hypothesis of whiplash injury using experimental and analytical approaches. *Frontiers in Whiplash Trauma*, IOS Press, Amsterdam, The Netherlands.

[14] Schmitt, K-U, Muser, M. H., Niederer, P. A new neck injury criterion candidate for rear-end collisions taking into account shear forces and bending moments. *Proceedings of the 17th International Technical Conference on the Enhanced Safety of Vehicles*, 2001, Amsterdam, The Netherlands, Paper No. 124.

[15] Kullgren, A., Eriksson, L., Boström, O., Krafft, M. Validation of neck injury criteria using reconstructed real-life rear-end crashes with recorded crash pulses. *Proceedings of the 18th International Technical Conference on the Enhanced Safety of Vehicles*, 2003, Nagoya, Japan, Paper No. 344.

[16] Yoganandan, N., Pintar, F. A., Kleinberger, M. (1998) Cervical spine vertebral and facet joint kinematics under Whiplash. *Journal of biomechanical engineering*, **120**(2): pp.305–7.

[17] Yoganandan, N., Pintar, F. A. *et al*. (2000) Biomechanics of human occupants in simulated rear crashes: documentation of injuries, determination of forces and moments at the occipital condyles, and comparison of injury metrics with proposed injury criteria. *Stapp Car Crash Journal*, **4**: pp.189–203.

[18] Yoganandan, N., Pintar, F. A. Single rear impact produces lower cervical spine soft tissue injuries. *Proceedings of IRCOBI Conference*, 2001, Isle of Man, UK, pp.201–11.

[19] Stemper, B. D., Yoganandan, N., Pintar, F. A. (2002) Intervertebral rotations as a function of rear impact loading. *Biomedical sciences instrumentation*, **38**: pp.227–31.

[20] Stemper, B. D. *et al.* Toward a lower neck rear impact injury criterion – correlation of lower neck loads with spinal kinematics. *Proceedings of IRCOBI Conference*, 2007, Maastricht, The Netherlands, pp.249–60.

[21] Ono, K., Kaneoka, K., Wittek, A., Kajzer, J. Cervical Injury Mechanism Based on the Analysis of Human Cervical Vertebral Motion and Head-Neck-Torso Kinematics During Low Speed Rear Impacts. *Proceedings of the 41st Stapp Car Crash Conference,* 1997, Lake Buena Vista , Florida, USA, pp.339–56.

[22] Ono, K., Ejima, S. *et al*. Prediction of Neck Injury Risk Based on the Analysis of Localized Cervical Vertebral Motion of Human Volunteers during Low-Speed Rear Impacts. *Proceedings of IRCOBI Conference*, 2006, Madrid, Spain, pp.103–14.

[23] Ono, K., Ejima, S. *et al*. Evaluation Criteria for the Reduction of Minor Neck Injuries during Rear-end Impacts Based on Human Volunteer Experiments and Accident Reconstruction Using Human FE Model Simulations. *Proceedings of IRCOBI Conference*, 2009, York, United Kingdom, pp.381–98.

[24] Carlsson, A., Linder, A. *et al*. (2001) Dynamic kinematic responses of female volunteers in rear impacts and

comparison to previous male volunteer tests. *Traffic Injury Prediction*, **12**(4): pp.347–57.

[25] Krafft, M., Kullgren, A. *et al.* Influence of crash severity on various whiplash injury symptoms: A study based on real-life rear-end crashes with recorded crash pulses. *Proceedings of the 19th International Technical Conference on the Enhanced Safety of Vehicles*, 2005, Washington D.C., USA, Paper No. 05-0363.

[26] Mordaka, J. K. (2004) Finite element analysis of whiplash injury for women. *Nottingham Trent University*, Nottingham, United Kingdom.

[27] Stemper, B. D., Yoganandan, N., Pintar, F. A. (2004) Gender- and region-dependent local facet joint kinematics in rear impact: implications in whiplash injury. *Spine*, **29**, pp. 1764–71.

[28] Radanov, B. P., Sturzenegger, M., Di Stefano, G. (1995) Long-term outcome after whiplash injury. A 2-year follow-up considering features of injury mechanism and somatic, radiologic, and psychosocial findings. *Medicine*, **74**(5): pp.281–97.

[29] Sturzenegger, M., DiStefano, G., Radanov, B. P., Schnidrig, A. (1994) Presenting symptoms and signs after whiplash injury: the influence of accident mechanisms. *Neurology*, **44**(4): pp.688–93.

[30] Zuby, D. S., Avery, M. BioRID responses and the effect of crash pulse characteristics. *Proceedings of the International Conference on Vehicle Safety*, 2002, London, UK.

[31] Linder, A., Olsson, T. *et al*. Dynamics of different seat designs for low to medium velocity rear impact. *Proceedings of the 45th Annual Conference of the Association for the Advancement of Automotive Medicine*, 2001, Barrington, IL, 187-201.

[32] Kleinberger, M., Sun, E., Saunders, J., Zhou, Z. Effects of head restraint position on neck injury in rear impact. WAD'99 Compendium/Traffic Safety and Auto Engineering. *World Congress on Whiplash Associated Disorders*, 1999, Vancouver, Canada.

[33] Kleinberger, M. (2000) Importance of Head Restraint Position on Whiplash Injury. *Frontiers in Whiplash Trauma*, IOS Press, Amsterdam, The Netherlands

[34]Viano, D. C., Gargan, M. F. (1996) Seating Position and Head Restraint Location During Normal Driving: Implications to Neck Injury Risks in Rear Crashes. *Accident Analysis and Prevention*, **28**(6): pp.665–74.

[35]Tropiano, P., Thollon L. *et al.* (2004) Using a Finite Element Model to Evaluate Human Injuries Application to the HUMOS Model in Whiplash Situation. *Spine*, **29**(16): pp.1709-1716.

[36]Esat, V., Acar M. A multi-body model of the whole human spine for whiplash investigations. *Proceedings of the 23rd International Technical Conference on the Enhanced Safety of Vehicles (ESV)*, 2007, Lyon, France.

[37]Kobayashi, S. Kitagawa, Y. Development of a finite element model of the human neck for whiplash simulation. *Proceedings of the 18th International Technical Conference on the Enhanced Safety of Vehicles (ESV)*, 2003, Nagoya, Japan.

[38]European new car assessment programme (Euro NCAP). Assessment protocol – Adult occupant protection. Version 5.0, April 2009.

[39] MADYMO Theory Manual, Version 7.4.1, *TASS B.V.*, Rijswijk, The Netherlands, 2012.

[40]European new car assessment programme (Euro NCAP). The dynamic assessment of car seats for neck injury protection testing protocol. Version 3.2, November 2014.

[41] MADYMO Human Body Models Manual, Version 7.4.1., *TASS B.V.*, Rijswijk, The Netherlands, 2012.

[42] Horst, M. (2002) Human Head Neck Response in Frontal, Lateral and Rear End Impact Loading - Modelling and Validation. PhD Thesis, University of Technology Eindhoven, The Netherlands.

[43] Meijer, R., van Hassel, E. *et al*. Development of a multi-body human model that predicts active and passive human behaviour. *Proceedings of IRCOBI Conference*, 2012, Dublin, Ireland.

[44] Meijer, R., Elrofai, H., Broos, J., van Hassel, E. Evaluation of an active multi-body human model for braking and frontal crash events. *Proceedings of the 23rd International Technical Conference on the Enhanced Safety of Vehicles (ESV)*, 2013, Seoul, South Korea.

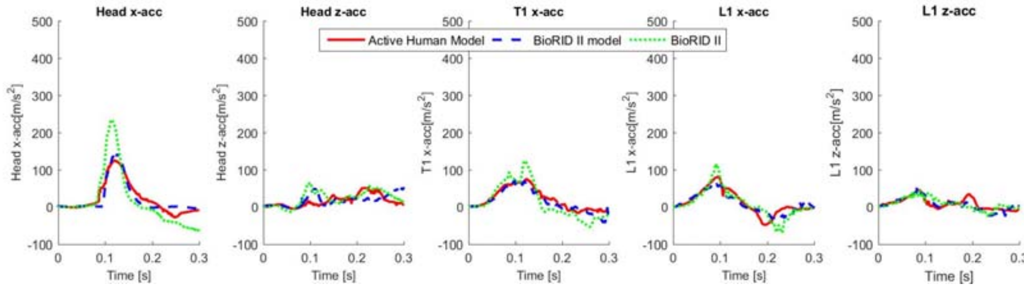
[45]Davidsson, J. *et al.* Evaluation of seat performance criteria for rear-end impact testing. *Proceedings of the 22nd International Technical Conference on the Enhanced Safety of Vehicles (ESV)*, 2011, Washington DC, USA.

[46]Kullgren, A., Fildes, B. *et al.* Evaluation of the Euro NCAP Whiplash Protocol Using Real-World Crash Data. *Proceedings of the 24nd International Technical Conference on the Enhanced Safety of Vehicles (ESV),* 2015, Gothenburg, Sweden.

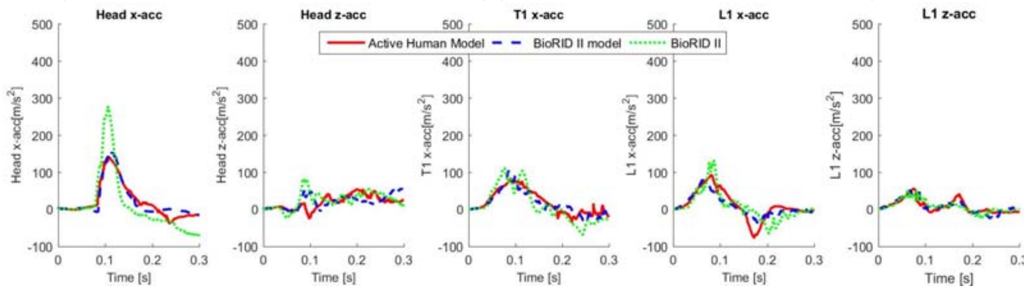
VIII. APPENDICES

Appendix A

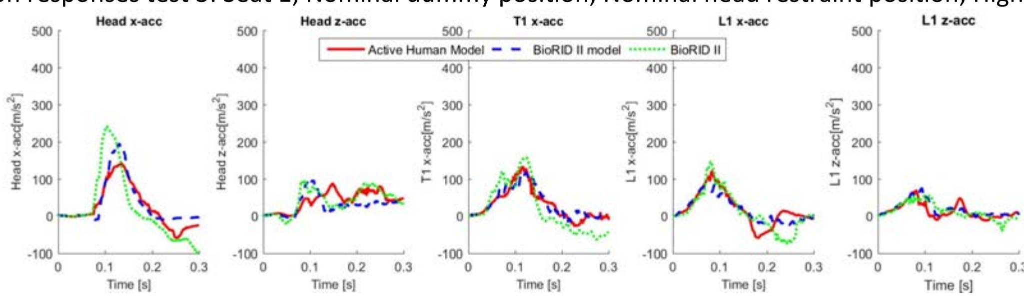
Acceleration responses test 1: Seat 1; Nominal dummy position; Nominal head restraint position; Low pulse.



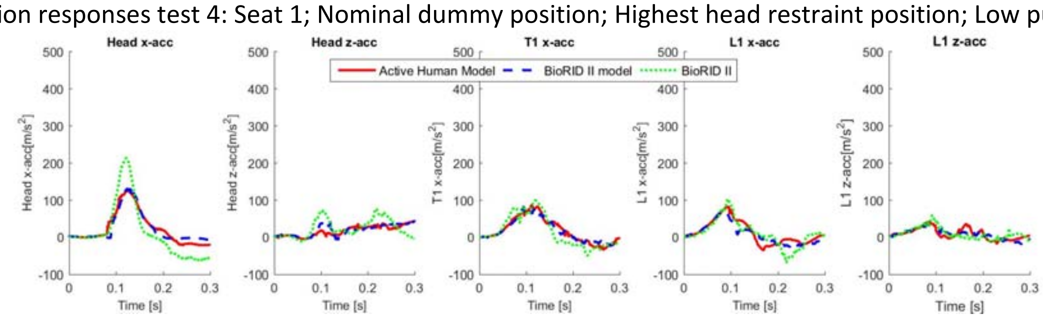
Acceleration responses test 2: Seat 1; Nominal dummy position; Nominal head restraint position; Mid pulse.



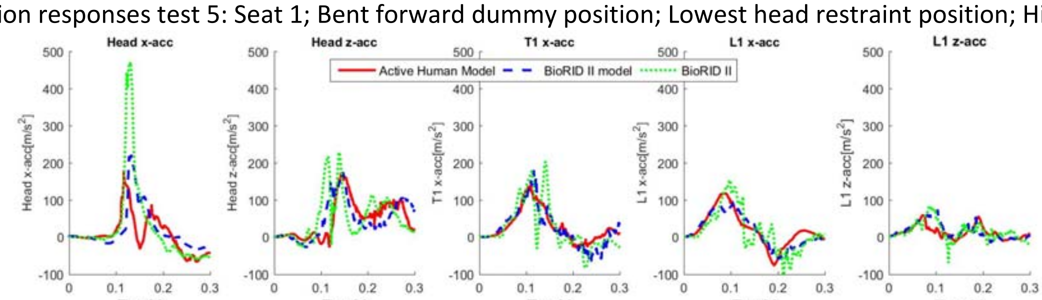
Acceleration responses test 3: Seat 1; Nominal dummy position; Nominal head restraint position; High pulse.



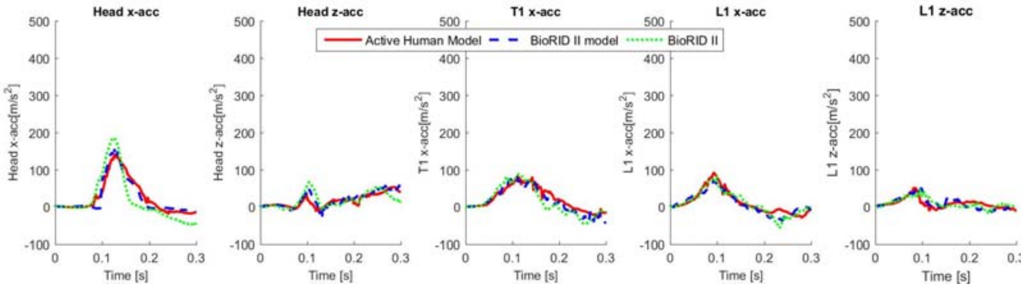
Acceleration responses test 4: Seat 1; Nominal dummy position; Highest head restraint position; Low pulse.



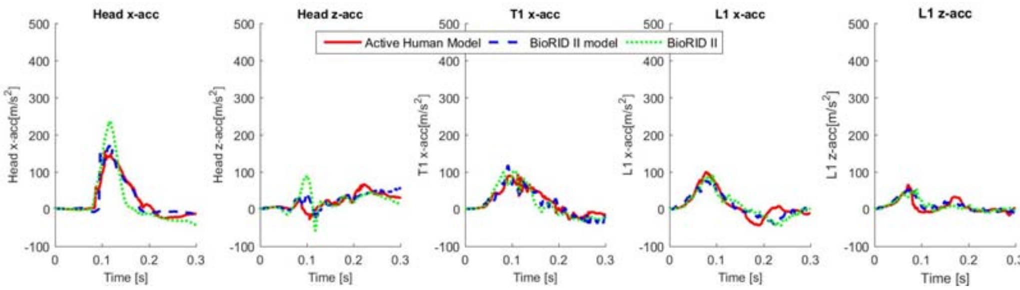
Acceleration responses test 5: Seat 1; Bent forward dummy position; Lowest head restraint position; High pulse.



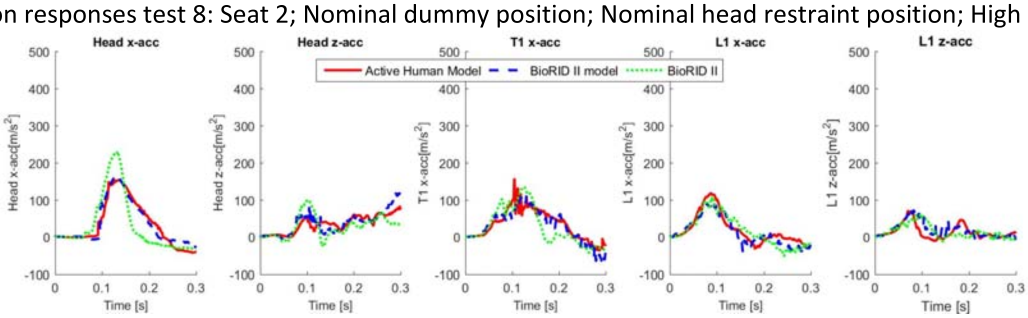
Acceleration responses test 6: Seat 2; Nominal dummy position; Nominal head restraint position; Low pulse.



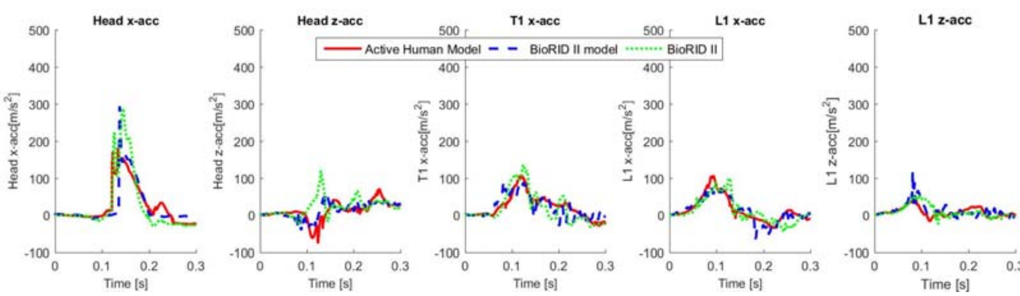
Acceleration responses test 7: Seat 2; Nominal dummy position; Nominal head restraint position; Mid pulse.



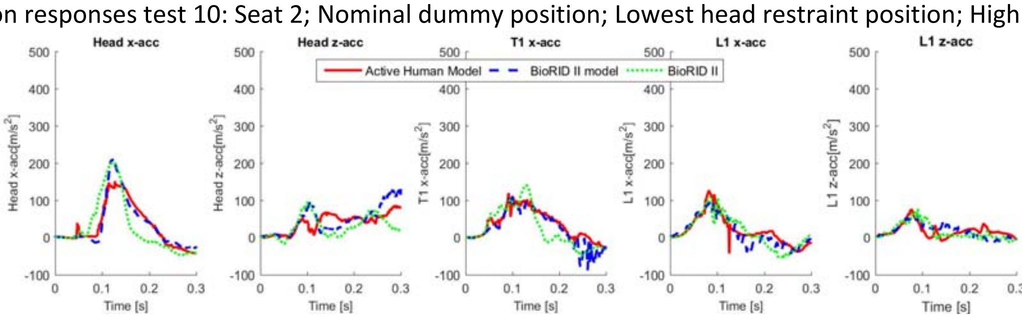
Acceleration responses test 8: Seat 2; Nominal dummy position; Nominal head restraint position; High pulse.



Acceleration responses test 9: Seat 2; Bent Forward dummy position; Highest head restraint position; Low pulse.

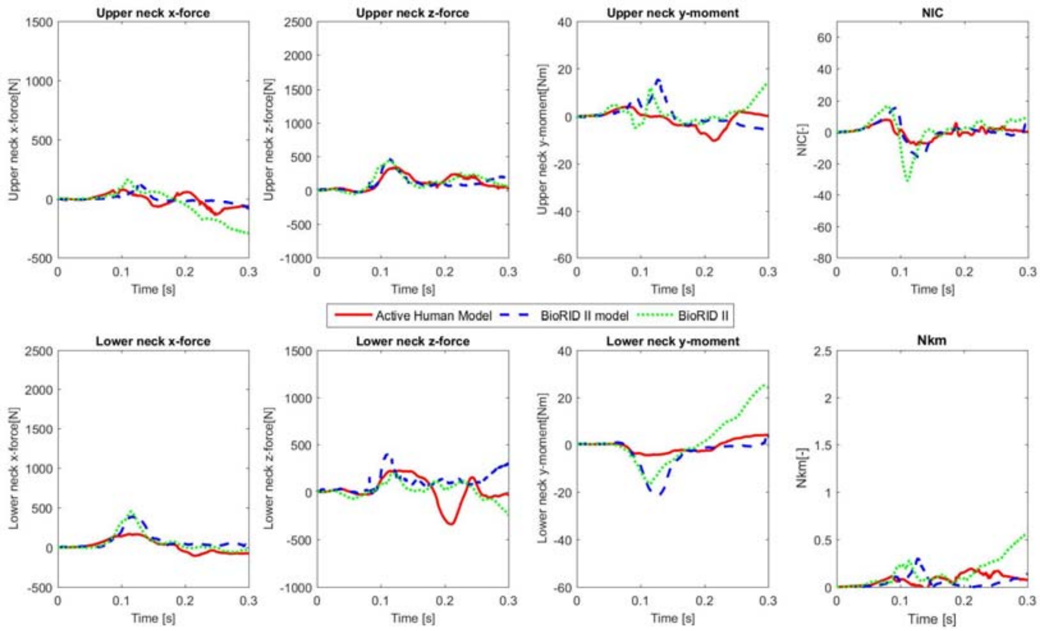


Acceleration responses test 10: Seat 2; Nominal dummy position; Lowest head restraint position; High pulse.

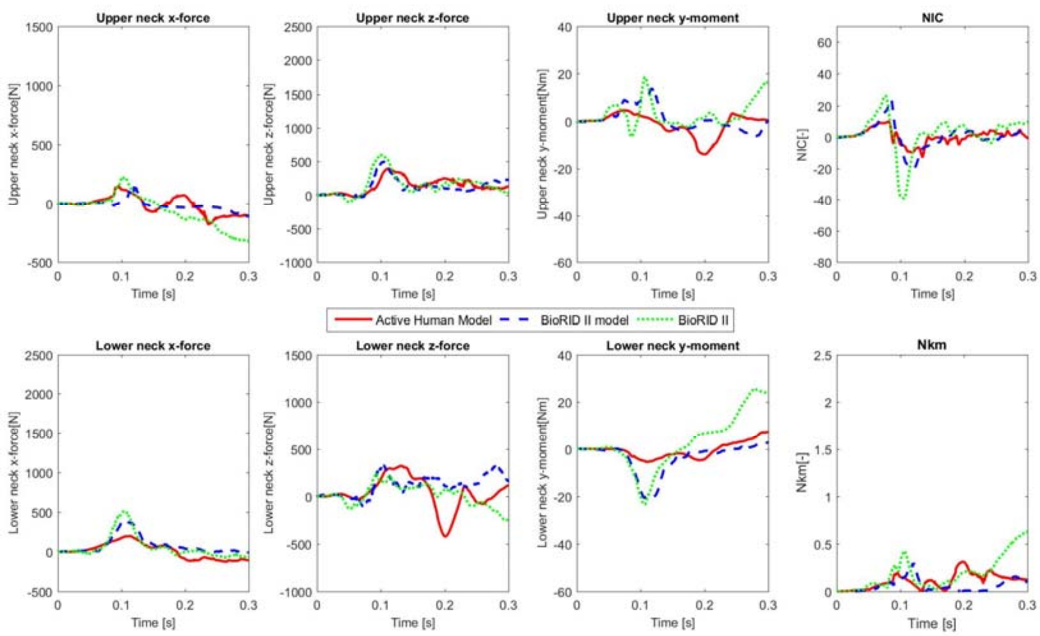


Appendix B

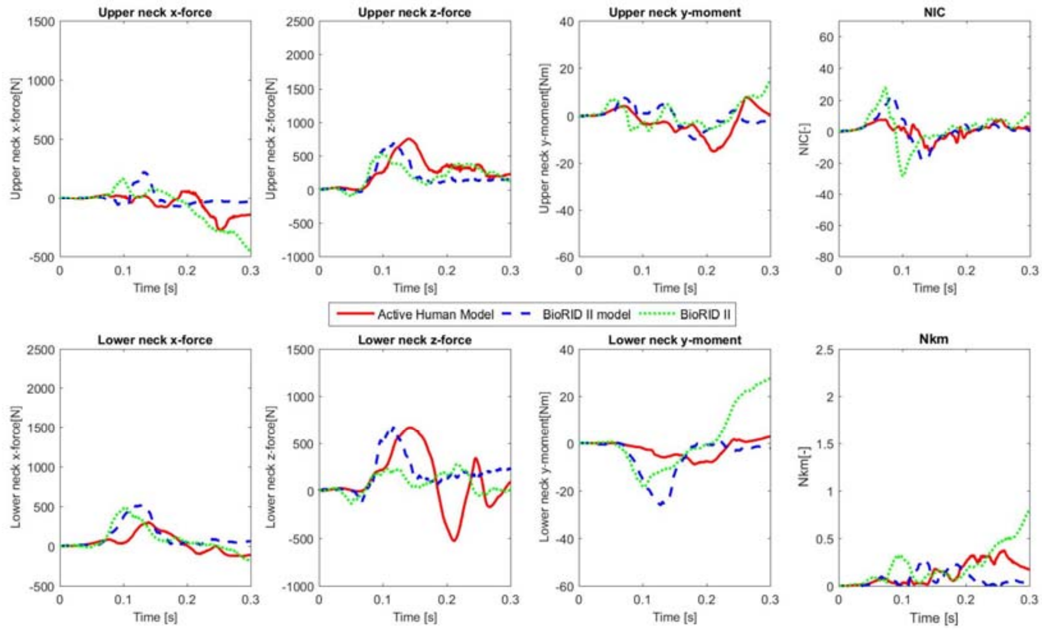
Injury responses test 1: Seat 1; Nominal dummy position; Nominal head restraint position; Low pulse.



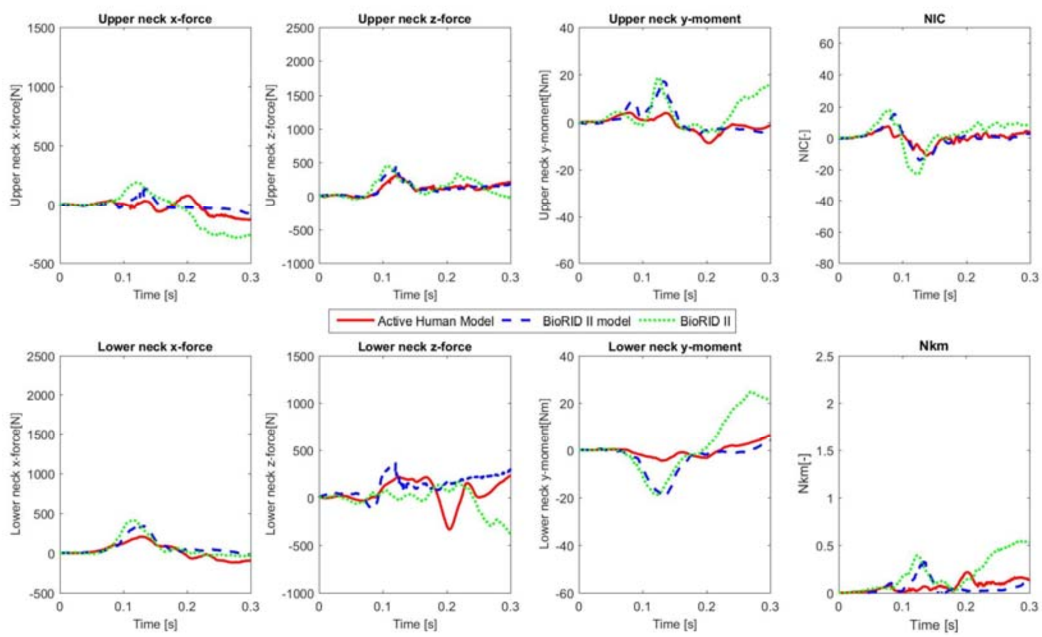
Injury responses test 2: Seat 1; Nominal dummy position; Nominal head restraint position; Mid pulse.



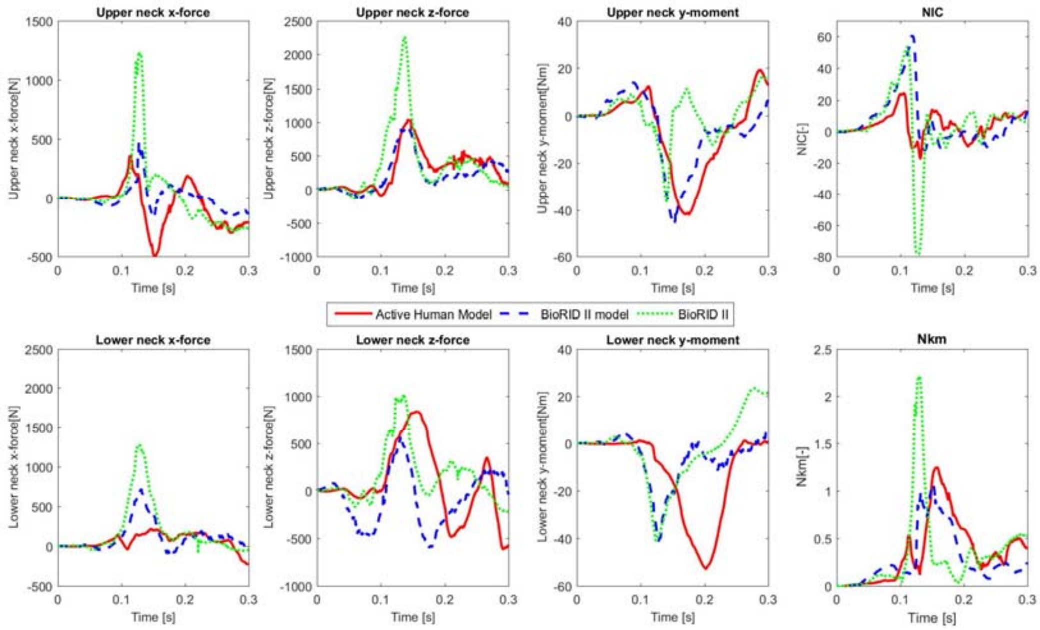
Injury responses test 3: Seat 1; Nominal dummy position; Nominal head restraint position; High pulse.



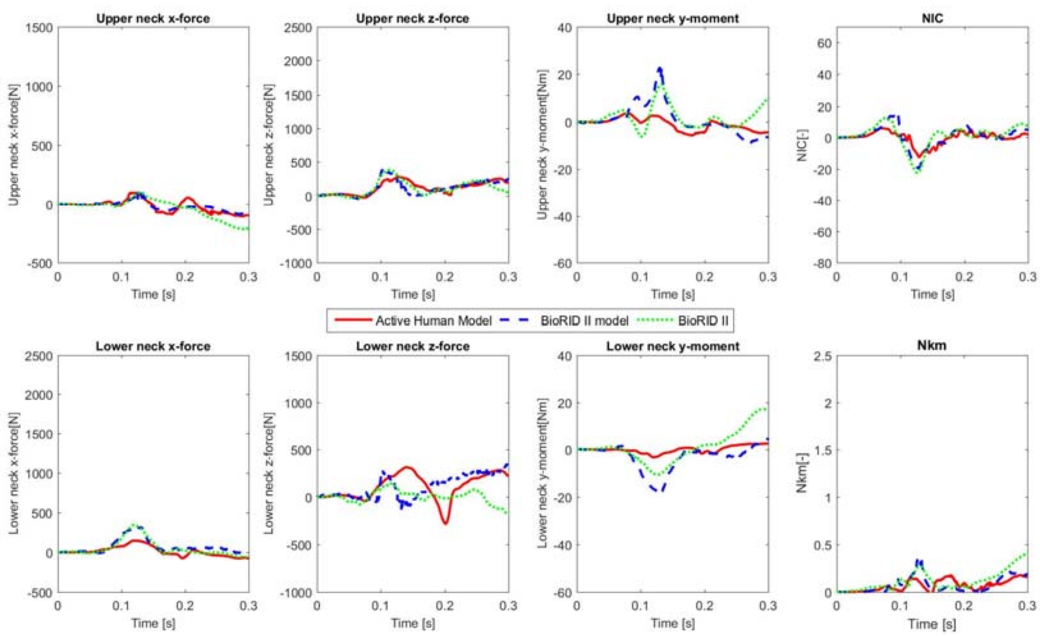
Injury responses test 4: Seat 1; Nominal dummy position; Highest head restraint position; Low pulse.



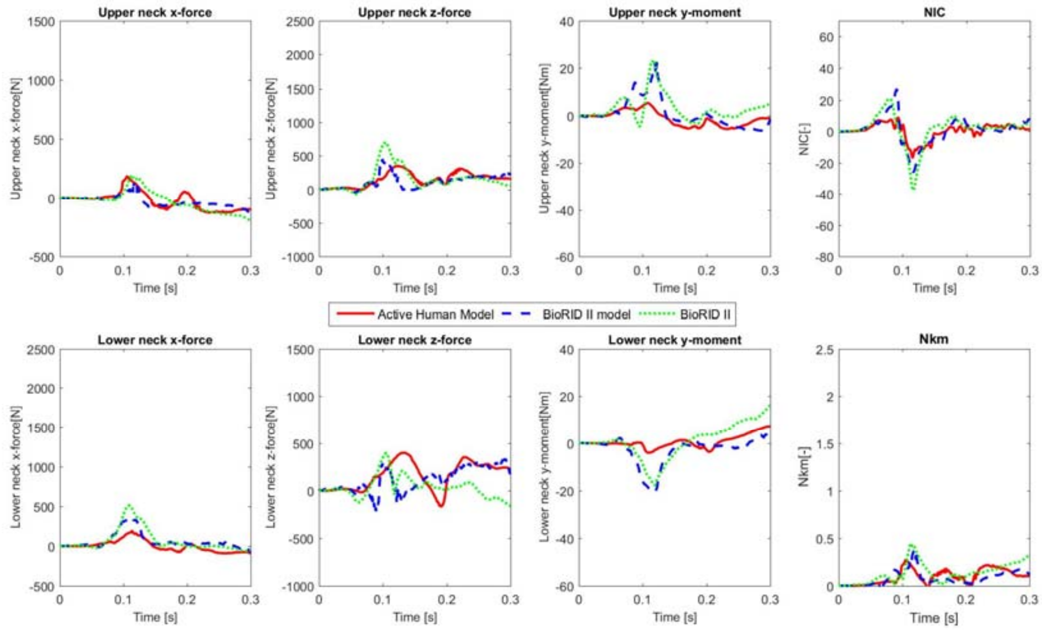
Injury responses test 5: Seat 1; Bent forward dummy position; Lowest head restraint position; High pulse.



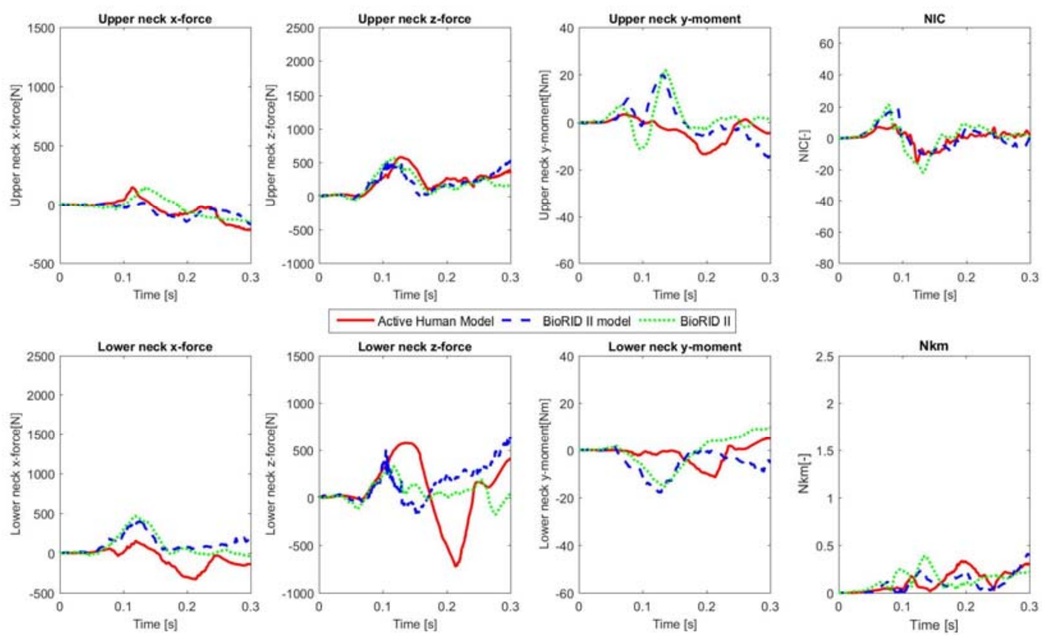
Injury responses test 6: Seat 2; Nominal dummy position; Nominal head restraint position; Low pulse.



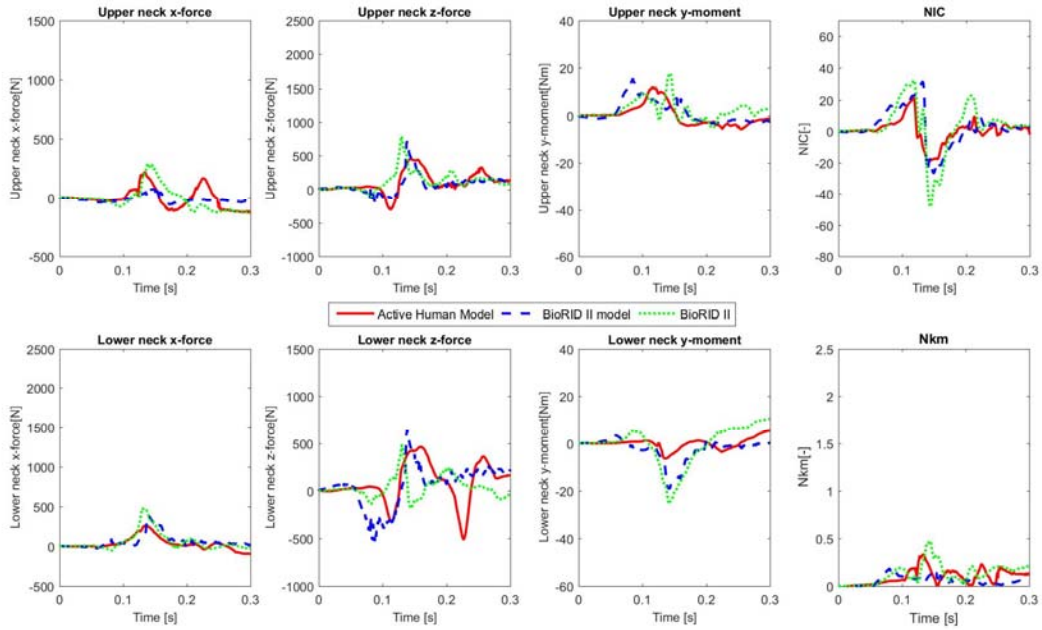
Injury responses test 7: Seat 2; Nominal dummy position; Nominal head restraint position; Mid pulse.



Injury responses test 8: Seat 2; Nominal dummy position; Nominal head restraint position; High pulse.



Injury responses test 9: Seat 2; Bent Forward dummy position; Highest head restraint position; Low pulse.



Injury responses test 10: Seat 2; Nominal dummy position; Lowest head restraint position; High pulse.