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Side Impact Neck Injury Criteria and Tolerances in Aerospace Safety

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ABSTRACT

The objective of this research is to develop neck injury criteria and injury tolerance levels to be used as a basis for a performance standard for the certification of side facing aircraft seats. This paper presents the results of the first part of this study. A literature review was performed on the neck injuries, kinematics and injury mechanisms in lateral loading and existing injury criteria and tolerance levels. Simulations with a human model were performed to determine the head and neck kinematics of the human body in a side facing aircraft seat and to determine the effect of belts and muscle activity.

Injury criteria that seemed to best correlate with neck injuries were the head angle and the head angular acceleration. A few studies also indicated that the upper neck lateral bending moment and upper neck lateral shear force could be related to neck injury. The simulations in which the human model was seated on a rigid side facing couch against a rigid wall at counter side of the impact direction and using a 5-point belt showed the largest head lateral angle and head lateral angular rotation. This 'worst case scenario' will be used in future PMHS tests for injury investigation. The phases in the kinematics during lateral loading described in literature coincide with the human model kinematics. The peak head lateral angle and head lateral angular acceleration resulting from the simulations coincide with the literature data. The simulations showed that muscle activity significantly reduces the head lateral angle and head lateral angular acceleration. The simulations showed that at a 16 G triangular impact a neck injury of AIS 2 or more can be expected.

INTRODUCTION

With the increasing use of side facing seats in small business aircraft, the FAA requires that specific procedures for the certification of side facing seats are included in the Federal Aviation Regulations (FAR). Since the resistance of the human body to lateral loading differs from that in frontal loading, the current regulations for forward facing seats cannot be directly adopted for side facing seats. Unfortunately, no information is currently known about lateral loading of the human body in aircraft crashes. The injury criteria and tolerance levels for forward facing aircraft seats were adopted from automotive crash safety regulations. However, the situation for an occupant in a side facing seat during a plane crash differs considerably from that in a side impact car crash (Lankarani, 1999, Soltis *et al.*, 2001). Therefore, injury criteria developed for automotive side impact cannot be directly adopted in regulations for side facing seats in aircraft.

Two earlier studies were performed by the FAA on the safety of side facing aircraft seats (Shams *et al.*, 1995, Lankarani *et al.*, 1999). Both studies interactively used full-scale seat/dummy impact tests and computer simulations to evaluate a number of potential occupant injury parameters. The Head Injury Criterion (HIC) and the Pelvic Injury Criterion, both adopted from automotive side impact regulations, were used to assess injury. Neck injury tolerance levels for the neck loads were adopted from Mertz and Patrick (1967, 1971) and a limit for the lateral neck moment from Patrick and Chou (1976). The impact complied with the 16 G 44 ft/s horizontal crash pulse for forward facing seats specified in FAR 25.652. The results of the studies showed fairly good agreement between the test and the simulations for a number of load and injury parameters. In both studies the injury parameter that consistently exceeded the tolerance level was the lateral neck moment. Specific requirements for a side facing seat were defined in none of these studies. In a literature study by Soltis (2001) a proposal was made for lateral load neck injury criteria based on N_{ij} intercepts. However, more data is required for validation of these criteria.

The objective of this study is to develop neck injury criteria and injury tolerance levels to be used as a basis for a performance standard for the certification of side facing aircraft seats. The criteria and tolerance levels are intended to be used in conjunction with a Side Impact Dummy for establishment of protection performance requirements of sideways mounted aircraft seats. This study will focus on the use of side facing seats in smaller (business) aircraft (FAR 25.562) rather than the large commercial carriers. However, the proposed injury criteria will be sufficiently general so that they can be applied to all category aircraft.

The study is approached as follows. A literature review will be performed on neck injuries, kinematics and injury mechanisms in lateral loading and existing neck injury criteria and tolerance levels. From the literature review, interim injury criteria and tolerance levels will be proposed. Computer simulations with a human model in a side facing seat at loading conditions according to FAR 25.562 will be performed to design Post Mortem Human Subject (PMHS) sled tests. Based on the PMHS test results and simulations, risk curves, neck injury criteria and tolerance levels will be developed. Tests and simulations with a side impact dummy will be performed in order to translate the established tolerance levels for humans to dummy output. The side impact dummy and its associated tolerance levels will be used to develop and validate a test procedure for the evaluation of sideways mounted aircraft seats and restraint systems. This paper presents the literature review and the computer simulations to prepare the PMHS and dummy tests.

METHODS

To get good insight into what neck tissues can be damaged during lateral loading, the functional anatomy of the neck was studied first. A literature review was performed on neck injuries, kinematics and injury mechanisms in lateral inertial loading and existing injury criteria and tolerance levels. The information was gathered from automotive, aviation, as well as medical related papers. Computer simulations were performed with a human model in a side facing seat subjected to the horizontal crash pulse specified in FAR 25.562 to predict the head and neck kinematics in lateral inertial loading. The effects of the use of belts and muscle activity were studied. Dummy model simulations were performed to determine the difference between human and dummy behavior in a side facing seat.

Literature review

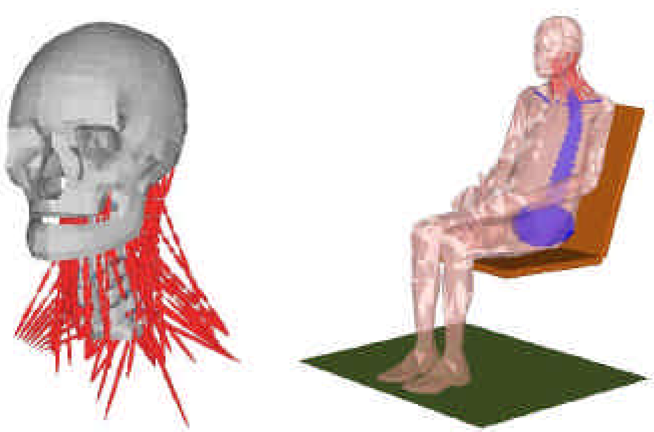
In literature studies of Lankarani *et al.* (1999) and Soltis (2001) it was mentioned that the situation for an occupant in an aircraft side facing seat during a survivable accident differs considerably from that in a side impact car crash. The main differences are:

- The acceleration prescribed in FAR 25.562, minimum peak 16 G reached in 90 ms and minimum 44 ft/s (50 km/h), is typically lower and its duration is longer than the resulting vehicle acceleration from standard side impact car crash regulations. The peak acceleration in a standard side impact car crash is approximately 80 G within 20 ms (FMVSS 214: 54 km/h, EU 96/27/EC: 50 km/h, US NCAP: 61 km/h, Euro NCAP: 50 km/h, NHTSA: 58 km/h).
- The most severe injuries of a side impact car crash victim are related to contact with the car interior, especially to head contact. Shams *et al.* (1995) and Lankarani *et al.* (1999) showed that the neck is most vulnerable in a side facing aircraft seat subjected to the acceleration prescribed in FAR 25.562.

Since in side impact car crashes head and thorax injuries are most common due to impacts with the car interior, most automotive literature is focused on head and thorax impact. Neck injury (other than whiplash) has not been a dominant occupant injury mode in car accidents, and therefore, research on this type of injury has been limited. This literature review is focused on neck lateral loading without head contact.

Simulations

Post mortem human subject. Simulations of the PMHS in lateral loading were performed using the MADYMO human model with detailed neck (Figure 1). The human model has been validated for frontal, rearward and lateral loading (Kroonenberg *et al.*, 1997, Happee *et al.*, 1998, 1999, 2000 Meijer *et al.*, 2001, Horst *et al.*, 2001, Horst, 2002). In lateral impact, the human body model has been validated for 4 to 37 G (Happee *et al.*, 2000, Horst, 2002).



a) b)

Figure 1: a) Detailed neck model; b) Combined 50th percentile male human model and detailed neck model.

Dummy. The SID, BioSID and EuroSID-1 have been used in evaluations of side facing aircraft seats by Lankarani *et al.* (1999). The EuroSID-1 was found most suitable. It was found durable, repeatable, and provided data that correlate well with the other tested ATDs. The SID was found less suitable, because it lacked the ability to measure rib deflection. It also did not provide accurate lateral flail response when restrained by only belts due to the lack of a clavicle for the shoulder belt to bear on. The BioSID was also not suitable, since it lacked a second arm and its 'far sided' spine design prevented its use in evaluating body to body contact. Durability of the BioSID was also a concern, since repairs were necessary after each test. There was also concern about the noisy nature of the acceleration data produced by the BioSID.

With regard to the EuroSID-I, the EuroSID-II has the advantage that it is equipped with an upper neck load cell and has a more integrated back-plate. Therefore, the EuroSID-II was considered to be the most suitable side impact dummy for the current study. For the simulations in this study, the MADYMO EuroSID-II facet model was used (MADYMO 2001). The EuroSID-II facet model will further be referred to as 'dummy model'.

Conditions. Simulations were performed with the human model and the dummy model on a rigid side facing seat subjected to a lateral acceleration. The geometry of the couch and the acceleration pulse were taken from earlier performed side facing seat sled tests at CAMI (Teulings *et al.*, 1998). The CAMI sled test pulse was according to the horizontal crash pulse specified in FAR 25.562 (triangular, peak 16 G, minimum duration 90 ms). A rigid wall was placed next to the subject's shoulder in order to create a worst case scenario for the neck in lateral loading. The effect of the load magnitude was determined by repeating the simulations with a pulse of half the magnitude (peak 8 G). The human model simulations were repeated with a 5-point belt. The human model 5-point belt simulations were also repeated with applying neck muscle activity of 50% of the maximum muscle forces.

a) b)

Figure 2: a) Human model in side facing seat; b) Dummy model in side facing seat.

RESULTS

Literature review

Neck injuries in lateral loading. Neck injuries in PMHS sustained by lateral inertial loading described in literature were found to vary from strained ligaments to total disc separations. The vertebrae C2 and C6 and their surrounding tissues were found to be especially vulnerable.

Neck injuries in volunteers were muscular strain and stiffness, and unconsciousness. An overview of the lateral neck kinematics and associated injuries are summarized in Table 1. The Abbreviated Injury Scale (AIS) for spine injury was used to indicate the injury severity (Table 2). Pain and AIS 1 were put in the same column, since the pain level was considered to be near injury.

Table 1. Overview of Lateral Neck Kinematics and Neck Injuries from Literature

Reference	Lateral kinematic peak values	Subject	Age (y)	Restraint	No pain/ injury	Pain/ AIS 1	AIS 2	AIS 3	AIS 4	AIS 5
Gadd *et al.* (1971)	Quasi-static Head 60º	PMHS 4 M	>66	no	0	4	0	0	0	0
Zaborowski (1964)	Sled 2.93-3.50 G, Head 2.33-6.20 G	Vol. 39 M	20-40	Lap	12	0	0	0	0	0
	Sled 5.48-9.95 G, Head 6.59-31.61 G				17	18	0	0	0	0
Zaborowski (1966)	Sled 3.69-8.41 G, Head 5-61º, 5.51-21.16 G	Vol. 52 M	20-43	4-point	39	0	0	0	0	0
	Sled 7.70-11.74 G, Head 15-66º, 9.55-45.41 G				18	26	0	0	0	0
Horsch *et al.* (1979)	Sled 10 G, 34-38 km/h Head ±30 G	PMHS 10 M+F	23, >56	2- and 3-point	1	0	3	4	2	0
Bendjellal *et al.* (1987)	Sled 6.6-9.2 G, Head 57º, ±18 G	PMHS 11 M	51-66		4	0	0	0	0	0
	Sled 12.2-14.7 G, Head 50-75º, 1588-2526 rad/s^2, 12.5-17.2 G,				6	0	1	0	0	0
Kallieris *et al.* (1987)	Car 40, 45, 50, 60 km/h Head 70-90º	PMHS 31 M+F	19-60	3-point	5	15	9	1	0	1
Kallieris *et al.* (1990, 1991)	Car 30, 35 km/h, Head 57-80º, 32-39 rad/s, 1610-2601 rad/s^2, 14-18 G	PMHS, 3 M 5 F	24-74	3-point	0	0	1	0	1	1
	Car 50 km/h, Head: 27-58º, 8-31 rad/s, 560-1460 rad/s^2, 13-26 G				1	4	0	0	0	0

M=male
F=female

Table 2. Abbreviated Injury Scale for Spinal Injuries

AIS	Spine injury
1	Acute strain (no fracture)
2	Minor fracture, no cord involvement
3	Disc rupture, nerve root damage
4	Incomplete spinal cord, cord syndrome
5	Quadriplegia

Kinematics in lateral loading. From the studies of Wismans and Spenny (1983), Wismans *et al.* (1986), Bendjellal *et al.* (1987) and Vibert *et al.* (2001) the global kinematics in a lateral loading condition can be characterized by the following four phases (Figure 3):

1. Head translation relative to the torso opposite to the sled direction.
2. Head lateral rotation about a center of rotation located in the mid-sagittal plane, opposite to the sled direction, until lock occurs at about 10º. The lateral rotation is directly followed by a head twist and head frontal flexion. The head twist and frontal flexion are smaller than the head lateral rotation.
3. Neck lateral bending and head rotation opposite to the sled direction.
4. Head and neck rotation in the sled direction (rebound phase).

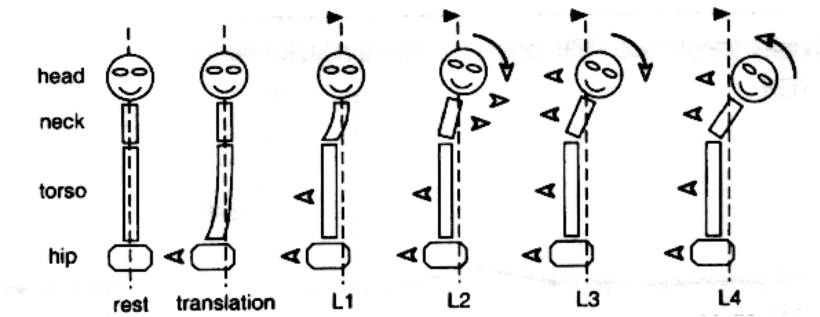


Figure 3: Sequence of trunk and head responses at lateral impact of volunteers (Vibert *et al.*, 2001).

The kinematics in lateral loading can be affected as follows (Vibert *et al.*, 2001):

- Without restraints the motion starts with a translation of the torso.
- The head motion starts earlier when restraints are used.
- The head rotations are larger when upper torso restraints are used.

Differences in the lateral loading kinematics between PMHS and volunteers are (Bendjellal *et al.*, 1987):

- The head lateral translations are larger for PMHS than for volunteers.
- Anterior flexion was seen in the PMHS tests and not in the volunteer tests.
- The head-neck lock is approximately twice as large for PMHS than for volunteers.
- The head twist is 50% less for PMHS than for volunteers.

The differences between PMHS and volunteers can probably be attributed to the absence of muscle tone in PMHS. However, the differences could also be age induced, since the average age of PMHS is about 30 years more than that of the volunteers (Table 2).

Mechanics in lateral loading. From studies of Schneider *et al.* (1975), Panjabi *et al.* (1991) and Vasavada *et al.* (2001) the following mechanics of the neck in lateral loading were found:

- The alar ligaments at the C0-C1 and C1-C2 joints provide resistance against flexion, lateral bending and axial rotation, but not extension.
- The maximum moments by voluntary neck muscle contractions in lateral bending are comparable to that in flexion, but significantly smaller than in extension.

Figure 4 shows the occipital-atlanto junction with the ligaments. This figure shows that the alar ligaments will get strained in anterior flexion and in lateral flexion. From this figure it seems that the transverse ligament of the atlas (C1) and the apical ligament of the dens will be vulnerable to anterior flexion and lateral flexion. The above-mentioned findings and the anatomical structure of the occipital-atlanto junction indicate that the neck injury criteria for anterior flexion might be applicable for lateral flexion as well.

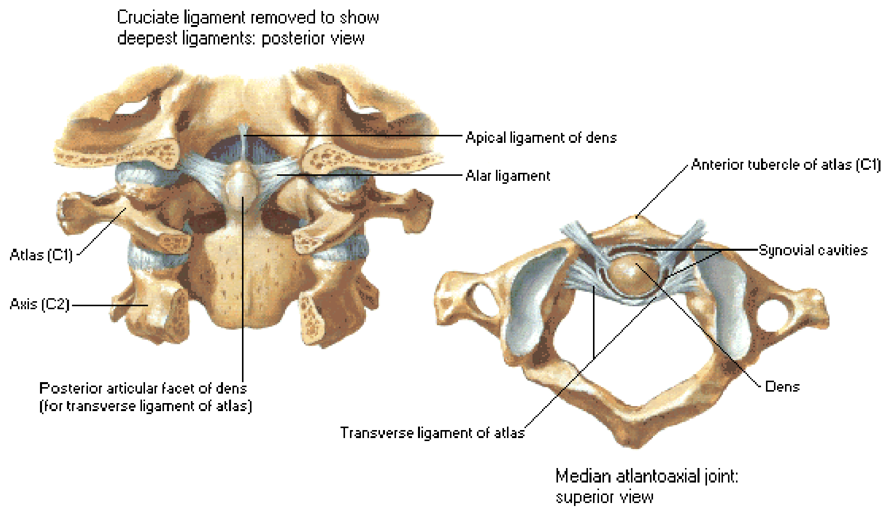


Figure 4: Ligaments at the axial-atlanto-occipital junctions (copyright Novartis 1995-1998, CD-ROM: *Atlas of human anatomy*).

Injury mechanisms in lateral loading. Mauradian *et al.* (1978) performed experiments with 11 PMHS specimens of C2-C1 segments that were laterally loaded (lateral shear). The lateral loading caused fracture at the base of the dens in 10 of the 11 specimens, and in one case a fracture into C2. The mean force required to break the dens was 542 lb. (2411 N). Dissection of the 11 specimens following failure revealed that the dens traveled with the atlas and transverse ligament as a unit. The transverse, alar, and apical ligaments were grossly intact. With a pre-programmed displacement limit of 20 mm, the anterior and posterior longitudinal ligaments were generally disrupted, but a consistent pattern was not observed.

Kallieris *et al.* (1987) mentioned that according to autopsy experience with lateral impacts, shearing load at the transition of the head-neck complex and bending of the cervical spine cause the neck injuries. Consequences of shearing loads are fractures of the occipital condyles and the dens. Hemorrhages, lacerations, transaction of the upper cervical spinal cord and vertebral fractures (processes as well as arch) occur as consequences of the lateral bending of the cervical spine.

Careme (1989) mentioned that the ligamentous junction at the level of the skull and atlas is a very tough and durable bonding, lacking the elastic properties of the cervical spine below this junction. This relatively inelastic fibrous bonding at the occipital-atlanto junctions appears to be especially vulnerable to lateral shearing forces. Rupture occurs at this junction with sparing of the more elastic structures in the cervical spine below. According to Ullrich (2001), the alar ligaments are especially vulnerable to rotational movements and have a resistance to rupture of about 240 N.

Possible neck injury criteria and tolerance levels for lateral loading. EEVC Working Group 11 (Lowne, 1996) proposed the injury criteria FNIC for frontal neck loading, based on the study of Mertz (1993) and data of Mertz and Patrick (1967, 1971). FNIC consists of three components: axial tension, axial compression and fore/after shear forces, as illustrated in Figure 5. This figure also shows that in FNIC the duration of the load is taken into account. For lateral neck bending, Patrick and Chou (1976) found that for volunteers the load at the occipital condyles at discomfort was 360 in-lb (40.7 Nm).

Soltis (2001) proposed two forms of candidate tolerance levels for neck lateral loading. The first form was based on the head and neck kinematics and the second on neck loads and moments. The tolerance levels for the head and neck kinematics were based on mainly the same publications as mentioned in Table 1. The tolerance levels for the neck loads were based on data published by Mertz and Patrick (1967, 1971). The tolerance level for the lateral neck moment was based on data published by Patrick and Chou (1976). The injury criteria and tolerance levels proposed by Soltis (2001) for a 50% male dummy are given in Table 3. The neck loads and moments were proposed to be used in NHTSA's injury criteria N_{ij}. A graphical depiction of the proposed N_{ij} lateral load neck injury criteria is shown in Figure 6.

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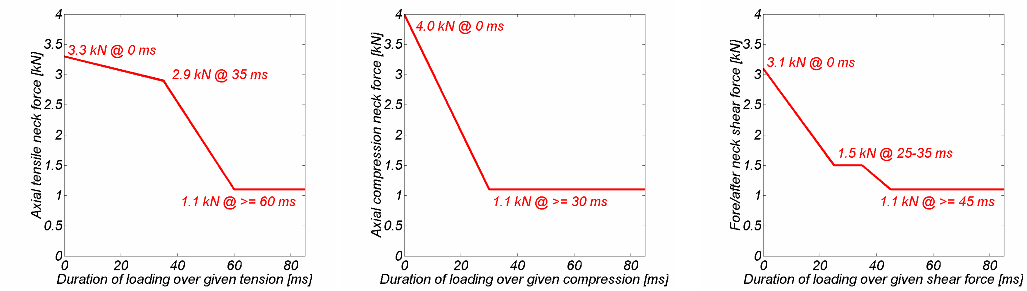

Figure 5: Tolerance levels for tension, compression and shear for frontal neck loading criteria FNIC proposed by EEVC Working Group 11 (Lowne, 1996).

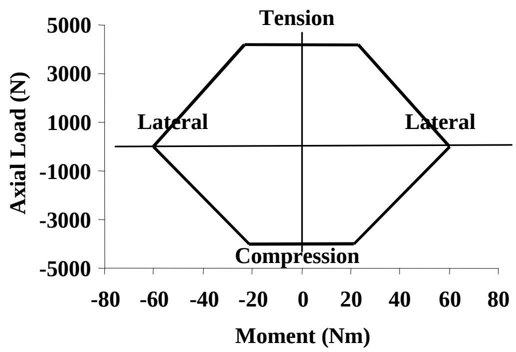

Figure 6: Tolerance levels for neck lateral load N_{ij} criteria proposed by Soltis (2001).

From the literature review, possible injury criteria and tolerance ranges for neck AIS 1 and AIS 2 were composed (Table 3). It must be noted that for AIS 2 the tolerance ranges are based on less subjects than for AIS 1. Too few literature data were found to compose a range for AIS 3 or higher. From Table 3 it can be seen that the impact velocity, head angular velocity and the linear head acceleration have a large range in which neck AIS 1 can be expected. This indicates that these criteria are not well correlated with neck injury. The head angle and angular acceleration seem to be good criteria. Most studies lack measurements or calculations of the neck bending moments, which does not mean that this criterion can be neglected. The axial tension and compression limits for the neck are not expected to be critical in lateral loading in a side facing aircraft seat if no head contact takes place. According to some literature the occipital-atlanto junctions are especially vulnerable to shearing loads. Therefore, the shear force can also be a good criterion.

Since the duration of the prescribed acceleration in FAR 25.652 is much longer than in automotive impacts, it is possible that this will lower the tolerance levels of the neck injury criteria. This can be seen in the injury criterion FNIC (Lowne, 1996), which accounts for the duration of the load.

Table 3. Approximated Neck Injury Criteria and Tolerance Levels in Lateral Loading Summarized from Various Literature Data.

		Tolerance range for neck injury		Soltis (2001)
		AIS 1	AIS 2	AIS 1 – AIS 2
Input:	Impact velocity	<40 km/h	30-60 km/h	
	Impact acceleration	5-10 G	10-14.7 G	
Kinematic criteria:	Head angle	50-70 degrees	57-75 degrees	60 degrees
	Head angular velocity	8-30 rad/s	32-39 rad/s	
	Head angular acceleration	680-1460 rad/s^2	1588-2601 rad/s^2	2600 rad/s^2
	Head linear acceleration	13-32 G	12.5-18 G	36 G
Load criteria:	Neck bending moment	22.6-40.7 Nm	40.7-60 Nm	60 Nm
	Tension	?	4170 N	4170 N
	Compression	?	4000 N	4000 N
	Shear force	>240 N	>900 N	

Simulations

Effects of belts and neck muscle activity. The human model responses with and without a 5-point belt and 50% neck muscle activity are shown in Figure 7 and Figure 8. The dashed horizontal lines in the figures show the AIS 1 range from Table 3. The continuous horizontal lines show the AIS 2 range. Comparing the peak responses of the human model in the three different situations, it can be seen that:

- At 16 G impact, as well as at 8 G impact, the 5-point belt increased the peak head lateral angle, but did not affect the peak head lateral angular acceleration significantly.
- At 16 G impact, as well as at 8 G impact, the simulated muscle activity decreased the peak head lateral angle and the peak head lateral angular acceleration significantly.
- The effect of the muscle activity was larger for the 8 G impact than for the 16 G impact.

Comparing the peak responses of the human model with the tolerance ranges given in Table 3, it can be seen that:

- At 16 G impact, the peak head lateral angle of the human model exceeded the AIS 2 range in all three different situations.
- At 8 G impact, the peak head lateral angle of the human model without belts exceeded the AIS 2 range, with 5-point belt was just within the AIS 2 range, and with 5-point belt and muscle activity was at the start of the AIS 1 range.
- At 16 G impact, the peak head lateral angular acceleration of the human model with and without 5-point belt is within the AIS 2 range, and with 5-point belt and muscle activity at the edge of the AIS 1 range.
- At 8 G impact, the peak head lateral angular acceleration of the human model with and without 5-point belt is within the AIS 1 range, and with 5-point belt and muscle activity far below the injury zone.

In the simulations it was also seen that the 5-point belt decreased the movement of the thorax and as a result decreased the rotation of T1. Thereby, the head rotation was increased with respect to T1. This explains the larger head rotation when upper torso restraints are used found by Vibert *et al.* (2001). No significant difference was seen regarding the translation of the torso and head between the simulations performed with and without 5-point belt, in contrast with Vibert *et al.* (2001). This was caused by the rigid wall placed next to the subject's shoulder in the simulations, which was not the case in the study of Vibert *et al.* (2001), mainly prevented the torso translation.

The simulations showed that muscle activity decreases the head lateral rotation as well as the head-neck lock angle. Thus, the larger head lateral translation and head-neck lock angle for PMHS than for volunteers found by Bendjellal *et al.* (1987) could probably be explained by muscle activity. In the simulations it was also observed that the muscle activity decreased the head anterior flexion and head twist. Consequently, the head anterior flexion seen in the PMHS tests and not seen in the volunteer tests performed by Bendjellal *et al.* (1987) could also be explained by muscle activity. The head twist was a bit decreased by the muscle activity in the simulations, while a larger head twist was seen in the volunteer tests than in the PMHS tests performed by Bendjellal *et al.* (1987). An explanation for this difference could be that in the simulations all the neck muscles were activated at the same level, while in reality this might not be the case. Different activation levels for different muscle groups can affect the magnitude of the head twist in lateral loading.

The AIS 1 range from Table 3 resulted from an input acceleration of 5-10 G, and the AIS 2 range from 10-14.7 G. Considering these input accelerations, the simulation results of the head lateral angle and head lateral angular acceleration were expected to be inside the AIS 1 range for the 8 G impact and just outside the AIS 2 range for the 16 G impact. Therefore, the simulation results generally coincide with the literature data. In addition, the movies of the human model simulations showed the four kinematic phases described in the literature.

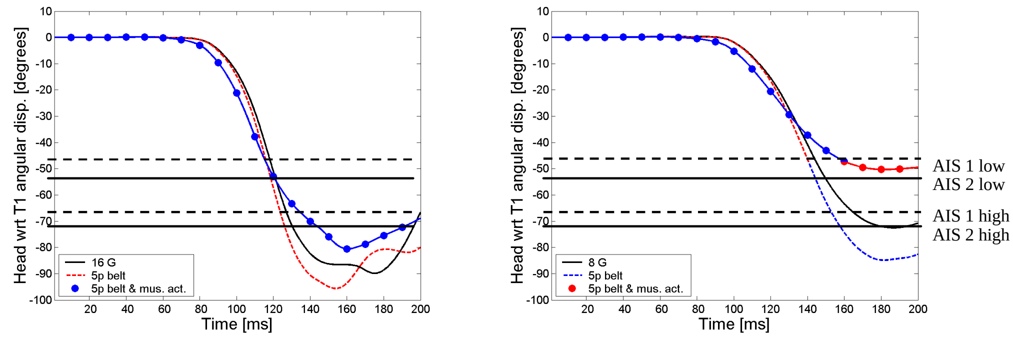

Figure 7: Head w.r.t. T1 lateral angular displacement. Human model situation without belts compared with 5-point belt and with 5-point belt plus 50% neck muscle activity.

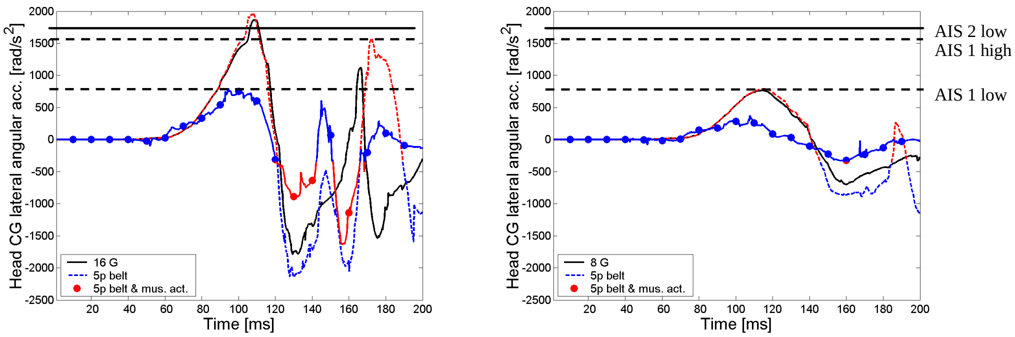

Figure 8: Head lateral angular acceleration. Human model situation without belts compared with 5-point belt and with 5-point belt plus 50% neck muscle activity.

Differences between human and dummy. The human model responses are compared with the dummy model responses for the 16 G and 8 G lateral impacts without belts in Figure 9 and Figure 10. Comparing the peak responses of the human model with that of the dummy model, it can be seen that:

- At 16 G impact, as well as at 8 G impact, the peak head lateral angle of the human and dummy model is comparable.
- At 16 G impact, as well as at 8 G impact, the peak head lateral angular acceleration of the human model is significantly lower than that of the dummy model. The difference is largest at 8 G impact.

These simulation results indicate that the dummy model is significantly stiffer than the human model. The simulations also showed a larger head twist for the human model than the dummy model. The dummy model simulations showed the four kinematic phases described in the literature.

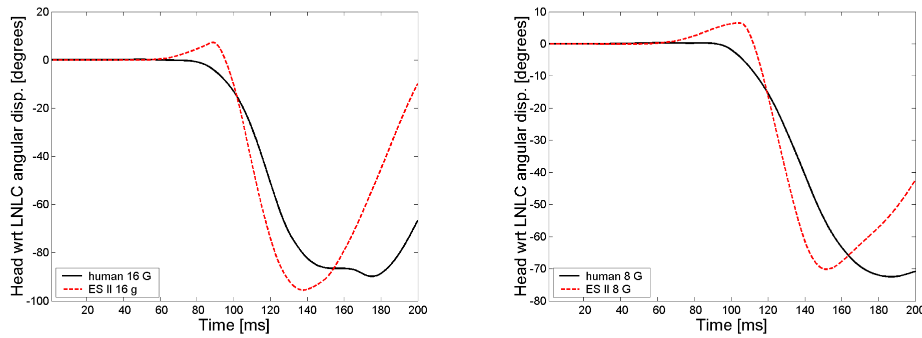

Figure 9: Head w.r.t. LNLC (lower neck load cell) or w.r.t. T1 lateral angular displacement. Human model compared with dummy model.

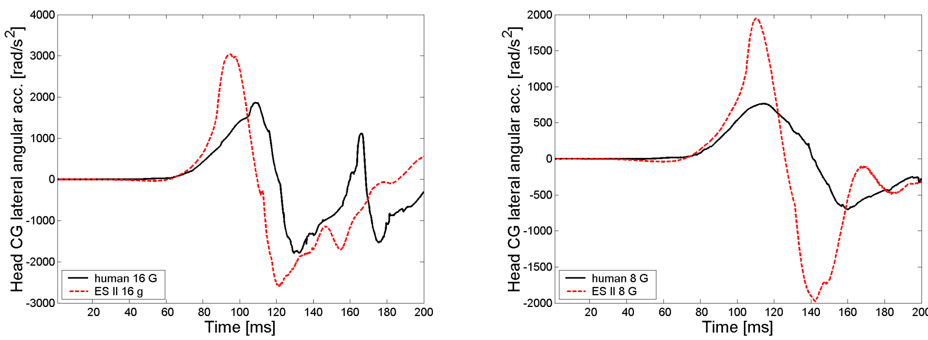

Figure 10: Head CG lateral angular acceleration. Human model compared with dummy model.

CONCLUSIONS

The objective of this study was to develop neck injury criteria and injury tolerance levels for a performance certification standard of side facing aircraft seats. In this paper, the literature review and computer simulations to prepare future PMHS and dummy tests were presented. Neck injuries, kinematics and injury mechanisms in lateral inertial loading and existing injury criteria and tolerance levels were studied. Computer simulations were performed using a human model with detailed neck to predict the head and neck kinematics. The effects of a 5-point belt and muscle activity were evaluated. Simulations with a EuroSID-2 model were performed to determine the difference between human and dummy behavior in a side facing seat.

Combining the results of the literature review and the simulations described in this paper, the following was concluded:

- The vertebrae C2 and C6 are expected to be especially vulnerable in lateral inertial loading.
- The head angle, head angular acceleration, upper neck moment of force and upper neck shear force are expected to be good injury criteria for neck lateral loading.
- PMHS are more vulnerable than volunteers. This could be age induced; however, the simulations showed that muscle activity significantly decreased the head angle and head angular acceleration. The effect of muscle activity was significantly larger for an 8 G impact than for a 16 G impact.

- The kinematic phases during lateral inertial loading described in the literature coincide with the human model and dummy model kinematics.
- The simulations indicate that the dummy model is significantly stiffer than the human model.
- The peak head lateral angle and angular acceleration of the human model coincide with the literature data.
- For a lateral 16 G horizontal impact complying with FAR 25.562, a neck injury of level AIS 2 or more can be expected.

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DISCUSSION

PAPER: **Side Impact Neck Injury Criteria and Tolerances in Aerospace Safety**

PRESENTER: ***Matt Philippens, TNO Automotive***

QUESTION**:** *Guy Nushultz, DaimerChrysler*
You're concluding that the post-mortem human subject will be harder to injure because they're older and they're–

ANSWER**:** They're not harder. Easier. Sorry.

Q: What?

A: Easier.

Q: I'm sorry. Easier. But, there's a lot of factors, a lot of injuries which cannot be detected with post-mortem sub[jects]–particularly for AIS 1, in which you won't be able to detect them. So, they functionally could become more - harder to injure than the actual live subject. Have you taken that into account? And particularly, you know, even for some AIS 3s you might have that problem. Have you taken that into account in trying to–?

A: Not yet, but the thing we're looking for is that we would like to have a tolerance level which maybe just allows an AIS 3. So, that means that the AIS 1 which may, as you say, is hard to detect in a PMHS is not important if you're looking at, let's say, the AIS 3 with respect to that there are injuries which will be rated AIS 3 but which will be hard to detect in the PMHS. You're right. I hope we could, by doing that and parallel to the PMHS test used to model, that maybe we could get an answer to that or focus or make this information more specific.

Q: Does the model have the ability to, say, rupture blood vessels or tear ligaments?

A: No vessels. It's only muscles.

Q: A second question has to do with your dummy that you're using. You're gonna try and find the mapping between the dummy neck, because the dummy neck is stiffer and behaves responsible, or you're actually going to try and change the dummy neck from–?

A: Why, that depends on what the result on the PMHS is. Let's say, that on the first phases of– Let's say on the first phase of dummy, of cadaver testing is done in combination with the simulations. From there, we have to conclude upon if it's still possible to use the dummy because it's–I think, actually, if you restrict or stay to the same restraint of five [point] belts that I use now that the dummy is really too stiff because you're relying on the interaction between the belt system and the shoulders. In this case, it seems that just a lap belt is better to prevent injuries. But, the problem is: It's a couch where you're sitting on, and there are three or two people [next] to each other, so they are going to hit each other and that's a secondary problem we haven't been looking at yet.

Q: Do you think you're going to be able to run enough cadavers to get enough information to make those decisions?

A: We hope so, in combination with the simulations that we're going to run. There are about 10 full-body tests planned at the moment, and that's not a lot. But-