

Whiplash Injury and Head Injury Criterion during Deceleration

K. Jelen^{*}, O. Fanta, R. Billich, D. Hadraba, P. Kubový

Department of Anatomy and Biomechanics, Faculty of Physical Education and Sport, Charles University in Prague, Czech Republic

** Corresponding author: jelen@ftvs.cuni.cz*

DOI: 10.2478/v10158-011-0022-8

ABSTRACT: Cervical spine injuries have become an urgent problem in modern society. Regardless of social status or background, the high rate of neck injuries is a serious healthcare issue worldwide. The cervical spine injury is mainly caused by external impact and is termed as whiplash injury. In addition, the head also performs a whiplash movement during rapid deceleration. The aim of this study is to monitor and describe physically the natural response of the head to rapid deceleration. The methodology of using an impact simulator was adopted for simulating a load which is applied to passengers wearing a seat belt in a head-on collision of a car at the speed of 30 km/h. Furthermore, a series of comparative tests of two versions (impact with and without a blindfold) were conducted to determine the influence of vision and consciousness on risk and the seriousness of trauma and the results were compared with measurements on a dummy.

KEYWORDS: Biomechanics, head injury, neck injury, whiplash, impact, deceleration.

1 INTRODUCTION

Scientists have been trying to reveal the mechanisms of acute neck injury and invent new treatment procedures for more than 50 years. The exercise regimes and general advice are considered to be the most effective treatment (McKinney, 1989). The implementation of prevention strategies into the automotive safety elements plays an important role. The mechanism of the cervical spine injury may occur during everyday activity (sport, etc.), however, it is mainly caused by a crash of two or more vehicles. The most common situation when neck injury occurs is a rear-end collision and almost one third of all neck injuries are diagnosed after a head-on collision (Kullgren, Krafft, Nygren, & Tingvall, 2000). Approximately 65 % of all cervical spine trauma are sustained at low speeds (Castro et al., 1997).

The injury which is caused by a rapid, unexpected head movement due to external impact is called “whiplash injury” (Fig. 1). The injury originates from the principle that the head, or the neck especially, is sharply flexed or extended and the movement is suddenly followed by a massive rebound in the opposite direction. The most common cause of whiplash injury is a traffic accident (head-on, rear-end, and side collision). Not only do drivers face the risk of whiplash injury, but also all other passengers. Whiplash describes a range of signs and symptoms following the injury that often progress to chronic pain which can be felt for months or even years.

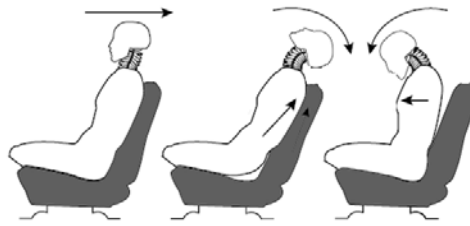


Figure 1: A demonstration of whiplash injury caused by the sudden motion of the head through an unexpected impact.

Hyperflexion of the neck resulting from car accidents is a serious health problem which can give rise to medical and economical complications (Spitzer et al., 1995). In 1969, a head restraint was integrated into the seat as an automotive safety element which limits or avoids extensive movements of the neck, thus preventing cervical spine injuries (Ruedmann, 1969).

The current papers have also discussed the possible mechanisms of cervical spine injuries. The model obtained by measuring pigs by (Svensson, Lovsund, Haland, & Larsson, 1993) proved that intracranial pressure massively increased after a sudden and powerful rear-end impact. The increase led directly to the degeneration of neurons. Other studies show cases where the ligaments of the cervical spinal region were damaged (Obelieniene, Shrader, Bovim, Misericiene, & Sand, 1999; Panjabi, 1998). In addition, damage of the cervical joints is considered to be another mechanism originating from external loading (Panjabi, 1998; Yoganandan, Pintar, & Kleinberger, 1999). An extensive number of studies have analyzed rear-end collision situations. This effort was driven by the general belief that cervical spinal injuries happen during a rear-end collision. Recent studies have suggested that a significant number of cervical spinal injuries occur after a head-on collision (Kullgren et al., 2000; Yoganandan et al., 1999). Furthermore, the muscles may play a pivotal role in the occurrence of cervical spinal injuries. Kumar, Narayan, and Amell (2002) believe that the muscles in the cervical spine area are mostly damaged by a rear-end impact at low speeds. Numerous experiments have been carried out, and dummies were chosen to be the first tested objects. Those studies were limited because no data of the muscle activity and neuromuscular responses were collected. Extensive development was achieved thanks to the study on the medical research cadavers, where obvious muscular and vertebral injuries were classified. Active research activities on human beings started in 1955 (Severy, Mathewson, & Bechtol, 1955). As the tests were executed at a speed of 50km/h and micro traumas and permanent injuries occurred, ethical committees in many countries soon set the maximum allowed speed for experiments at 36km/h. It is crucial to realize that, according to unpublished studies, the cause of the acute syndrome as a result of cervical spine injuries lies at low speeds. Moreover, the injury-causing speed does not frequently exceed the speed of 36km/h.

Considering head injuries that occur during rapid deceleration, at low speed there is no direct contact between the head and a solid obstacle (car interior), thus no head injuries are listed. There are two major mechanical causes of the various mechanisms leading to head injuries. One is a direct impact involving a collision of the head with another solid object at an appreciable velocity, and the other is a non-contact impact involving a sudden head motion without direct contact with another solid object (Goldsmith, 1972). The research on the direct contact impact of the head has been led by many researchers. Kenner and Goldsmith (1973) investigated experimentally the problem of a striker impacting on a simple model of the human head. Ruan, Khalil, and King (1994) studied the impact response of the human head using a 3D finite element analysis. Willinger, Kang, and Diaw

(1999) simulated anatomical details, including the skull, falx, tentorium, subarachnoid space, cerebral hemisphere, and brain.

This study aims to reveal the issue of non-contact impacts. The data were obtained from accelerometers which had been fixed on human participants' heads, as well as on the dummies' heads. The head injury criterion was calculated (the 3ms criterion), and the influence of the impact expectancy on the results was stated.

2 MATERIAL AND METHODS

The experiment was run using eight participants who were measured during impact at the simulator. The head acceleration in three axes and the neck muscle activity (EMG) were scanned. Firstly, each participant underwent the impact blindfolded and then again without a blindfold. The values for the head acceleration were obtained through accelerometers fixed on the participants' forehead; furthermore, the acceleration on the impact desk was measured. The information from the left and right m. sternoleidomastoideus and from the left and right m. trapezius was scanned by using an EMG. The whole setting of the impact was additionally monitored by Qualysis and a digital video camera which enabled slow motion recording.

The HIC (Head Injury Criterion) and 3ms criterion were employed to compare the seriousness of the head injury. This criterion is applicable not only for the head injury. The limiting value is 80g and this means that acceleration higher than 80g must not last over 3 ms.

The head Injury Criterion (HIC) is defined as:

$$HIC = \left\{ (t_2 - t_1) \left(\frac{1}{t_2 - t_1} \int_{t_1}^{t_2} a(t) dt \right)^{2.5} \right\}_{\max}$$

where $a(t)$ is the resultant acceleration of the head and t_1 and t_2 are variable initial and final time intervals during which HIC reaches its maximum value. Head injury criterion HIC use is based on the proposal of the National Highway Traffic Safety Administration (NHTSA), 1972 (Marjoux, Baumgartner, Deck, & Willinger, 2008). For the effects of direct impact it has been demonstrated that HIC is an acceptable discriminator between severe and less severe injuries (Tarriere, 1981). It also correlates with the risk of fractures of the skull. Determination of time interval appears to be an important parameter for calculating the HIC value. According to regulation EHK 94 (First, 2008) the threshold is then determined as $HIC = 1000$ and acceleration that is greater than 80 g for no longer than 3 ms. HIC_{36} was designed to protect the head against injuries, such as fractures of the skull, with a longer exposure when there is no contact with the hard parts of the interior.

The software package HyperWorks (application HyperView and HyperGraph) was used for evaluating the measured data. The data were filtered after importing according to the EuroNCAP methodology (EuroNCAP, 2011).

The experiment plan description: In this study, a model of the experiment and methodological study was used. The keystone of the experiment was a practical measurement framework. The point of the methodological study was to summarize possible neck injuries and compare the differences between the real head and the dummy; furthermore, it was to consider the influence of the blindfold on the seriousness of head injuries.

Test group: The dummy Manikin and eight participants of middle age (1-6 were men, and 7-8 were women) were tested. The participants were healthy and had never suffered from pain in the cervical spine.

The measurement of variables and used instruments: The biomechanical response of the participants wearing a seat belt to the impact was measured. A simulator was borrowed from the BESIP Team, for the Ministry of Transport provides the agency EuroNet.CZ, spol. s.r.o. The simulator imitates rapid deceleration.

3 RESULTS

The design of the experiment is shown in Fig. 2. Each participant was placed in the seat and a 3-point seat belt was fastened during the sled impact test. The first two measurements were taken on the dummy, followed by two measurements on each participant. Firstly, the experiment was conducted with blindfolded participants, and secondly, with full vision ability. According to the video analysis, the impacts at such low speeds do not induce the whiplash head movements.



Figure 2: The experimental video record, left - the dummy during impact, middle – a participant before impact, right – a participant during impact.

The data from the accelerometers were imported into the software HyperGraph and filtered according to the method of Euro NCAP – filter CFC 1000. CFC 1000 is characterized by the following parameters: 3 dB limit frequency is 1650 Hz, stop damping is -40 dB, and sampling frequency is at least 10 kHz.

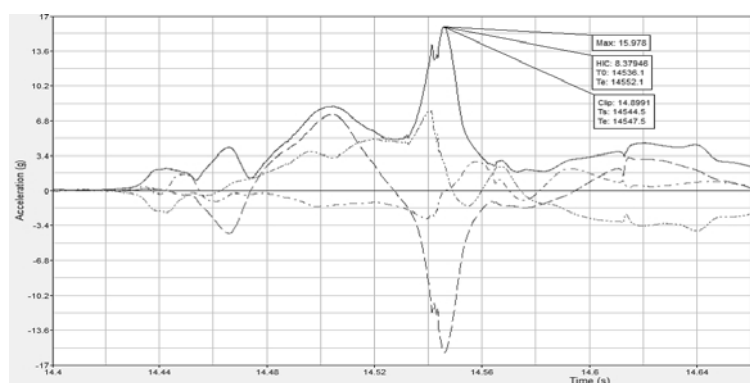


Figure 3: The measured values of the head acceleration in three axes x, y, and z and the resultant acceleration.

The values of the recorded accelerations in three axes were converted to the resultant acceleration of the head (Fig. 3) and the values of the HIC, 3ms criterion, and the maximum acceleration were computed. At the same time, acceleration of the simulator was measured to compare it with the magnitude of individual impacts.

The summarized results are listed in the tables below. For the participants see Table 1 and for the dummy see Table 2.

Table 1: The measured results for the participants (AV = Average, SD = Standard Deviation, X = failed measurement).

	Max. head acceleration (g)		HIC 36		3 ms criterion (g)		Simulator acceleration (g)	
	Without eyes	With eyes	Without eyes	With eyes	Without eyes	With eyes	Without eyes	With eyes
Person 1	10,3	7,7	5,1	3,1	10,0	7,6	29,1	29,3
Person 2	16,0	10,0	8,4	4,9	14,9	9,6	26,3	24,0
Person 3	14,1	7,0	7,0	2,6	13,5	6,8	26,8	26,4
Person 4	8,2	5,8	2,5	2,5	7,7	5,7	27,5	28,4
Person 5	13,7	14,7	7,1	11,4	12,3	13,2	X	X
Person 6	9,6	6,0	3,7	2,1	9,2	5,8	X	X
Person 7	12,9	7,6	6,3	3,5	12,3	7,3	X	X
Person 8	12,0	6,6	5,7	2,0	11,8	6,5	X	X
AV	12,1	8,2	5,7	4,0	11,5	7,8	27,4	27,0
SD	2,4	2,8	1,8	2,9	2,2	2,3	1,1	2,1
T-test	0,004		0,129		0,004		0,563	
	p<0,05		p>0,05		p<0,05		p>0,05	

Table 2: The measured results for the dummy (AV = Average, SD = Standard Deviation).

	Max. head acceleration (m/s/s)	HIC 36	3 ms criterion (m/s/s)	Simulator acceleration (m/s/s)
Dummy 1	20,0	17,8	17,5	28,7
Dummy 2	21,9	22,2	19,5	31,6
AV	20,9	20,0	18,5	30,2
SD	1,0	2,2	1,0	1,5

4 DISCUSSION

According to the results from the accelerometer located on the simulator the magnitudes of the impacts were equal (mean 25.2g) for all the tests. In addition, the linear regression approach proved there was no relationship between the magnitudes of the accelerations on the simulator and the maximum values of acceleration measured on the head, thus the matched tests (with and without a blindfold) were compared. The statistics show a positive relation between the blindfold and the maximum head acceleration (the head acceleration is higher in the second test while wearing a blindfold). The mean decrease of the maximum value between the two tests was 3.9g (38.3 m/s²), which is equal to 32 %. On the other hand the significant effect (p value .05) on the HIC₃₆ was not evidenced in the tests; however,

the HIC_{36} was lowered by 1.7 (29 %) while wearing a blindfold. The 3ms criterion reached lower values (decreased by 3.7g – 32 %) when the test was completed with full vision ability.

It is important to state that the measurements on the dummy showed higher values for all the injury criteria than on the participants. The maximum value of the dummy's head acceleration was greater (by 106 %) than that of the human head; moreover, the HIC_{36} was higher by 312 % and the 3ms criterion by 92%.

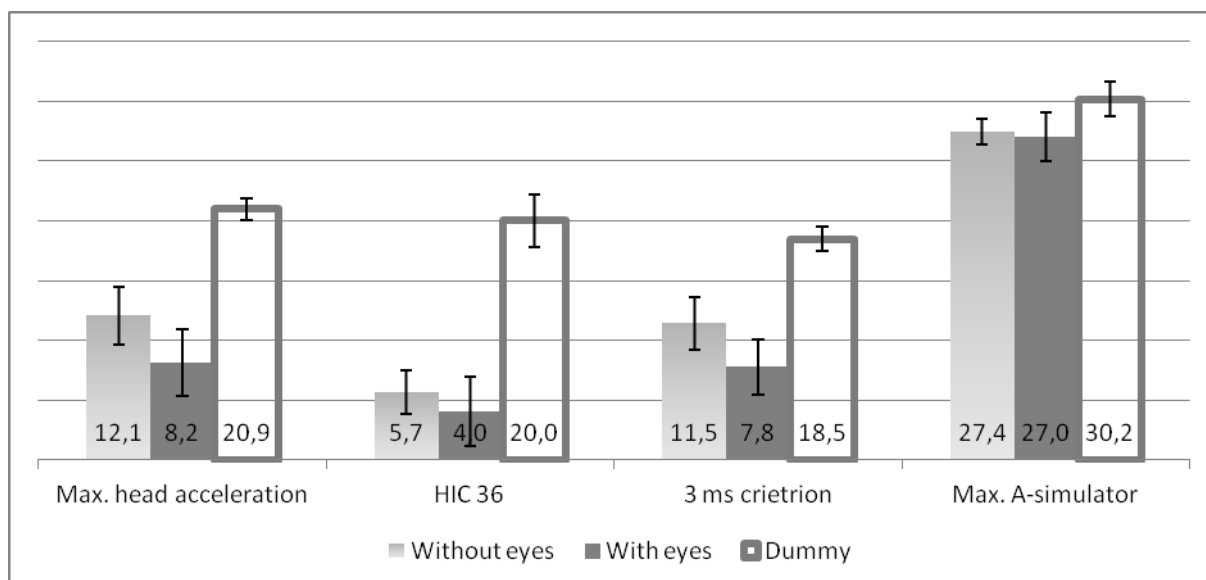


Figure 4: The values comparing the results between the dummy and the participants (with and without visibility).

A series of non-impact tests published by Gong, Lee, and Lu (2008) employed the finite element method (FEM) approach (a model of the head and neck) which was integrated with the Articulated Total Body (ATB) (McHenry, 2004), see Fig. 5. This model was placed on the model of a seat, fastened with a seatbelt, and the accelerations of 13.3, 23.5, and 33.7g were applied.

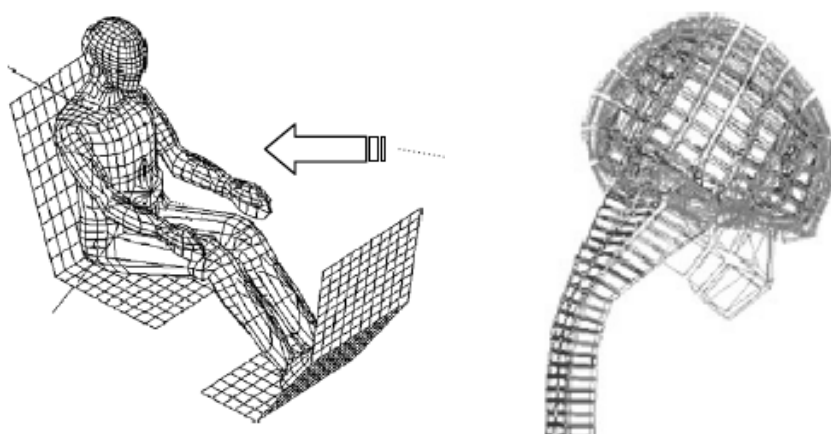


Figure 5: The model built by Gong et al. (2008) for non-impact tests.

The main goal of the study (Gong et al., 2008) was to identify intracranial pressure and shear stress. Nevertheless, the values of the head acceleration were of three orders higher than the initial acceleration of the impact (see Fig. 6). Viewed from the perspective of biomechanics, the results (Gong et al., 2008) of intracranial pressure, brain injury,

considering the resultant head acceleration, matched the aforementioned results in this study. However, the real simulation of impact did not validate the abovementioned injuries in account of the initial deceleration.

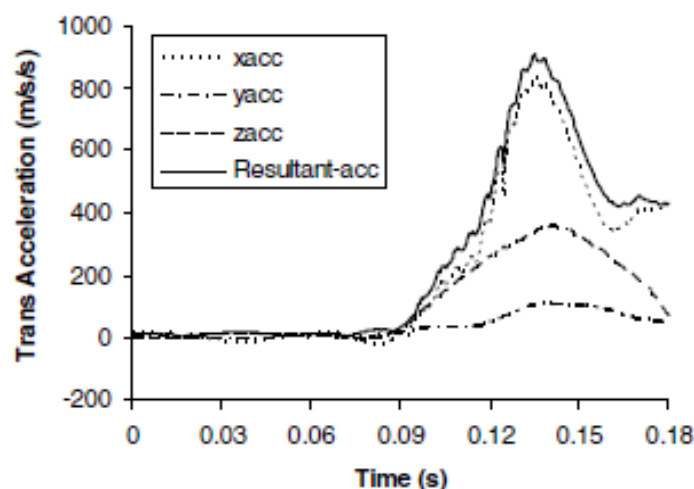


Figure 6: Resultant acceleration (Gong et al., 2008) during non-impact tests with initial acceleration 33.7 g.

ACKNOWLEDGEMENTS

This project is supported by GAUK 111310, GAČR P 407/10/1624 and SVV-2011-263 601.

REFERENCES

- Castro, W. H. M., Schilgen, M., Meyer, S., Weber, M., Peuker, C., & Wortler, K. (1997). Do “whiplash injuries” occur in low-speed Real impacts? *Eur. Spine J*, 6, 366–375.
- EuroNCAP. (2011). The tests explained Euro NCAP - For safer cars crash test safety rating. Retrieved from <http://www.euroncap.com/testprocedures.aspx>
- First, J. (2008). *Zkoušení automobilů a motocyklů*. . Praha: S&T CZ s.r.o.
- Goldsmith, W. (1972). *Biomechanics of head injury*: Prentice-Hall International, Inc.
- Gong, S. W., Lee, H. P., & Lu, C. (2008). Computational simulation of the human head response to non-contact impact. *Computers and Structures*, 86, 758-770.
- Kenner, V. H., & Goldsmith, W. (1973). Impact on a simple physical model of the head. *J. Biomech*, 6, 1-11.
- Kullgren, A., Krafft, M., Nygren, A., & Tingvall, C. (2000). Neck injuries in frontal impacts: influence of crash pulse characteristics on injury risk. *Accident Analysis Prevention*, 32, 197–205.
- Kumar, S., Narayan, Y., & Amell, T. (2002). An electromyographic study of low-velocity rear-end impacts. *Spine*, 27.

- Marjoux, D., Baumgartner, D., Deck, C., & Willinger, R. (2008). Head injury prediction capability of the HIC, HIP, SIMon and ULP criteria. *Accid Anal Prev*, 40(3), 1135-1148. doi: 10.1016/j.aap.2007.12.006
- McKinney, L. A. (1989). Early mobilization and outcome in acute sprains of the neck. *British Medical Journal*, 299, 1006–1008.
- McHenry, B. G. (2004). *Head injury criterion and the ATB*. Paper presented at the The 2004 ATB users' conference. <http://www.mchenrysoftware.com/>
- Obelieniene, D., Shrader, H., Bovim, G., Misericiene, I., & Sand, T. (1999). Pain after whiplash: a prospective controlled inception cohort study. *Journal of Neurology Neurosurgery and Psychiatry*, 66, 279–283.
- Panjabi, M. M. (1998). Cervical spine models for biomechanics research. *Spine* 23, 2684–2700.
- Ruan, J. S., Khalil, T. B., & King, A. (1994). Dynamic response of the human head to impact by three-dimensional finite element analysis. *J. Biomech*, 116, 44-50.
- Ruedmann, A. D. J. (1969). Automobile safety device—headrest to prevent whiplash injury. *JAMA* 164, 1889.
- Severy, D. M., Mathewson, J. H., & Bechtol, C. O. (1955). Controlled automobile rear-end collisions, an investigation of related engineering and medical phenomena. *Can. Serv. Med. J.*, 111, 727–759.
- Spitzer, W. O., Skovron, M. L., Salmi, L. R., Cassidy, J. D., Duranceau, J., Suissa, S., & Zeiss, E. (1995). Scientific monograph of the quebec task force on whiplash-associated disorders. *Spine* 120, 8, 1S–73S.
- Svensson, M. Y., Lovsund, P., Haland, Y., & Larsson, S. (1993). *The influence of seat-back and head-restraint properties on the head-neck motion during rear impact*. Paper presented at the IRCOBI International Conference Biomechanics of Impacts, Eindhoven, Netherlands.
- Tarriere, C. (1981). Head and Neck Injury Criteria - a Consensus Workshop (D. o. Transport, Trans.) (pp. 218-228). Washington D.C. : U.S.: NHTSA.
- Willinger, R., Kang, H. S., & Diaw, B. (1999). Three-dimensional human head finite-element model validation against two experimental impacts. *Ann. Biomed. Eng.*, 27, 403-410.
- Yoganandan, N., Pintar, F. A., & Kleinberger, M. (1999). Whiplash injury. *Biomechanical experimentation*, 24, 83-85.

**The article was produced under the support of the project
Transport R & D Centre (CZ.1.05/2.1.00/03.0064).**