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HUMAN FACE RESPONSE TO 30°-ORIENTED IMPACTS

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Abstract

The development of airbag allowed the decrease of the severity of facial injuries on face, which are often coded as minor or moderate. Yet, such injuries can lead to long-term cosmetic sequels and the study of facial bone response during impact is still an important stake, particularly in view to improve frontal crash dummies or to develop numerical models. Our study aimed firstly, to propose corridors for the head and face response during impact on frontal bone, zygoma or maxilla and secondly to define injury risk criteria for these facial bones.

Since 1995, 69 impact tests have been performed on PMHS (Post Mortem Human Subjects) faces including 34 on frontal bone and 11 on maxilla. Impacts were delivered by a horizontal guided impacting mass (17kg) equipped with a rigid cylinder at its extremity (diameter = 25 mm). Subjects were seated and the direction of impact was 30°-angled laterally from the sagittal plane of the head. The head was equipped with a metallic rig bearing 4 triaxial accelerometers localised on the top, rear, left and right of the head. These 12 acceleration measurements were used to compute the linear acceleration of the centre of gravity of the head. After tests, head autopsies were carried out and facial injuries were coded according to the AIS (Abbreviated Injury Scale) scale.

From the test results, time history corridors of the head linear acceleration and of the impact force were proposed to describe the response to impact of the frontal bone and the maxilla, with several impactor energy levels. Maximum values of head linear acceleration, impact force and HIC (Head Injury Criterion) were compared. Injury risk curves were statistically assessed for the different facial bones.

To conclude, this study provides firstly, reference data for the validation of mechanical and numerical models of the human face, in terms of corridors for time histories of mechanical response of the head during impact on facial bone. Secondly, injury mechanisms were analysed and different risk curves were defined in order to estimate with mechanical or numerical human surrogates the risk of injury for a road user involved in an accident.

Introduction

The development of the air bag resulted in decreased severity of injuries of the head and face [Krafft et al., 1998]. Using the Abbreviated Injury Scale (AIS), fractures of facial bones are often coded as minor or moderate. Yet, the personal consequences of faces injuries are considerable because of the long-term cosmetic repercussions.

Based on research on the prevention of severe injuries to the head, injury criteria have been established. The National Highway Traffic Safety Administration (NHTSA) proposed the Head Injury Criterion: it uses time duration of impact and linear acceleration of the center of gravity of the head. Past research has also proposed fracture tolerances for major facial bones using different

impactor sizes. These reference data was summarised in a SAE report in 1980 [SAE, 1980]. Later, this data was completed [Allsop et al., 1988] and overall facial response analyzed. Since contact on the face with the steering wheel is still most frequent [Thomas et al., 1991], several studies analyzed facial bone fractures following dynamic contact with the steering wheel or substitute [Nyquist et al., 1986] [Allsop et al., 1988] [Yoganandan et al., 1988] [Bermond et al., 1999].

Energy absorbed during the crush of facial bones has an influence on both facial injuries and overall head dynamics. Thus, the mechanical response of the face and its injury mechanisms must be considered to define injury criteria for head impacts. Moreover, the knowledge of mechanical response of face is essential to assess and improve the biofidelity of mechanical and numerical models of the head. Recent works performed as part of the European project called Advanced Crash Dummy Research for Injury Assessment in frontal test conditions (ADRIA) brought new data concerning dynamic response of malar bone in comparison with frontal bone under impact with a steel cylinder [ADRIA, 1998] [Bermond et al., 1999].

This paper compares impacts on the maxilla bone with the frontal bone. The aims of the analysis were firstly to propose corridors of the dynamic behaviour of head during impact on these bones, and secondly to define injury risk curves.

Material and Method

PMHS Preparation

Since 1995, 69 tests have been performed on 18 PMHS. For each PMHS, a set of anthropometric data was collected to characterise the whole corpse (height, weight) and the head (circumference, height and weight). The Frankfurt plane was materialised by four small lead balls, placed at infraorbital notches and auditory meati. The anatomical frame was defined as shown on Figure 1. The centre of gravity of the head (G) was defined according to the results of Beier et al. (1980) with $x = 8.3\text{mm}$, $y = 0$ and $z = -31.2\text{ mm}$ in the anatomical frame.

Impact conditions

A horizontal guided impactor was used. To simulate steering wheel rim, the extremity of the impactor was equipped with a horizontal steel cylinder with a diameter equal to 2.25 cm. This rigid portion directly contacted the face of the subject during tests. The weight of impacting mass was 17 kg as used since 1989 at INRETS [Welbourne et al., 1989] [Césari et al., 1989] [Bermond et al., 1999].

During tests, subject was seated, linked to the armchair by a strap and a cervical collar was used in order to hold the head in good position (Frankfurt plane in the horizontal) and to reduce the neck's flexibility. The direction of impact on maxilla or frontal bones was chosen as 30°- angled from the mid-sagittal section plane in order to obtain impact conditions closer to real impact conditions, where the seat belt leads to the rotation of the upper torso and the head.

The impactor energy taken from the mass and the velocity of the impactor was chosen on the basis of results from previous tests [Césari et al., 1989] and in a way to observe either facial bone fractures or not.

Some subjects have been submitted to both right and left impacts on maxilla. In these cases, the low energy impacts were performed first and it has been verified that they were sub-injurious impacts. Some subjects have been submitted to both impacts on frontal bone and maxilla. In this case, it was verified that the impact on frontal bone would not affect the integrity of the maxilla region.

Instrumentation

The head was equipped with a metallic rig bearing 4 blocks of triaxial accelerometers located on the top, rear, left and right of the rig (Figure 2). This rig was screwed onto the head and weighed 0.34 kg, which was generally lower than 10% of the head mass. The location of the centre of gravity of the head was adjusted by considering this additional weight. The impact force and the speed and acceleration of the impactor were also measured.

The data was prefiltered at 2.5kHz and digitised at a sampling rate of 10kHz. Then, data was filtered according to the ISO normalisation: accelerations of the head were filtered with a FIR1000 filter and data from the impactor was filtered at 180Hz.

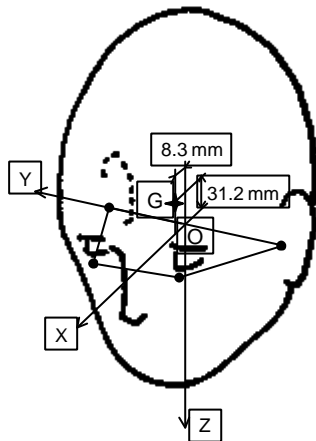


Figure 1 : Anatomical frame of the head (O = origin of the anatomical frame, G = center of gravity of the head)

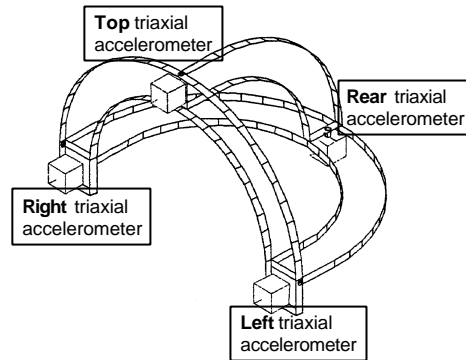


Figure 2 : Scheme of the metallic rig with 4 triaxial accelerometers

Analysis

Linear acceleration of the centre of gravity of the head The 4 triaxial accelerometers were used to calculate the linear acceleration at the centre of gravity of the head during the impact. The method used was the Nx1 method, developed by APR [Oudenard et al., 1991]. Computing is based on equations for the head motion, considering it as a rigid body. These equations contain 6 unknown parameters, which are the linear and angular accelerations of the centre of gravity of the head along X, Y and Z axis. Using information supplied by N accelerometers with $N \geq 6$ leads to a set of N equations which are resolved by an explicit integrating method.

The Nx1 method requires the orientation and position of each accelerometric sensors relative to the centre of gravity in the head anatomical frame. This information was provided by X-Ray photographs of the head and rig system, taken in frontal and sagittal planes before the test. The use of a metallic rig made the orientations and positions of the accelerometric sensors easier to obtain from X-Ray photographs.

The linear acceleration of the centre of gravity of the head was plotted as a function of time. Response corridors were proposed at several energy levels for frontal bone and maxilla bone. The maximum value of the linear acceleration was also considered.

HIC criterion The HIC criterion was computed according to the NHTSA definition [NHTSA, 1997]:

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

where $a(t)$ is the linear acceleration of the centre of gravity of the head and t_1 and t_2 are two points in time which are separated by not more than a 35 ms time interval.

Impact force The impact force was computed from the measured force of impactor, considering the mass of the impacting cylinder. As for the linear acceleration of the head, response corridors were established for impact force time-histories. The maximum value of the impact force was also considered.

Medical Investigation Autopsy of head included an inspection of the soft tissue injuries of the face, and of the skull and facial bones. Injuries were coded according to the AIS90 [AIS, 1990].

Statistics Statistical analysis was performed using the SAS® Software. The relationship between the injury risk and the measured parameters was evaluated by the test of maximum likelihood. Logistic regressions were also evaluated by test of maximum likelihood, considered as significant for a probability $p \leq 0.05$.

Results

In this paper, we present the results from 34 tests on frontal bone including 5 tests with impactor energy close to 280J and from 11 tests on maxilla including 7 tests with impactor energy close to 280 J.

Dynamic response of the head Time histories of linear acceleration of the centre of gravity of the head are given for impact on the frontal bone and the on maxilla in case of impactor energy close to 280J (Figures 3 and 4). Results in terms of maximum linear acceleration, HIC and maximum impact force are given in Table 1 for the same impactor energy level (close to 280 J). These results showed higher values in case of impact on frontal bone.

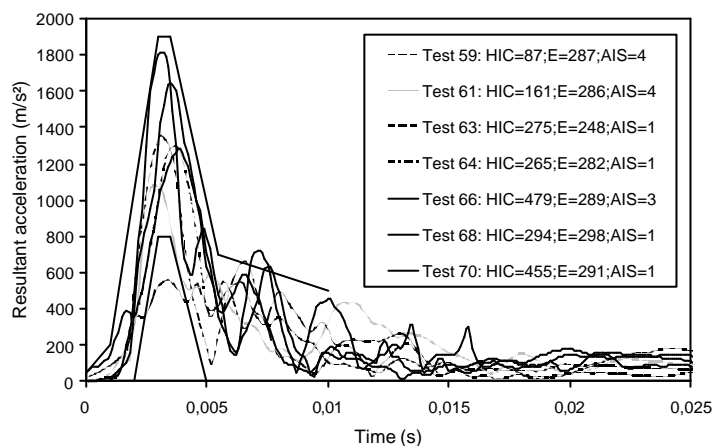


Figure 3 : Maxilla bone impact : linear acceleration of the centre of gravity of the head versus time plots and proposed response corridor

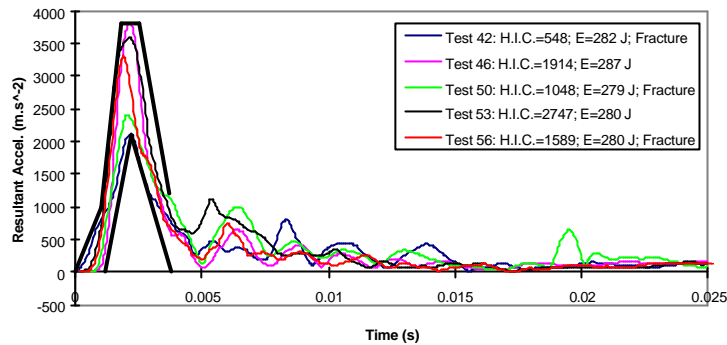


Figure 4 : Frontal bone impact : linear acceleration of the centre of gravity of the head versus time plots and proposed response corridor

Table 1 : Impact on the frontal bone and the maxilla with impactor energy close to 280 J : dynamic response of the head and impact force

	<i>Maxilla bone (n=7)</i>		<i>Frontal bone (n=5)</i>	
	Mean (SD)	Range	Mean (SD)	Range
Impactor energy (J)	283 (16)	[248 – 298]	282 (3)	[279 – 287]
Max. impact force (N)	2409 (622)	[1802 – 3331]	6818 (1144)	[5149 – 7866]
Max. linear acceleration of the cog of the head (m/s ²)	1291 (403)	[561 – 1810]	2948 (719)	[1972 – 3542]
HIC	288 (142)	[87 – 479]	1569 (840)	[548 – 2747]

SD : standard deviation; cog : centre of gravity

Observed injuries Localised abrasions or wounds of the skin were always observed and coded as AIS 1. Impacts on maxilla bone conducted to a caving in of the maxilla body and more severe impact also conducted to complete fracture of the palatine bone and/or displacement of the face with rupture of the suture between frontal and zygomatic bones. In case of severe impact on frontal bone, fractures of this bone and of the orbital cavity were observed. For both impacts on frontal bone and maxilla, an impactor energy close to 280 J seems to correspond to a juxta-injurious level, with AIS=1 (no fracture) or more (bone fracture).

Injury risk curves In order to assess the measured parameters as injury risk criteria, their values have been compared to the risk of injury, considering it equal to 100% for AIS \geq 2. For impacts on maxilla, risk distributions obtained did not allow the interpolation of the injury risk function by logistic curves since the probability of dependence was not significant. For frontal bone impact, the probability of dependence was significant for the impactor energy and the impact force. Logistic regressions are illustrated by Figure 5. Considering the impactor energy as injury criterion, a 50% injury risk is obtained for a value of 265 J. For the impact force, it corresponds to a value of 7500 N.

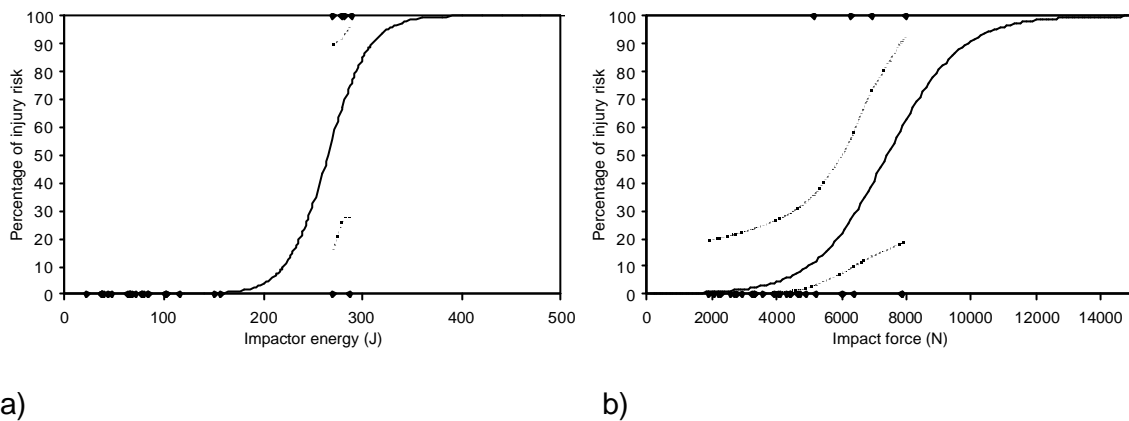


Figure 5 : Injury risk curve for impact on frontal bone using
a) impactor energy, b) impact force, as injury criteria

Discussion and conclusion

Such study using PMHS present some bias. First, it implies variations in anthropometry and in the biological tissue properties. The bone strength, which depends on many parameters (such as age, sex, living conditions, bone geometry), directly influences the mechanical behaviour and strength of the studied body part. Thus, mean PMHS behaviour is considered by building corridors and statistics are performed to obtain trends of this behaviour. Secondly, differences in the mechanical behaviour of PMHS and living persons exist, namely due to the lack of muscle tone in PMHS. However, the PMHS is still the best substitute of the human body to study particularly the injury mechanisms.

In our study, corridors proposed for the head response to facial impacts can be used as recommendations for the mechanical and numerical models. These models must be particularly biofidelic in case of juxta-injurious impact in order to predict the injury risk.

Moreover, the comparison of the head mechanical response when maxilla or frontal bone is impacted with the same impactor energy, showed the protection of the intra-cranial contents by the facial bones crushing. This comparison also shows that the injury risk curve with 50% of injury risk for HIC=1000 can not be applied in case of impact on face. Thus, new risk curve using HIC and/or new injury criteria must be defined for facial impacts.

For impact on maxilla, injury risk distribution with regard to the measured parameters did not show any threshold and did not allow us to establish logistic regressions. It seems that the impact severity in terms of impactor energy or impact force corresponds to a 50% injury risk. In order to define injury risk curve for impact on this facial bone, more injurious tests will be performed. For impact on frontal bone, it has been possible to define injury risk curve with impactor energy and impact force as injury criteria. As expected, the impact force level obtained for 50% injury risk on frontal bone (7500 N) is higher than the one obtained by Yoganandan et al. for 50% injury risk in case of steering wheel impact on zygoma (1500 N) [Yoganandan et al., 1993].

In conclusion, despite the variations in the mechanical responses of PMHS and the necessary caution in applying the results to living persons, trends in the mechanical response of

the human face to impacts with a substitute of steering wheel have been shown. Moreover, the comparison of the head behaviour when maxilla or frontal bone is impacted showed the protection of the intracranial contents by the facial bones crushing. In further research concerning the face response to impact, impacts with flat surface and/or airbag will have to be considered.

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