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## Comparison of Human Facial Tolerance and Mechanical Models

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### ABSTRACT

Facial injury detection in the crash laboratory needs new test devices and procedures. The increasing usage of improved restraint systems has reduced the severe head injuries. Instead a pattern of less severe facial injuries has emerged, e.g. the facial airbag contact.

The paper presents a comparison of the stiffness characteristics and the initial kinematics response of the facial of cadavers and mechanical models. The two mechanical models, based on a modified Hybrid III dummy head, detect localized facial forces. One was equipped with a frangible facial insert, and for the second, a load sensing face (LSF) was developed with piezoelectric sensors providing pressure times histories. Design properties and characteristics of the LSF are described as well as the sensor technique used.

Cadavers and dummy were fitted with triaxial accelerometers installed in the sagittal plane of the head at the coronal and lambdoidal sutures. The faces of all test subjects were impacted at the sub-nasal maxilla and nasion by a horizontal steel bar of 25 mm diameter.

Furthermore, a mathematical model of the human face is now in progress in an explicit finite element code.

To complete the knowledge of bone injuries obtained previously, specific tests to analyse the soft tissue injuries, were carried out.

All these results from tests are used to appreciate the means to evaluate the facial lesions and to understand the facial behaviour such as an interface, e.g. between the airbag and the head.

### ACCIDENTOLOGY

An accident data about the lesion typology was carried out, at INRETS, firstly in 1982-83 and then finally ten years after.

The aims of these two studies were to highlight the lesion typology by the types of all the roadusers, and to verify the possible evolution through these two periods of the lesions nature which should be correlated with the vehicle safety evolution and the environment evolution.

The first study (1982-1983) underscore a high frequency of face lesions, 33% of the road accident victims and 31.2% for light vehicle users. In 1993, the number of the face lesions increases for the first category to 37.2% and for the second to 35%.

We are mainly interested in lesions of the small car users for which protection solutions are in progress towards sophisticated devices, for a long time.

In 1982-1983 (Ramet, Vallet, 1987), 60% of the lesions of the face and the eyes belong to the small car users. Ten years after, the small car users represent 70% of the face injuries while all the others categories are slightly in reduction.

The drivers represent 68% of face injuries. In fact the front seat passenger are injured in 22% whereas the rear seat passenger in 10%.

The table 1 shows the repartition of lesion types in the two studies.

Tableau 1: Distribution of facial injuries

<i>LESION NATURE</i>	<i>TYOLOGY 1982/1983</i>	<i>TYOLOGY 1993/1994</i>
Laceration	80%	81.7%
Nasal Bone Fracture	10.2%	9.6%
Inferieur Maxilla Fr.	3.9%	4.3%
Naso-Maxillary Fr.	4.9%	1.4%
Cranio-Faciale Disjunction	0.08%	2.4%

We constate that if the frequency of facial lesions increase, the lesion nature and their allocation are very similar and that although the seat belt are more and more fasten. When we realize crash tests in laboratories, the dummy head hit the rim of the wheel despite the fact that the dummy thorax is more rigid than an average small car user, and the neck flexion are more limited than a human neck.

The face lesion origin are mainly due to the contacts with the steering wheel and specially with the wheel rim (Karlson, 1986).

We found quite the same lesion allocation in the literature (Jettner et al, 1986), (Chapman, 1985) and a typology difference notion in function of the contact type with the steering wheel. Principally face lesion origins are due to the contact with the wheel rim, while the contact with steering spider create cerebra lesions.

Why are we questioning us about the type of lesions ? Effectively, if these lesions are numerous, they are seldom serious. According to the AIS scale, fractures of the face are coded minor or moderate.

On 208 small car user who have face injuries ; 198 are coded AIS 1, 26 are coded 2 and only 7 are evaluated AIS 3.

But the consequences for the injured individual are considerable because of the long term psychic and cosmetic sequelae. These lesions give considerable damages and the

plastic and reconstructive surgery acts are very expensive. A study (Viano, 1985) shows effectively that face lesions cost two times more for the society than cerebral lesions.

In fact, if the nasal bone fractures give seldom partial permanent disability superior to 10%, the laceration and many face fractures can give partial permanent disability between 20% and 50% ( Fournier, 1978), (Karlson, 1986).

## FACIAL INJURY BIOMECHANICS

### Anatomy

The face is the front part of the head: it is defined as the area from the forehead to the lower jaw and includes fourteen bones, onely, the mandibule is free and has two movable joints attached to the base of the skull. The face is covered by skin and muscles. The skin is very thin particularly along the Langer lines (Fig.1).



Figure 1 : Facial bones.

### Facial skin tolerance

Facial skin is very thin. Soft tissue injuries of the face are abrasion or laceration more or less (AIS 1 or 2). These types of injuries decreased with the use of laminated windshield and seat-belt use.

A lacerative index was defined by Leug (1977) using tests performed on human skin, (living person or fresh cadavers) the data obtained found that the severities of laceration injures are depending to the lacerative resistance, the applied load and the shape of the impactor. The lacerative index is a 4 points sytem (0-4).

In the same way, the VRTC developed a chamois laceration scale to take into account cuts sustained by this "skin": The CLS is divided with 6 divisions (Jettner, Hiltner, 1986).

## Facial bones tolerance

Identified in 1900 by Le Fort which gives his name to complex types of facial fractures, automotive accidents are the leading cause of facial lesions, especially mandibular fractures (Fig.2).

INJURIES OF ALPHABETIC AND CEREBRAL F-THREMIY  
100 CARAVER TESTS  
FRACTURES OF BONES OF THE FACE.  
BASIS INJURY (15 CASES) WITH IMBINATION TO THE FALL (5 CASES)

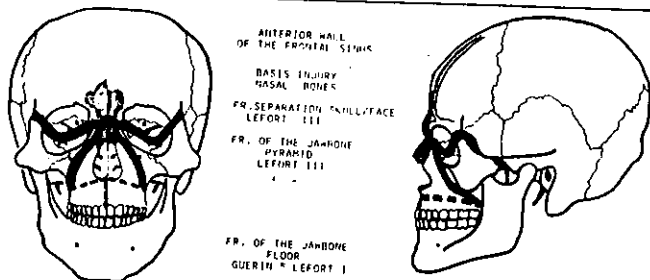


Figure 2 : Illustration of facial fracture.

The most frequent studies concern the mid-face and are conducted on fresh cadavers. The accident data indicate that the steering, specially the rim : so, numerous tests used an impactor which is an horizontal steel bar of 25mm diameter which can be compared to the ring of the steering wheel.

### Cadavers tests

A serie of tests was conducted at INRETS Laboratory to provide information on the fracture tolerance of bonny face.

Height fresh human cadavers are impacted under impact conditions simulating an impact on the steering wheel rim. The results are compared with tests conducted under similar conditions on Hybrid III dummy equipped with a frangible face.

Cadavers are impacted with a T shaped head of which the horizontal steel bar was of 25 mm diameter : The subjects are seated in front of the impactor and the sequence of impacts consisted of the following.

- Sub-injurious impact to the chin at 45° to the Franfort plane.
- Sub-injurious impact to the chin parallel with the Franfort plane.
- Potentially injurious impact to the maxilla.
- Potentially injurious impact to the nasion.

## Results of these tests

Table 2 : Mandibule tests

	impact speed (m/s)	impactor kinetic energy(j)	maximum force (N)	Remarks
1	1,40	16,3	387	no injury
2	1,85	28,4	352	no injury
3	1,94	32,2	525	no injury
4	2,67	59,2	663	cutaneus erosion
5	3,11	80,3	683	condyle fracture
6	3,06	77,7	654	wound, no fracture
7	0,00			
8	2,24	43,6	426	no injury

Table 3 : Sub-nasal maxilla tests

	impact speed (m/s)	impactor kinetic energy(j)	maximum force (N)	Remarks
1	1,40	16,2	516	fracture of nasal spine
2	1,35	15,1	660	no fracture
3	1,98	32,5	673	no fracture
4	3,14	81,7	1254	fracture of nasal spine
5	1,95	31,5	788	partial fracture of nasal spine
6	2,30	43,8	1148	fracture of nasal spine
7	3,19	88,2	1361	fracture of nasal spine
8	3,22	89,9	1049	no fracture

Table 4 : Nasion tests

	impact speed (m/s)	impactor kinetic energy(j)	maximum force (N)	Remarks
1	3,25	87,5	2503	fracture of nasal bone and septum
2	3,08	78,6	2918	no fracture
3	3,86	123,4	2781	Lefort III fracture
4	3,03	76,1	3403	no fracture
5	3,67	111,6	3760	fracture at junction of nasal and frontal bones
6	3,67	111,6	2225	Lefort III fracture
7	2,27	44,7	1875	fracture of nasal bones
8	2,26	44,2	1789	fracture at junction of nasal and frontal bones

A single impact was made at each site on each subject. During each impact, the motion of the head was recorded by high speed cameras and 6 head accelerations, the impactor acceleration and the impactor forces were recorded.

After the 4 tests, the subject was X ray radiographed. An autopsy is made and the head is defleshed to identify the fractures (Figure 3).



Figure 3 : Nasal bone fractures.

## MECHANICAL DEVICES

### A frangible face

This work provide tolerance data and permit to compare them with results obtain on frangible face (figure 4).

The same tests are conducted at INRETS (Welbourne, 1989) with hybrid III dummy equipped with a frangible face, but only to the nasion and to the sub nasal maxilla.

The data from the frangible face showed well defined thresholds at both impact sites. The frangible face provide a quite coinservative indication of the strength of the human face of the maxilla. For the nasion, the fracture threshold of the frangible face coincides approximately with the 40th percentile of the distribution of the cadaver tolerance data.

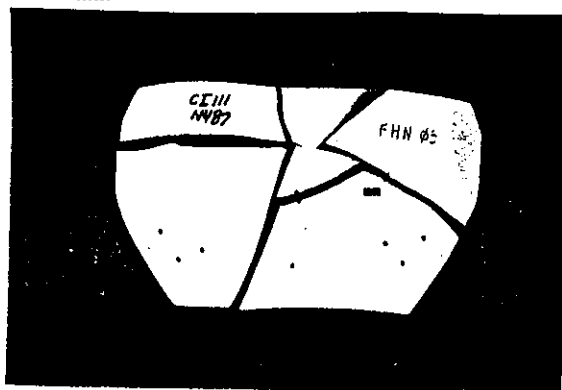


Figure 4 : Cracked frangible face.

### A load sensing face

It seems that the FMVSS 208 procedure concerning head accelerations at the head gravity center are not sufficient to detect most important facial injuries.

So, works are conducted to realize a load sensing dummy face, qui should enable researchers and designers to progress in amenagements of vehicles interior surfaces (Warner, 1986).

Impactor tests carried out at INRETS with a load Sensing Face (LSF) equipped dummy (Planat, 1988). The sensing face is the face of an Hybrid II dummy covered by twenty five piezo-electrics sensors. An instrumented face form can record time histories of impact, related to pressures.

Results are showed in figure 5.

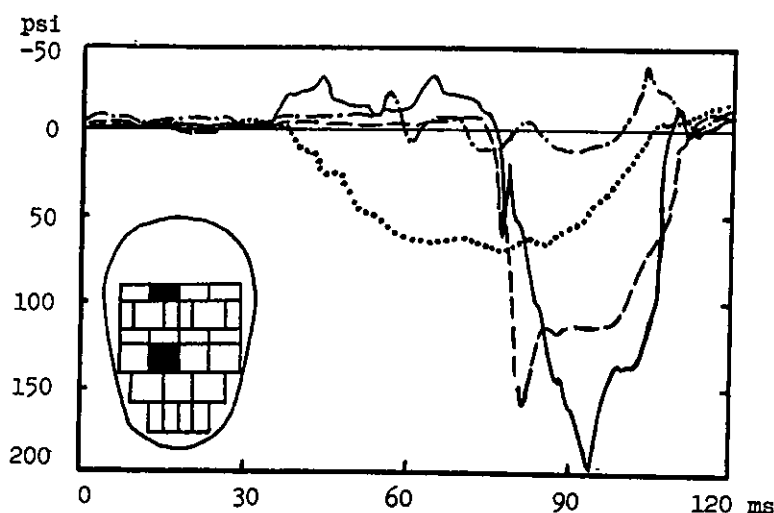


Figure 5 : Pressure time histories from LSF.

### The Bimass 150 principle

The aim of a study (Willinger, 1993) is to transfer the biomechanical head study results to a physical model intended for "measuring" the shock severity. The chosen principle is that of two masses which model the brain and skull and whose mechanical liaison has a rigidity such that the systems natural frequency is at 150 Hz. The validation of the model relies on the superimposition of the mechanical impedance physical model with the mechanical impedance of the human head in vivo.

The tests were carried out at INRETS on an impacting device setting a 16.6 kg mass in motion. The model was fixed in the neck of a Hybrid III Dummy placed in front of the impactor as illustrated in figure 6.



Figure 6 : Bimass 150 tests.

#### Remarks

The fragible face, as the load sensing face, as the bimass 150 are some needs to can measure induced trauma to the occupant face. An alternative and complementary way of the mechanical models and the cadaver tests is the mathematical model approach, to improve the knowledge in the field of the human head tolerance against the impacts.

### MATHEMATICAL MODEL

#### Mathematical model interests

A validated model can provide a rational basis for injury prevention and treatment. Experimental and mathematical modeling of head injury are two parallel methods used by head injury researchers to perform their head injury simulations. In experimental models, animals, cadavers and anthropomorphic dummies are used. Experimental modeling of head injury yields measured force, acceleration, displacement, related to injury tolerance, injury severity and type. These data help the Bioengineer in the understanding of human response and tolerance and also permits him to validate the mathematical models of head impact.

On the other hand mathematical modeling of head injury applies the principles of mechanics and numerical analysis to define the problem, formulate the solution and predict the potential of injuries. Mathematical models furnish the head trauma analysts with a powerful tool to extrapolate major experimental findings in animals and other surrogates to the human and to correlate mechanical parameters to clinical observations. In addition to that, the use of these models represents considerable cost reduction when compared to physical tests on actual specimens in the laboratory and also parameters that are virtually impossible to measure in the laboratory can be examined in detail with these simulations.

### Human head models

The development of a validated mathematical human head model has challenged many head injury researchers. If such a model can simulate the response of a living head, it will help us to answer many questions in head injury. Due to the complex geometry of the head, its non-linear material property of head impact and the difficulty of the numerical computations, mathematical models always need certain idealizations, assumptions and simplifications.

Early, the geometry was restricted to simple spherical or cylindrical shapes. A few models represented the human head more closely by finite element method. Models developed up to the 1980's have been summarized in two survey papers by King and Chou (1976), and Liu (1979).

More recently, Cheng et al (1990) formulated a two dimensional finite element model representing a coronal section of the brain to study the diffuse axonal injury (DAI) problem.

A plane strain model of a parasagittal section of an average human head was created by Chu and Lee (1991) to study the mechanisms of cerebral contusion.

Ruan et al (1991) formulated three finite element models, an axisymmetric model, a plane strain model of coronal section of human head with and without the falx and tentorium, respectively, to study the response of the head to side impact.

DiMasi et al (1991) have developed a 3-D brain model to study DAI on the premise that it is caused by cerebral strain. The model predicted shear and normal strains in the brain in response to an impact of the head to an automotive A-pillar. The model was that of an upper cerebral cortex with a longitudinal fissure to provide distinctive sagittal and coronal geometric feature and the surrounding dura including the falx partition. The dura and cortex were enclosed by a relatively rigid skull to simulate direct impact events with padded and unpadded A-pillars.

A three dimensional human head model simulating a three-layered skull, cerebral spinal fluid and brain has been developed by Ruan et al (1994) to study in more detail the coup/contrecoup response in the brain. This model was validated against cadaveric intracranial pressure reported by Nahum et al (1977). It predicted higher skull stresses and higher negative intracranial pressures in contrecoup region from occipital impact than from frontal impact. This result explains the clinical observations that more severe contrecoup injuries are presented in occipital impacts than frontal impacts.

Based on the 3-D geometry, previously presented (Ruan et al, 1994), a new 3-D human head finite element model has been developed (Jesse et al, 1993). A feature of this model was the computation of the impact force due to impact by a rigid cylinder moving at a given initial velocity, simulating the cadaveric tests of Nahum's et al (1977) whose data allowed to validate the model.

Willinger et al (1992) proposed a theoretical modal analysis of a finite element model of the human head in its sagittal plane. The results applications to the crash biomechanics field are presented. For the model validation, this approach allowed them to determine the mechanical properties of the material used to model the subarachnoid space so as to obtain the first natural frequency and mode shape in accordance with mechanical impedance recorded on the human head in vivo.

#### Remarks

Such models previously mentioned have been developed essentially for simulating the biomechanic response of the brain to dynamic loading which is vitally important for understanding brain injury mechanisms but most of them still lack prominent anatomical feature of the human head or have not been validated.

#### Methodology of the skull model

##### Geometry of the skull model

A 3D mathematical model of the human skull allowing the analysis of the human skull fracture tolerance is under development at the INRETS.

This model, utilizing the three-dimensional non-linear explicit finite element computer code PAM-CRASH (1992), is based on a real human skull, which gives a complexity structure.

The current skull model represents the face of the human that is a skull without its top and without the inferior jaw.

To obtained the geometry of a real skull, their internals and externals surfaces are punctually swept with the stylet from on ultrasonic device (GP8 from Science Accessories Corporation), which gives on a file the three dimensional coordinates of the point reached.

A mesh was drawn by hand on the real skull. The nodes of the mesh where chosen to defined volume element (most of them are represented by 8-node solid brick elements). The skull model is discretized with around 400 nodes and 200 elements

The figure 7 shows the human skull model.

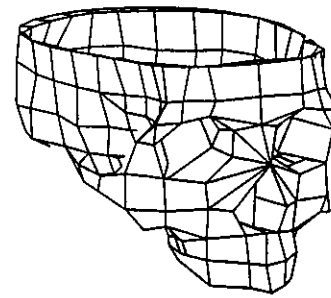


Figure 7 : Skull model.

##### Material properties of the skull

The bone of the skull is mainly cortical bones. However, certain part are composed with bone more or less dense. This influence the mechanical behaviour of the bone. So the skull model have to be shared in regard with the mechanical properties. Although there is more reliable information available on bone than any other material since it is easier to measure the constitutive properties of a hard, stiff material such as bone than to measure the properties of the softer more compliant human tissues, we did not found the specific mechanical properties of the skull bone. Also we take for the human skull model bone properties those which were estimated from available data about other bones like those from the human leg bones.

##### Mechanical properties of bones

The characteristic of the bone (Burstein et al, 1976, Keaveny et al, 1993), (table 5) is viscoelastic quality. Cortical bone is similar in properties to other fibrous materials such as wood, and has substantially more compliance than engineering materials such as metals, but more rigidity than spongy bone. Bone is one of the most rigid biological materials in the body and has a significantly greater stress carrying capacity than soft tissues which are frequently used to link long bones through joints or cover the musculoskeletal system.

Table 5 :Mechanical properties of bones

<i>Cortical bone</i>	<i>Femur</i>	<i>Tibia</i>
Density : kg/m <sup>3</sup>	1900	1900
Elastic modulus : N/m <sup>2</sup>	17.6 10 <sup>9</sup>	18.4 10 <sup>9</sup>
Plastic modulus : N/m <sup>2</sup>	0.754 10 <sup>9</sup>	1.2 10 <sup>9</sup>
Yield stress : N/m <sup>2</sup>	0.12 10 <sup>9</sup>	0.13 10 <sup>9</sup>
Ultimate stress : N/m <sup>2</sup>	0.14 10 <sup>9</sup>	0.15 10 <sup>9</sup>
Poisson's ratio	0.326	0.326

## Impactor description and model validation

To validate the human face model, it is impacted first at the nasion by a horizontal cylinder of 25 mm diameter to simulate the cadaveric tests performed at both Wayne State University (Nyquist et al, 1986) and Institut National de REcherche sur les Transports et leur Sécurité (Welbourne et al, 1989). Hence, a similar rigid impactor was built with solid brick elements.

All the impactor nodes can displace in the only direction of the impact to be in the same conditions as the experimental tests (Nyquist et al, 1986 - Welbourne et al, 1989). The model responses will be compared with the measured cadaveric test data in terms of head acceleration and impactor penetration or fracture tolerance (probability of fracture as a function of impact energy).

The analysis is performed by the PAM-CRASH finite element code (1992). This is an explicit, large displacement, Lagrangian, dynamic finite element program which is commonly used in crash worthiness analysis. The simulations are both qualitative and quantitative.

The graphical output of the post-processor (Daisy) presents the investigator with pictures that form important qualitative impressions that contribute to his ability to judge the reasonableness of the simulation and may suggest further changes in the model (e.g., mesh rezoning, changes in geometry, etc.) to improve its performance. Quantitative data is available in the form of element time histories of selected stresses and strains and nodal time histories of displacement, velocity and acceleration. In addition, the ability to plot stress and strain contours on selected cross sections of the model is also available.

## CONCLUSIONS

The aim of this paper was to compile different approaches of facial tolerance: cadavers tests, mechanical models and mathematical models.

Facial fractures are numerous and the consequences for the injured people are considerable

Moreover the fracture and collapse of the facial bones during distributed loading reduces the peak forces and resulting head accelerations in comparison to those produced by similar impact tests to the skull.

The skull model in frontal impact with Finite Element Method gives information about the internal mechanical behaviour, the kinematics, the displacement, the velocity, the acceleration, the maximum effective surface stress, the Von Mises stress and the contact force between the impactor and the tibia. We still have to simulate other use conditions of the model as a function of the stiffness, the speed or the energy of the impactor.

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