Biofidelity impact response requirements for an advanced mid-sized male crash test dummy

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BIOFIDELITY IMPACT RESPONSE REQUIREMENTS FOR AN ADVANCED MID-SIZED MALE CRASH TEST DUMMY

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ABSTRACT

The frontal crash is still an important contributor to deaths and serious injured resulting from road accidents in Europe. As the Hybrid-III dummy used in crash tests is over two decades old, the European Enhanced Vehicle-safety Committee is studying the potential for a new test device. Key is the availability of a well-defined set of requirements that identifies the minimum level of biofidelity required for an advanced frontal dummy. In this paper, a complete set of frontal impact biofidelity requirements, consisting of references, description of test conditions and corridors, is presented.

INTRODUCTION

Every year in Europe, more than 42 000 people are killed in road accidents, and over 1,7 million injuries are caused, of which several thousand give rise to severe disability (EU, 2001). The frontal crash is still a significant contributor to these numbers, accounting for an estimated 40-66 percent of the impacts that cause severe and fatal injuries to car occupants (EEVC, 1996). Hence occupant protection in frontal impact remains an important subject of research for regulatory bodies, research institutes and car manufacturers. In particular within the European Enhanced Vehicle-safety Committee (EEVC) protection in frontal impact has been subject of strong collaboration, considering ways to further improve the level of protection offered to the European car user.

In view of this, the Steering committee of the EEVC has directed a mandate to establish requirements for an advanced frontal impact dummy which should be used to assess injury risk in frontal impact and could replace the existing adult male Hybrid-III crash test dummy. This mandate has provided the start of WG 12. Considering other research programs in the area, especially the Advanced Anthropomorphic Test Device (AATD) and subsequent THOR development programs of the US Department of Transport – National Highway Safety Administration (NHTSA), the EEVC has decided to focus on the development of a worldwide acceptable frontal dummy.

Further enhancement of the biofidelity of frontal impact dummies is considered essential, especially for improved interaction with the vehicle interior and various restraint systems. For this, an up to date and agreed set of frontal impact biofidelity requirements must be available. For side impact dummies, such a set of requirements has recently been proposed by IHRA (IHRA, 2003). Accordingly the EEVC has derived a comprehensive set of frontal impact requirements that defines the minimum level of biofidelity required for an advanced adult male frontal dummy from a European perspective. The aim of this paper is to present the requirements for each part of the human body and to provide information why the respective test conditions were selected.

METHODOLOGY

Achieving desirable humanlike response is probably the most difficult part of the design of a dummy. Mechanical characteristics, such as the stiffness of the dummy at the points at which it is struck, and where it is likely to strike the vehicle, should be similar to those of similar parts of the human body. This means that the dummy should inflict damage on the vehicle similar to that found from human impacts in accidents. Similarly the dummy should deform where struck in a representative manner as particularly specified for each body component. Biofidelity impact response requirements should therefore reflect the loading conditions in real life accidents and be meaningful in terms of the injuries observed for each body region.

With the above in mind, a frontal accident survey has been undertaken based on the two largest European accident databases available: the database of the UK Co-operative Crash Injury Study (CCIS) and the German database of the Medical University of Hanover (MUH) which is financed since 30 years by BASt. The objective of the study was to prioritize injuries for different body regions of belted drivers and front seat passengers in frontal accidents and to attribute the injuries to the airbag, seatbelt and/or the internal vehicle structures. A list of injury types of the most severe and most frequent injured body parts was derived to identify dummy performance and instrumentation needs.

Following the accident survey, a critical evaluation of biomechanical data from experiments with surrogates was carried out. This included consideration of the latest biomechanical test results generated in the EC sponsored FID project (FID, 2000). The aim of the work was to study for each body region the source data available and assess their appropriateness and completeness as biomechanical design target. In this selection process, preference was given to tests in which the subjects did not sustain fractures or severe injuries (AIS<3), exceptions tolerated were deemed acceptable. Furthermore, data should be of high quality, accepted by experts in the field and, more importantly, well documented. Also, priority was given to human data - animal data was only used in the case that animal to human scaling data were available - and to dynamic above static data, as the dynamic condition better represents the actual condition in which both human and dummy are being exposed in the vehicle. Finally, preference was given to rigid above padded impacts and to contact impact tests, which provide force-deflection responses.

RESULTS

Accident Survey

The data used were from frontal accidents (11-1 o'clock), focusing on car to car and car to obstacle impacts, EES < 80 kph and vehicles not older than 1990. Occupants were 12 years or older and belted (driver and passenger). Table 1 shows the comparison of the two databases, categorized for driver and passenger and airbag deployment and not fitted/no deployment. It can be observed that the CCIS database includes more airbag cases. Further study showed that the overall accident severity of MUH data seems less than in CCIS cases, which is due to the different sampling strategy used for both databases. This difference added to the different size of the databases and different EES calculation applied, made that the databases could only be analyzed independently.

One part of the analysis examined the maximum injury to each body region for all occupants. This provided information on the frequency and severity of the injuries in the samples by body region. For example, Figure 1 shows the comparison of the CCIS sample distribution for AIS2+ injuries of restrained drivers, with and without airbag deployment.

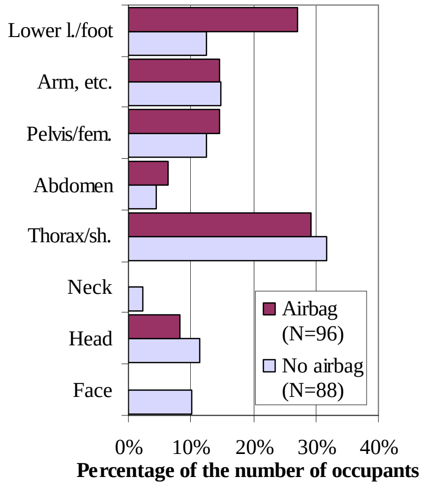

Figure 1. Distribution of AIS2+ injuries per body region for restrained drivers (CCIS)

Table 2 shows the AIS 2+ injury distribution of the MUH data set for those body regions, which will directly be addressed by frontal airbags.

Table 2.
Distribution of AIS2+ injuries per body region for restrained frontal occupants (MHH)

Percentages of AIS 2+ Injuries	Injury caused by	Injured Body Region			
		Head	**Neck**	**Thorax**	**Other**
with Frontal Airbag (n = 54 Injuries)	Airbag	14.8		1.9	
	Steering Wheel			5.6	
	Body Movement	3.7		1.9	
	Belt			3.7	
	other / unknown	3.7		1.9	
	TOTAL	**22.2**	**0.0**	**14.8**	**63.0**
without Frontal Airbag (n = 740 Injuries)	Steering Wheel	3.1		1.5	
	Belt	0.1		2.6	
	Body Movement	0.7	1.2	0.4	
	Windscreen	1.5		0.1	
	other	3.2	0.1	1.1	
	unknown	24.7	2.2	7.8	
	TOTAL	**33.4**	**3.5**	**13.5**	**49.6**

On the whole, in spite of in part small case numbers, the analyzed accidents from CCIS and MUH databases demonstrate that head and thorax are the body regions of vehicle occupants injured most severely and most frequently in frontal accidents. In accidents with airbag deployment, the injury severity of most body regions decreases considerably but remain relatively important nevertheless.

Further analysis was carried out to identify the main types of injury for the various body parts in the two samples. Severe head injuries mostly are brain injuries without fractures, but also severe skeletal injuries occurred to the head of belted drivers in accidents without airbag deployment. Severe thoracic injuries are mainly organic injuries and fractures to the ribs. In particular for drivers the percentage of fractures to the feet in accidents with

Table 1.
Comparison of accident databases

All EES < 80	CCIS	MUH
Driver – no airbag	151	747
Driver - airbag	152	57
Passenger – no airbag	87	281
Passenger – airbag	10	11

airbag deployment increases considerable. Furthermore, significant face injuries were found for drivers in accidents without airbag deployment, and for front seat occupants in accidents with airbag deployment.

Regarding the requirements for an advanced frontal dummy, the accident study has pointed out that head and thorax area are the most important part to protect. Injury assessment for these body regions, as well as for the abdomen, may require different criteria for different types of contact (e.g. steering wheel, belt or airbag) and hence the responses of these body areas should be appropriate and sensitive to different loading conditions. Injuries to the lower extremities need more attention than given so far.

Biofidelity Requirements

A short summary of the literature review, the data on which the requirements are based, and the variables for which the requirements are defined are discussed below. The test procedures and the biofidelity requirements are described in more detail in the annex.

_Face. For the definition of the biofidelity requirements of the face it is of important to define the requirements for the different areas of the face (maxilla, zygoma, frontal bone) as well as for the whole face (simulating a driver site airbag deployment), and for an impact within the direction 0±30 degrees.

In most studies concerning the face the objective has been to define fracture loads. Only a few studies has dealt with below AIS<3 injury level. On the data of the latter studies, except from the tests in which added impacts were performed, biofidelity requirements can be based. Padding materials act as mechanical filters therefore it is better to avoid their use in order to characterize the response of the face independently of the impactor properties.

Nyquist et al. (1986) and Allsop et al. (1988, 1991) published force-deflection data, which were quite variable, primarily due to the large differences in the initial low-stiffness region of the concave-upward response curves. The variability was stated to be due to the deformation of the soft tissues of the face and the deflection of the nasal bone, factors that can vary greatly between subjects. Above loads of 0.25 kN the response stiffness and the subject exhibit force deflection behavior that were more consistent. Melvin and Shee (1989) proposed two response corridors for impactor force time histories for facial impact. The first corridor, for impact to the nose with a rigid cylinder, is based on the results of the tests of Nyquist et al. (1986), in which only the nasal bone was fractured. A second corridor was defined for full-face impacts with a flat surface, based on tests performed with fresh human specimens.

Recently many oblique face impact tests have been performed (Cesari et al., 1989, ADRIA, 1998, Bermond et al., 1999, Bruyere et al., 2000). In the ADRIA (1998) study response corridors were defined for the resultant head CG for the impact to the frontal bone, zygoma and the maxilla. It should be noted that the corridor defined for the maxilla was based on tests including AIS=3 injuries. Bermond (1999) provided response corridors for impactor force and resultant acceleration of the head CG versus impact velocity for the frontal bone, zygoma, and the maxilla. It should be noted that at higher impact velocities AIS>3 occurred. Because preference should be given to tests in which the subjects did not sustain fractures or severe injuries (AIS>3), the ADRIA (1998) is used to base biofidelity requirements instead of the data obtained by Bermond (1999).

To evaluate the biofidelity of the face, four tests are proposed. The defined biofidelity requirements are based on the studies of Nyquist et al. (1986), Melvin and Shee (1988) for the frontal impact condition. For the oblique impact condition, the requirements are based on the test performed in the ADRIA project (1998). For the frontal impact condition, requirements have been defined for the following variable: The impactor force versus time response. For the oblique impact condition, requirements have been defined for the resultant head CG acceleration vs. time.

_Head. In the literature various studies can be found in which head impacts were performed. For example to look at injury mechanism (Canaple et al., 1999) or the influence of head protection (Newman et al. (1999). Both studies did not provide suitable data to base biofidelity requirements on. Data on head impact has also been obtained by Stalnaker et al. 1977, Rizitti et al., 1997, Rojanvanich et al., 1991, Troseille et al., 1992 and Ward (1985). However based on the lack of detailed description of the test set-up or due to the small number of tests performed, biofidelity requirements for the head could not be based on data obtained in these studies.

Melvin et al. (1985) has provided the most complete and detailed overview of suitable data and test protocols. In this study two corridors were defined. One corridor for a non-fracture head impact test at low impact speed (2.0 m/s) and one for a test at high impact speed (5.5 m/s). The corridors were based on the data of UMTRI (Prasad et al., 1985) and Hodgson and Thomas (1975) and are suitable for frontal, as well as lateral and rear impact tests.

To evaluate the biofidelity of the head, one test and two requirements are proposed. The requirements are based on the data of UMTRI (Prasad et al., 1985) and Hodgson and Thomas (1975), adopting the proposed corridors by Melvin (1985). Requirements have been defined for the peak impact force versus impact duration relationship.

_Neck. In literature, two sets of performance requirements have been defined to evaluate the frontal flexion/extension performance of the neck. The first set is based on volunteer and cadaver sled tests performed by Mertz and Patrick (1971) and Patrick and Chou (1976). The second set of requirements is based on volunteer sled tests carried out at the Naval BioDynamics Laboratory (NBDL) in New Orleans (Ewing et al, 1973, 1975, 1976, 1978; Grunsten et al., 1989).

As of the moment the most complete set of neck data available is the data obtained at the Naval BioDynamics Laboratory in New Orleans (NBDL data), therefore the newly defined biofidelity requirements defined were based on this data set.

To define a complete set of biofidelity requirements for the neck, the performance should be evaluated globally (in other words with regard to the environment like airbag etc) and locally focussing on the performance of the neck with regard to occurring injuries and defined injury criteria.

To evaluate the biofidelity of the neck, one test condition for the frontal and one for the oblique impact condition were defined. Requirements have been defined for the following variables, for the frontal as well as the oblique impact condition: The peak head center of gravity (CG) displacement w.r.t the sled; timing of peak head CG displacement w.r.t the sled; peak first thoracic vertebra (T1) displacement w.r.t to the sled; timing of peak T1 displacement w.r.t to the sled; peak T1 rotation w.r.t the sled; timing of peak T1 rotation w.r.t sled ; peak force at the OC joint; timing of peak force at the OC joint; peak moment at the OC joint and; timing of peak moment at the OC joint

_Shoulder. Biomechanical research concerning the behavior of the shoulder during frontal impact conditions is rare. L'Abbé et al. (1982) and Cesari et al. (1990) performed thorax belt loading compression tests. No special attention was paid to the shoulder behavior and no corridors, to be used for evaluation of the biofidelity of the shoulder, were/can be defined on the data obtained.

Vezin et al. (2002) has performed post mortem human subjects (PMHS) sled tests with and without airbag, and defined biofidelity requirement corridors for the shoulder. It should be noted that the corridors are based on a maximum of three tests, therefore further testing should be performed to refine the corridors provided. Requirements have been defined for the left and right acromion resultant acceleration, and; left and right upper humerus resultant acceleration versus time

_Spine. Biomechanical data, which properly define the kinematics of the spine under external loading of the whole body, is rare. Data, which provide the response of isolated segments of the spine, is available. However, it is difficult to extrapolate this information to how the spine would behave when it is integrated with the complete thorax and pelvis.

Vezin et al. (2002) developed biofidelity requirement corridors based on PMHS sled performed with and without airbag. Requirements have been defined for the sacrum, T1, T8 and T12 resultant acceleration versus time relationships.

_Thorax. Many studies, with the emphasis on the biofidelity aspect of the thorax have been published. To asses the thorax performance, impactor tests (Kroell 1971 &1973; Stalnaker et al. 1973; Nusholz et al. 1985, Bouquet et al. (1994), belt-loading tests (L'Abbe et al , 1982; Backaitis and St-Laurent, 1986 Cesari and Bouquet, 1990; Riordain et al. , 1991); and sled tests (Patrick et al., 1965, 1967; Yoganandan et al., 1991&1993, Kallieris , 1994,Vezin et al. 2002) have been performed.

Regarding the impactor tests the current thoracic biofidelity testing relies heavily upon the published work of Kroell (1971), despite it being over 20 years old. Due to the large number of cadavers tested, almost 50, in a variety of test configurations it is clear to see that valuable data can still be obtained from these results. In the paper by Lobdell, Kroell et al (1973) Kroell proposed performance corridors for low speed (4.92 m/s) and high speed (7.14 m/s) impactor tests. The tissue thickness and obliqueness of the impact etc. and muscle tone was taken into consideration. Kroell's work was conducted as a continuation of the work presented by Nahum et al (1970) where the same experimental test set-up had been used but the impact velocities were lower and aortic pressurization was not done. Stalnaker et al (1973) used a similar test configuration with the impact velocity varying from 5.35-6.71 m/s. No performance corridors were developed. The Stalnaker data represents a relaxed individual and provided data representing impactor penetration whereas the Kroell corridors represented sternal deflection. Neathery (1974) has re-examined the data of Kroell and Stalnaker. And concluded that, the Kroell and Stalnaker data could not be considered as a common database. Neathery (1974) proposed response corridors for the 5th, 50th and 95th percentile crash dummies similar to those proposed by Kroell, but scaled by the equations developed in the study.

Nusholtz et al (1985) has reported on a series of steering wheel impact tests. The focus of the research was on the trauma to the soft-tissue organs surrounded by the thoracic cage, as well as the kinematic response of the thoracic cage. Bouquet et al (1994) also performed impactor tests to the sternum. Two tests were performed on each PMHS, first at a sub-injury level test (impact speed ~ 3.5m/s) followed by an injury level test (impact speed ~ 5.8m/s). Corridors for the force-time history during the sub-injury test were presented. However it should be noted that these corridors are based only on two tests, therefore not providing a good representation.

In all of the above studies frontal impactor tests have been performed. Data with regard to oblique impactor tests to the thorax are rare. However Yoganandan et al. (1997) defined response corridors for the overall lower ribcage to oblique impact conditions.

Regarding the belt-loading tests. L'Abbe et al (1982) has performed early dynamic and static belt loading to examine the thoracic deflection characteristics of human volunteers in comparison with the Hybrid-III. Backaitis and St-Laurent (1986) re-analyzed the data of L'Abbe et al. (1982) focussing on the deflections of the mid-clavicle, mid-sternum and the 7th rib. Continuing on the work of L'Abbe et al. (1982) , Césari and Bouquet (1990) reported tests with 13 cadavers and a Hybrid III dummy in the same test configuration. Riordain et al. (1991) also used a similar test set-up. The paper included the results on 13 cadavers from the 1990 paper and extended the data with seven cadavers using a high mass (76.1 kg) impactor. Continuing on previous work Césari and Bouquet (1994) presented the results of an additional 9 nine PMHS. From the above, it would appear that the data of belt loading tests performed with cadavers/volunteers lying in supine position should be used to define biofidelity requirements. The loading pattern would be more representative of the real world than the Kroell test. However a limitation of these tests is that the back of the cadaver is fully supported against a rigid table, which could compromise rib deflection. A second limitation of these tests is that the test apparatus is not available/missing, therefore the tests can not easily be reproduced.

Regarding the sled tests. A few studies have been performed to evaluate the performance of the thorax in sled test. Patrick et al (1965, 1967) performed a series of sled test with unrestrained embalmed cadavers to determine the human tolerance of the head, chest and knee impact based on skeletal fracture. Four tests with different velocities were performed, in only two tests no fractures were observed. No performance corridors were defined. Yoganandan et al. (1991) performed horizontal sled impact tests (impact velocity 14 m/s, 16 G). A three-point belt was used to restrain the PMHS. Due to the occurrence of the multiple fractures the data set has not been used for defining biofidelity requirements. In a further paper, Yoganandan et al. (1993) an additional sled tests (impact velocity 9 or 13 m/s) were presented. The following restraint system configurations were used: air bag with knee bolster (AK), air bag with lap belt (AL) and air bag with three point belt (A3). It was seen that the biomechanical response of the human thorax was very different between the air bag with three-point belt loading compared to the air bag with knee bolster and airbag with lap belt restraint combinations. The data set has not been used for defining biofidelity requirements.

Kallieris (1994) performed frontal impact sled tests (impact velocity 48-55 km/h, 17 G). The following conditions restraint configuration were used: a 3-point belt, a driver side air bag and knee bolster, or a 3-point belt with supplemental driver side air bag. Biomechanical responses and the thoracic deformation contours and deflection time histories are also available in the paper. Also due to the high number of thoracic injuries occurring the data set has not been used to define the biofidelity requirements on.

Vezin et. al (2002) has performed sled tests using a sled velocity and deceleration pulse representative to the real world crashes. Tests with only seatbelt and with seatbelt and airbag were performed. Performance corridors were defined for significant variables. More data, bas the corridors on, will become available in future projects.

To evaluate the performance of the thorax three tests are proposed. The requirements are based on the data of Kroell (1971), Yoganandan et al. (1997) and Vezin et al. (2002). For the Kroell and Yoganandan tests, requirements have been defined for the force deflection relationship; for the Vezin test for the upper and lower sternum resultant acceleration versus time relationships.

_Abdomen. Nahum and Melvin (1998) have published an extensive review concerning the human abdomen biomechanics, clinical data, injury mechanism and tolerance levels. It could be seen that many studies are concerned with the biofidelity of the abdominal area for side impact conditions, only a few studies focus on the frontal impact conditions. No tests have been performed to evaluate the different impact responses caused by the asymmetric distribution of the internal organs in the abdominal cavity. Only the abdominal impact response is known for the upper and lower part without any distinction between the left, central and right region.

Cavanagh et al. (1986) and Begeman et al. (1990) performed sled tests to evaluate the abdomen behavior during frontal impact. Cavanagh et al (1986) performed low (mean 6.1 m/s) and high impact velocity (10.4 m/s) tests. Force- deflection corridors for the lower abdomen at the two impact velocities were generated. Begeman et al. (1990) performed impact tests using rigid steering columns with a soft wheel and an energy-absorbing column with rig and soft wheels. Nusholtz et al. (1985) performed a PMHS test to determine the thoraco-abdominal response with a deformable steering wheel impact. These tests were complex and it was difficult to determine the load paths, therefore the responses are not adequate to base the biofidelity requirements on. Later tests performed by Nusholtz et al. (1994) were less complicated. Impacts were performed with a ballistic pendulum fitted with a simulated rigid steering wheel assembly. A force-penetration corridor for the upper abdomen was presented.

Belts loading tests using porcine cadavers for evaluation of the abdomen have been performed by Rouhana et al. (1989). Comparisons were made between the force-deflection curves from porcine cadavers, human cadavers and living porcine subjects. Observed differences were used to extrapolate the human cadaver force deflection to living human force deflection data.

To evaluate the biofidelity of the abdomen three tests are proposed. The requirements are based on the data of Nusholtz et al. (1994), Cavanagh et al. (1986) and Rouhana et al. (1989). For the Nusholtz et al (1994) and Cavanaugh et al (1986) tests and the Rouhana et al. (1989) tests, requirements have been defined for the force versus penetration relationship.

_Femur/Knee. Only a few studies in which the femur/knee performance has been evaluated at sub-injury or at moderate injury level (AIS<3) have been published. Melvin et al, (1975), Horsch and Patrick (1976), Nusholtz et al. (1982) and Haut et al. (1995) all performed sub-injury level tests. However, Melvin et al (1975) performed padded impacts, the test set-up of Nusholtz et al (1982) is considered too complicated and Haut et al (1995) performed only isolated knee flexion tests.

Within the FID project (2000) tests similar to tests of Horsch and Patrick (1976) have been performed No response corridors have been provided by Horsch and Patrick (1976). However within the FID project corridors have been defined.

To evaluate the biofidelity of the femur/knee two tests are proposed. One with impact velocity of 2.8 m/s and the other with 4.0 m/s. For the 2.8 m/s condition requirements have been defined for the following variables: the knee impactor force versus time, the femur acceleration versus times and the iliac crest acceleration versus time. For the 4.0 m/s condition requirements have been defined for the following variables: knee impactor force versus time and the femur acceleration versus time.

_Lower Leg. Most studies concerning the lower leg performance have been aimed especially at the determination of injury mechanisms and criteria for the lower leg. The data obtained in these studies is not adequate to base biofidelity requirements on.

Yoganandan et al. (1996) performed pendulum impact test on unembalmed human cadaver legs at the Medical College of Wisconsin (MCW). Of the 26 tests performed, 13 resulted in fracture. This dataset was combined with data obtained at Calspan and Wayne State University. Age and axial force were found to be the most discriminating variables that define the risk function. Kuppa et al. (1998) analyzed Yoganandan’s data to characterize the dynamic response of the lower leg. Estimated values for the stiffness and the damping coefficient for the human lower leg were derived.

Crandall et al. (1996) reported on a series of static and dynamic tests carried out by Renault Biomechanical Research Department and the University of Virginia (UVA). Data for the volunteers were taken from a previously work by Hirsch and White (1965). Portier et al. (1997) published a detailed description of the dynamic tests performed.

Manning et al. (1998) performed a series of subinjuries heel (2 and 4 m/s) and toe impact (2,4 and 6 m/s), to assess the effect of active muscle tension. Comparative tests were performed on braced and umbraced volunteers and PMHS with an artificial Achilles tension applied. The test procedure used was based on the EEVC “ Tibia and foot certification tests”. The data of Manning et al.

(1998) has been developed into performance corridors for each impact condition and different specimen (PMHS and volunteer) by Wheeler et al (2000).

In addition to injury generating impact tests, McMaster (2000) has performed articulation tests. A basic articulation test was conducted. The ankles were tests through dorsiflexion to plantarflexion, plantarflexion to dorsiflexion, eversion to inversion and inversion to eversion. This study provided information concerning the range of motion of the ankle joint.

From the above overview of studies it can be stated that most of the data is unsuitable for basing biofidelity requirements on for the following reasons: unclear test conditions for reproduction (i.e for Yoganandan et al. (1996), Kuppa et al. (1998), Portier et al. (1997)), injury generating tests (i.e. for Portier et al. (1997) and McMaster (2000)), complex test set-up (McMaster (2000)), or static tests (Crandall et al. (1996)).

To evaluate the biofidelity of the lower leg two tests are proposed based on the corridors provided by Wheeler et al. (2000). For the toe impact condition the requirements have been defined for the following variables: the pendulum acceleration versus time; the tibial force versus time and the bending moment versus time. For the heel impact condition requirements have been defined for the following variables: pendulum acceleration versus time and tibial force versus time.

DISCUSSION

This paper presents a complete set of frontal biofidelity requirements that defines the minimum level of biofidelity required for an advanced adult male frontal dummy in Europe. The current set contains only those test conditions for which test specifications are well defined and corridors are established. Moreover, the set addresses the key body regions taking into account the various types of contact that may occur (belt, airbag, steering wheel, facia) according to the accident study.

The National Highway Safety Administration has defined a set of biomechanical response requirements for the THOR advanced frontal dummy (NHTSA/GESAC, 2001). A comparison of both sets of requirements indicates that the two sets do not lie far apart, increasing the potential for harmonization between the regions. For instance, the biofidelity requirements for the head, abdomen and femur/knee are identical between the two sets. The differences between EEVC and NHTSA proposed requirements are most notable for the neck - the oblique versus lateral NBDL impact condition- and the lower leg/ankle and foot for which NHTSA/GESAC defines a series of tests in the dynamic heel impact condition (Kuppa et al, 1998), dynamic dorsiflexion (Crandall et al, 1996), quasi static inversion (Petit et al, 1996), quasi static eversion (Crandall et al., 1996 and Petit et al., 1996), quasi static dorsi-flexion (Crandall et al., 1996), quasi static plantar flexion (Paranteau et al., 1996), quasi static internal/external rotation (Siegler et al., 1988) and dynamic inversion/eversion (Jaffredo et al., 2000). Recognizing that the requirements presented in this paper form a minimal set, it must be further investigated whether such a large set of tests for the lower leg/ankle and foot can be justified.

CONCLUSIONS

This paper proposes a set of biofidelity responses requirements for an advanced adult male frontal impact crash dummy. The set consists of requirements for the face, head, neck, spine, thorax, abdomen, femur/knee and the lower leg/ankle/foot complex, and defines the minimal level of biofidelity that is required for this dummy. The set of biofidelity requirements proposed in this paper will be used to compare the biomechanical performance of existing adult male frontal impact crash test dummies, including the Hybrid-III and THOR-alpha dummies. Comparing the presented set with the biomechanical response requirements for the THOR advanced frontal dummy shows that there is good possibility for harmonization of requirements through IHRA, and hence, for the test device that meets of these requirements.

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ANNEX: Description of Test Procedures and Biofidelity Requirements

FACE

_Face test 1 is based on the data of Nyquist et al (1986). The test is a horizontal guided impactor test (impact velocity 3.6 m/s). Using an impactor with a horizontal steel cylinder (mass 32 kg, diameter 25 mm). The dummy needs to be positioned with the anterior-posterior axis of the head horizontal and the sagittal plane vertical. Impact is performed to the nose.

Instrument the pendulum with a load cell, to be able to obtain the impact force versus time relationship. Filter all response data according to the requirements of SAE Recommended Practice J211.

The impactor force versus time response has to be within the corridor shown in Figure 1.

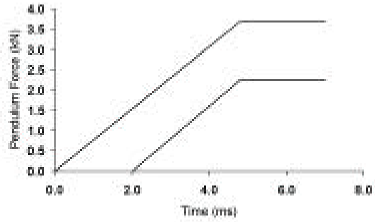

Figure 1. Pendulum Force vs. Time.

_Face test 2 is based on the data/corridor provided by Melvin and Shee (1988). The test is a full-face impact test, using a flat disk horizontal guided impactor (mass 13 kg, diameter 152 mm) impact velocity 6.7 m/s. The dummy needs to be positioned with the anterior-posterior axis of the head horizontal and the mid sagittal plane vertical. The extremity of the impactor is to impact the mid-point of the line joining the two maxilla plates on the face.

Instrument the pendulum with a load cell, to be able to obtain the impact force versus time relationship. Filter all response data according to the requirements of SAE Recommended Practice J211 with Class 1000 for the head data and Class 180 for the impactor data.

The impactor force versus time response should be within the corridor shown in Figure 2.

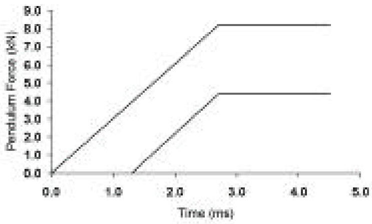

Figure 2. Pendulum Force vs. Time

_Face test 3 is based on the data/corridor provided in ADRIA (1998). The test prescribed is a 3.84 m/s, 30 degree angled horizontal impact test to the frontal bone using a rigid impactor (17 kg, diameter 25 mm). The dummy needs to be positioned with the anterior-posterior axis of the head horizontal, its mid sagittal plane vertical and adjusted to obtain an impact direction 30°-angled from the mid sagittal plane of the head.

Instrument the dummy's head with a tri-axial accelerometer located at the head CG and place a load cell between the major part of the impacting mass and the extremity. The influence of the extremity mass on the load measurement must be corrected. Filter all response data according to the requirements of SAE Recommended Practice J211.

The resultant head CG acceleration has to be within the corridor shown in Figure 3

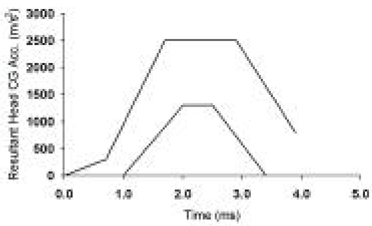

Figure 3. Resultant Head CG Acceleration vs. Time (frontal bone impact)

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Face test 4 is identical to face test 3, only the seat height should be adjusted such that impact to the zygoma instead of the frontal bone occurs. Instrumentation is similar to the instrumentation described by face test 3.

The resultant head CG acceleration has to be within the corridor shown in Figure 4.

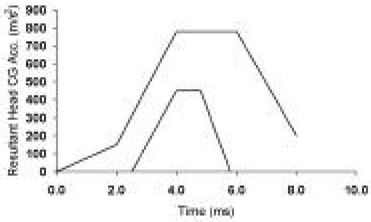

Figure 4. Resultant Head CG Acceleration vs. Time (Impact to the Zygoma)

Head

The head test is a frontal impact test, impact velocity 2.0 m/s and 5.5 m/s using a rigid flat surface impactor (mass 23.4 kg, diameter 15.2 mm). The dummy needs to be seated upright without support.

Instrument the dummy's head with a tri-axial accelerometer located at the head CG. Place a load cell on the impactor. Filter all response data according to the requirements of SAE Recommended Practice J211.

The peak impactor force for 2.0 m/s and 5.5 m/s impact respectively should be corridors provided by Melvin (1985) as shown in Figure 5.

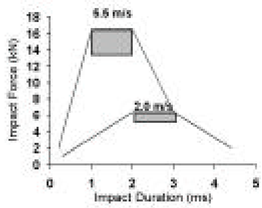

Figure 5. Peak impactor force vs. impact duration.

Neck

_Neck test 1 is a full dummy 15-G frontal impact sled test. The sled pulse used should be similar to the NBDL sled pulse. Fasten a rigid seat, functionally similar to the one used by Ewing and Thomas (1977), to the HyGe sled, facing the direction of sled travel. The dummy needs to be positioned with the anterior-posterior axis of the head horizontal and its mid sagittal plane vertical. Use a restrain system similar to the restrain system used by Ewing and Thomas (1977).

Instrument the dummy with tri-axial accelerometers at the head CG and T1. Use cameras to record the photographic markers of the head CG, OC and T1. Filter all response data according to the requirements of SAE Recommended Practice J211.

The following responses should be within the corridors provided. The peak T1 displacement and timing of the peak T1 displacement with respect to the sled, the peak CG displacement and timing of the peak CG displacement with respect to the sled, peak flexion of the head and the timing of the peak flexion of the head.

Measurement	Units	Lower Bound	Upper Bound
Peak T1 Displacement in X-Direction	mm	53	60
Timing of Peak T1 Displacement in X-Direction	ms	152	158
Peak T1 Displacement in Z-Direction	mm	-32	-17
Timing of Peak T1 Displacement in Z-Direction.	ms	149	160
Peak CG Displacement in X-Direction	mm	191	214
Timing of Peak CG Displacement in X-Direction	ms	154	159
Peak CG Displacement in Z-Direction	mm	-237	-208
Timing of Peak CG Displacement in Z-Direction	ms	162	169
Peak Flexion of the Head	deg	70	87
Timing of Peak Flexion of the Head	ms	165	176
Peak T1 Rotation about the Y-axis	deg	17	27
Timing of Peak T1 Rotation about the Y-axis	ms	145	155
Peak Force at OC Joint in X-Direction (1st minimum)	N	-1381	-801
Timing of Peak Force at OC joint in X-Direction(1st minimum)	ms	95	101
Peak Force at OC Joint in X-Direction (2nd minimum)	N	-1098	-908
Timing of Peak Force at OC Joint in X-Direction (2nd minimum)	ms	145	154
Peak Force at OC Joint in Z-Direction (1st minimum)	N	-793	-546
Timing of Peak Force at OC Joint in Z-Direction (1st minimum)	ms	89	95
Peak Force at OC Joint in Z-Direction (2nd minimum)	N	-899	-530
Timing of Peak Force at OC Joint in Z-Direction (2nd minimum)	ms	128	141
Peak OC moment about the flexion axis	Nm	-56	-46

_**Neck test 2** is a full dummy 11-G oblique impact sled test. Fasten a rigid seat, functional similar to the one used by Ewing and Thomas (1977), to the HyGe sled at an angle of 45 degrees from the forward facing direction. Attach a vertical light padded wooden board against the seat to restrict upper torso rotation and to support the torso during sled translation. The dummy needs to be positioned with the anterior-posterior axis of the head horizontal and its mid sagittal plane vertical. Use a restrain system similar to the restrain system used by Ewing and Thomas (1977). Instrument the dummy with triaxial accelerometers at the head CG and T1 Use cameras to record the photographic markers of the head CG, OC and T1. Filter all response data according to the requirements of SAE Recommended Practice J211.

The following responses should be within the corridors provided. The peak T1 displacement and timing of the peak T1 displacement with respect to the sled, the peak CG displacement and timing of the peak CG displacement with respect to the sled, peak flexion and the timing of the peak flexion of the head, and the peak twist and timing of the peak twist of the head.

Measurement	Units	Lower Bound	Upper Bound
Peak T1 Displacement in X-Direction	mm	63	83
Timing of Peak T1 Displacement in X-Direction	ms	160	165
Peak T1 Displacement in Y-Direction	mm	-27	-20
Timing of Peak T1 Displacement in Y-Direction	ms	157	162
Peak CG Displacement in X-Direction	mm	196	247
Timing of Peak CG Displacement in X-Direction	ms	156	163
Peak CG Displacement in Y-Direction	mm	-43	-7
Timing of Peak CG Displacement in Y-Direction	ms	137	165
Peak CG Displacement in Z-Direction	mm	-211	-139
Timing of Peak CG Displacement in Z-Direction	ms	171	180
Peak Flexion of the Head	deg	54	80
Timing of Peak Flexion of the Head	ms	172	187
Peak Twist of the Head	deg	-38	-24
Timing of Peak Twist of the Head		144	175
Peak T1 Flexion	deg	4	15
Timing of Peak T1 Flexion	ms	149	159
Peak T1 Twist	deg	-16	-8
Timing of Peak T1 Twist	ms	151	169
Peak Force at OC Joint in X-Direction	N	-810	-698
Timing of Peak Force at OC joint in X-Direction	ms	154	164
Peak Force at OC Joint in Y-Direction	N	363	475
Timing of Peak Force at OC Joint in Y-Direction	ms	159	167
Peak Force at OC Joint in Z-Direction	N	-702	-412
Timing of Peak Force at OC Joint in Z-Direction	ms	108	112
Peak OC moment about the flexion axis	Nm	-56	-46
Peak OC moment about the twist axis	Nm	10	21

Shoulder

_**Shoulder test 1** is a frontal impact sled test at 50 km/h with a sled deceleration close to the ECE R44-03 regulation (Child's restraint regulation), corresponding to a maximum deceleration of 22 G's and seat belt (4 kN load limiting system) and airbag as restraint system

Instrument the dummy with an tri-axial accelerometer at the upper and lower left and right arm. Filter all response data according to the requirements of the SAE Recommended Practice J211.

The following responses should be within the corridors provided. Left and right acromion resultant acceleration versus time, left lower and upper humerus resultant acceleration versus time and the right upper and lower humerus resultant acceleration versus time.

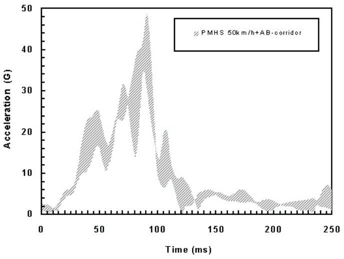

Figure 6. Left acromion resultant acceleration versus time

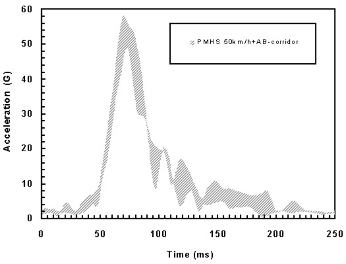

Figure 7. Right acromion resultant acceleration versus time

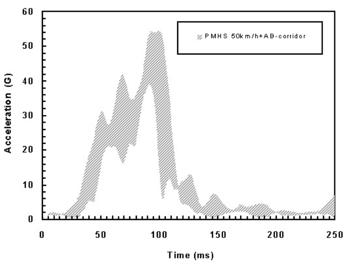

Figure 8. Left humerus resultant acceleration vs. time

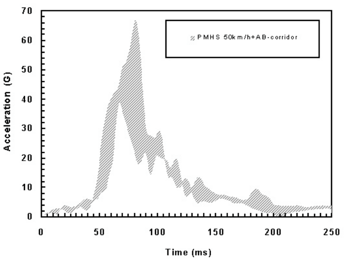

Figure 9. Right humerus resultant acceleration vs. time

_**Shoulder test 2** is a frontal impact sled tests at 30 km/h with a sled deceleration of 15 G's, close the deceleration used at the University of Heidelberg (Kallieris, 2001) and only set belt as restraint system (4 kN force limiting system).

Instrument the dummy with an tri-axial accelerometer at the upper and lower left and right arm. Filter all response data according to the requirements of the SAE Recommended Practice J211.

The following responses should be within the corridors provided. Left and right acromion and upper humerus resultant acceleration versus time.

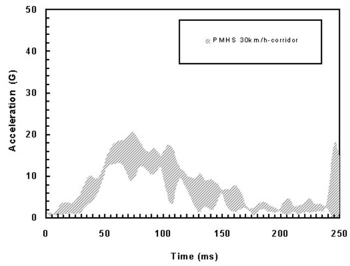

Figure 10. Left acromion resultant acceleration versus time

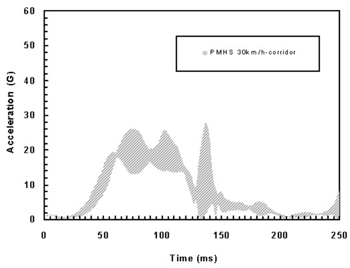

Figure 11. Right acromion resultant acceleration versus time

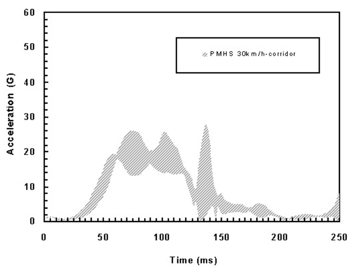

Figure 12. Left humerus resultant acceleration vs. time

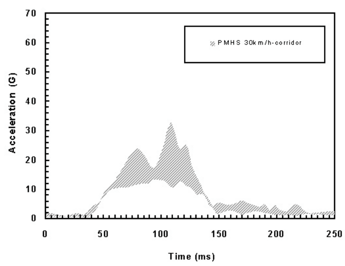

Figure 13. Right humerus resultant acceleration vs. time

Spine

_Spine test 1 is a frontal impact sled test at 50 km/h with a sled deceleration close to the ECE R44-03 regulation (Child's restraint regulation), corresponding to a maximum deceleration of 22 G's and seat belt (4 kN load limiting system) and airbag as restraint system

Instrument the dummy with a tri-axial accelerometer at T1, T8 and T12. Filter all response data according to the requirements of the SAE Recommended Practice J211.

The following responses should be within the corridors provided. The sacrum, T1, T8 and T12 vertebra resultant acceleration versus time.

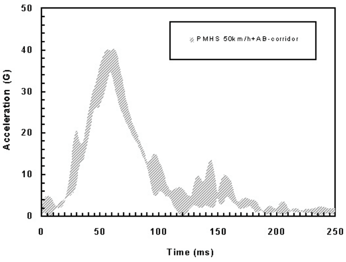

Figure 14. Sacrum resultant acceleration vs. time

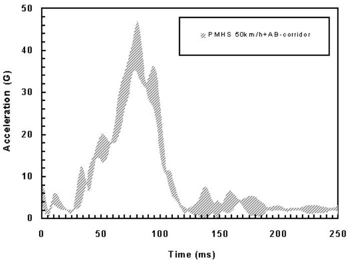

Figure 15. T1 resultant acceleration vs. time

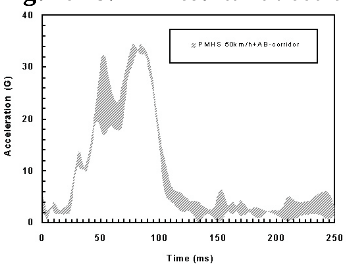

Figure 16. T8 resultant acceleration vs. time

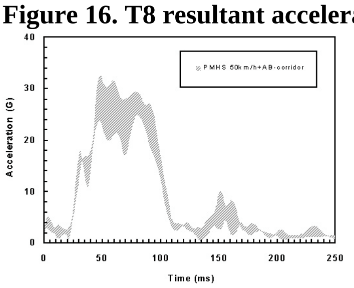

Figure 17. T12 resultant acceleration vs. time

_Spine test 2 is a frontal impact sled tests at 30 km/h with a sled deceleration of 15 G's, close the deceleration used at the University of Heidelberg (Kallieris, 2001) and only set belt as restraint system (4 kN force limiting system).

Instrument the dummy with a tri-axial accelerometer at T1, T8 and T12. Filter all response data according to the requirements of the SAE Recommended Practice J211.
The following responses should be within the corridors provided. The sacrum T1, T8 and T12 vertebra resultant acceleration versus time.

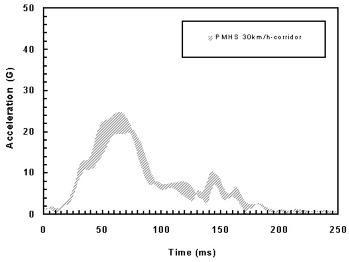

Figure 18. Sacrum resultant acceleration vs. time

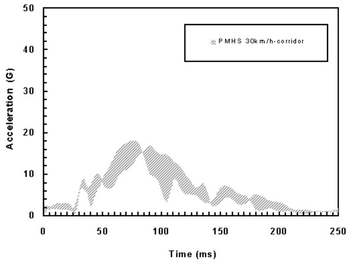

Figure 19. T1 resultant acceleration vs. time

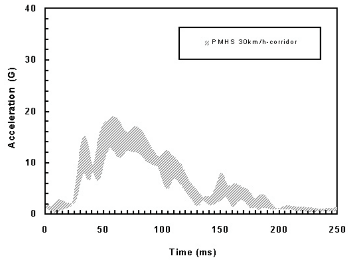

Figure 20. T8 resultant acceleration vs. time

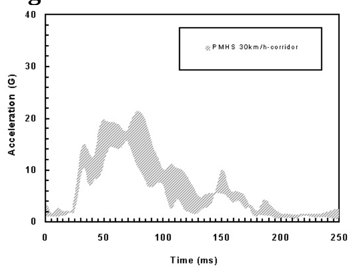

Figure 21. T12 resultant acceleration vs. time

Thorax

_Thorax test 1 is based on the published work of Kroell (1971). The test is an impactor test (mass 23.4-kg, contacting interface 152-mm diameter wooden block with a 12.8-mm edge radius. Impact applied horizontally with the contact surface perpendicular to the direction of loading and centred mid-sagittal over the fourth costal interspace at the sternum. Impact velocities 4.3 and 6.7 m/s. Dummy positioned such that the surface of the thorax is line with the impactor centerline is vertical with the longitudinal centerline of the impactor at the same height as the mid-sternum and guided in the mid-sagittal plane of the subject.

Instrument the pendulum with two uniaxial accelerometers and a velocity sensor and the dummy with chest deflection sensor.

The force-deflection response for the 4.3 m/s and the 6.7 m/s impact velocity conditions should be within the well-known corridors based on the published work of Kroell (1971) see Figure 22.

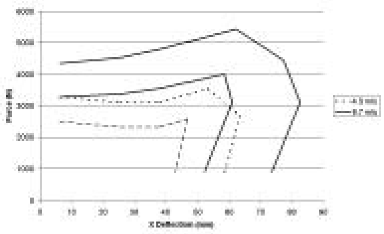

Figure 22. Force deflection corridor

_Thorax test 2 is based o the data of Yogananadan et al. (1997). The test is an impactor test (mass 23.4 kg, diameter 150 mm) to the right antero-lateral thorax, impact velocity 4.3 m/s. The lower extremities are stretched out horizontally, upper extremities are extended, back of the torso is unsupported, and the torso rotated 15º from right to left.

The force-deflection response, the force-time history and the deflection time history response should lie within the corridors provided by Yoganandan et al. (1997).

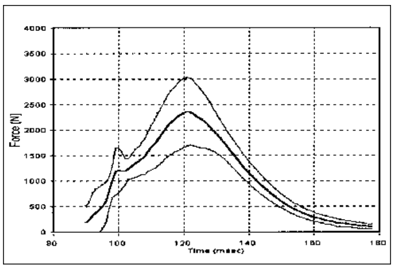

Figure 23. Force-time corridor

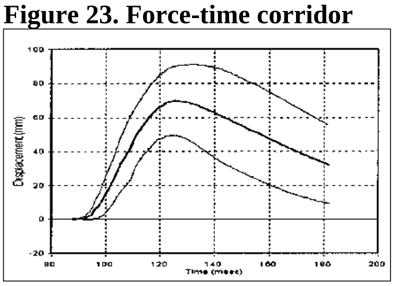

Figure 24. Deflection-time corridor

_Thorax test 3 is a frontal impact sled test at 50 km/h with a sled deceleration close to the ECE R44-03 regulation (Child's restraint regulation), corresponding to a maximum deceleration of 22 G's and seat belt (4 kN load limiting system) and airbag as restraint system

Instrument the dummy with tri-axial accelerometer at the upper and lower part of the sternum. Filter all the response data according to the requirements of SAE Recommended Practice J211

Upper sternum and lower sternum resultant acceleration response should be within the corridors provided.

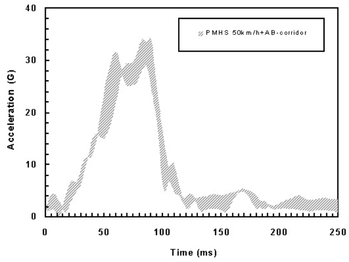

Figure 25. Upper sternum resultant acceleration vs. time

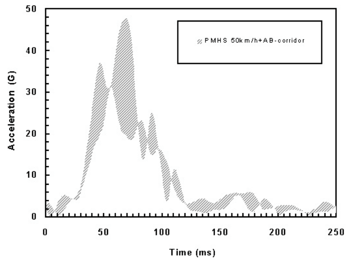

Figure 26. Lower sternum resultant acceleration vs. time

_Thorax test 4 is a frontal impact sled tests at 30 km/h with a sled deceleration of 15 G's, close the deceleration used at the University of Heidelberg (Kallieris, 2001) and only set belt as restraint system (4 kN force limiting system).

Instrument the dummy with tri-axial accelerometer at the upper and lower part of the sternum. Filter all the response data according to the requirements of SAE Recommended Practice J211

Upper sternum and lower sternum resultant acceleration response should be within the corridors provided.)

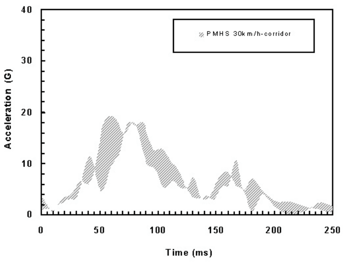

Figure 27. Upper sternum resultant acceleration vs. time

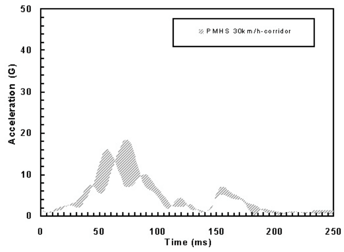

Figure 28. Lower sternum resultant acceleration vs. time

Abdomen

_Abdomen test 1 is a frontal impact test defined for the upper abdomen. The impact face of the pendulum consists of a rigid steering wheel, mounted with an angle of 45º. The effective mass of the pendulum should be 18 kg, impact speed 8.0 m/s. The dummy needs to be seated upright, with no back support and its legs and the arms raised. The impact point should be located at the level of the L2 vertebra.

Two uniaxial accelerometers rigidly mounted on the impactor with its axis of measurement collinear with the impact direction. The dummy will be instrumented with T12 triaxial accelerometer. The displacement and velocity of the spine and pendulum for the principal direction will be determined using the concept of a moving frame. Filter all response data according to the requirements of SAE Recommended Practice J211.

The upper abdomen force versus penetration response has to be within the well-known corridor provided by Nusholtz et al. (1994).

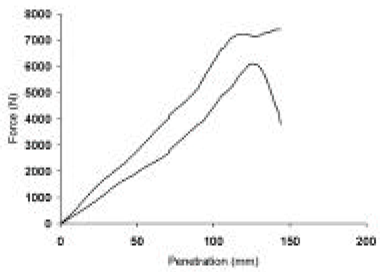

Figure 29. Force vs. Penetration

_Abdomen test 2 is a frontal impact test defined for the lower abdomen. The impact face of the pendulum consists of a rigid bar with above defined dimensions rigidly mounted on the pendulum. The effective mass of the pendulum should be 32 kg, impact speed 6.1 m/s for low impact and 10.4 m/s for high impact speed. The dummy needs to be position in an upright sitting position, with no back support and its legs and the arms raised. Impact point is located at the level of the L3 vertebra.

Two uni-axial accelerometers rigidly mounted on the impactor with its axis of measurement collinear with the impact direction. The dummy will be instrumented with an uni-axial accelerometer rigidly mounted in the rearward part of the spine and its measuring axis collinear with impact direction. Filter all response data according to the requirements of SAE Recommended Practice J211.

The lower abdomen force versus penetration response has to be within the well-know corridor provided by (Cavanaugh et al., 1986).

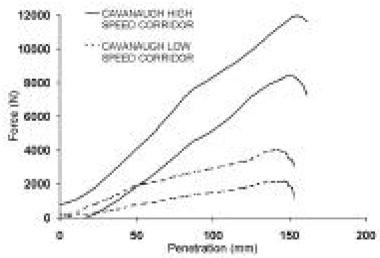

Figure 30. Force vs. penetration

_Abdomen test 3 is a seatbelt-loading test. The dummy needs to be positioned upright with the legs outstretched, and will be loaded about the mid abdominal region (approximately level of the umbilicus).. The device to achieve seatbelt loading should provide a peak-loading rate of approximately 3 m/s and an penetration speed-time history.

The load versus penetration response should be within the corridor provided in Figure 31.

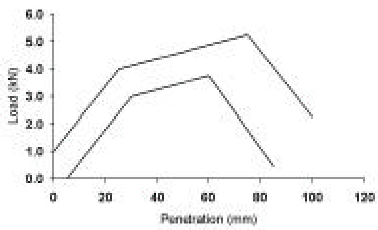

Figure 31. Load vs. Penetration for (Seatbelt Loading Condition)

_Abdomen test 4 is a surrogate airbag-loading test. The device used includes a pneumatic firing mechanism that accelerates a lightweight (approximately 1-kg) aluminum impactor constructed of welded thin-wall tubing. Face of the impactor is the sidewall of a 7.6 cm diameter tube that is 20 cm long. Impact speed 13 m/s.

The load versus penetration response should be within the corridor provided in Figure 32.

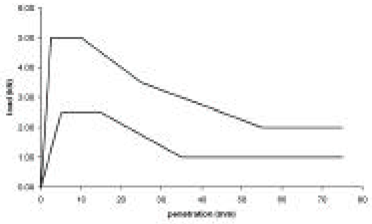

Figure 32. Load vs. Penetration (Surrogate Airbag Loading Condition)

Femur/Knee

_Femur/knee test 1 is a frontal impact test using a rigid flat face impactor (mass 5 kg, diameter 75 mm). The dummy needs to be seated on a rigid surface in front of the pendulum. The pelvis and the lower torso free to translate rearward such that no loading will occur on the pelvis due to the back support. Impact velocity 2.8 and 4.0 m/s.

Instrument the pendulum with one uni-axial accelerometers. Instrument the femur of the dummy with a load cell and four tri-axial accelerometers.

The knee impact versus time response, results femur acceleration versus time response and the iliac crest versus time response for the impact velocity of 2.8 m/s conditions should be within the corridors provided in Figure 33-37.

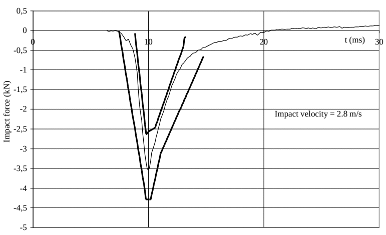

Figure 33. Knee impact force vs. time

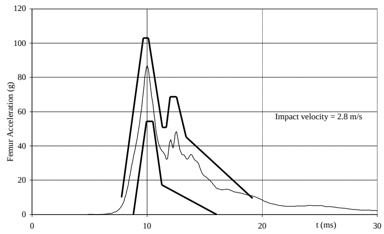

Figure 34. Femur acceleration vs. time

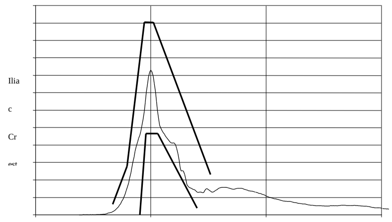

Figure 35. Iliac crest acceleration vs. time

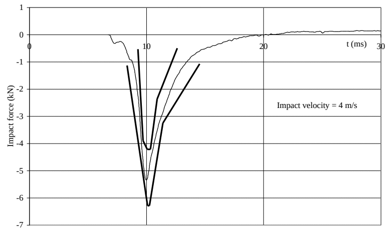

Figure 36. Knee impact force vs. time (4.0 m/s)

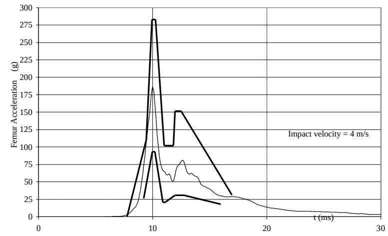

Figure 37. Femur Acceleration vs. time (4.0 m/s)

Lower Leg

_Lower leg test 1 is a pendulum impact to the toe, impact speed 6 m/s. The impactor comprises of a horizontal cylinder (mass 1.25, diameter 50 mm) and a lightweight support arm. The test procedure is based on the EEVC 'Tibia and Foot Certification The lower leg is attached via the knee clevis to a rigid back-plate. The foot is orientated such that the second metatarsal pointed vertically upwards, with the ankle maintained at 90° (neutral position). The line joining the knee clevis joint and the ankle is horizontal and the sole of the foot is vertical. The foot is fitted with a 3 mm thick PVC to represent the sole of a shoe. Laterally the center of the cylinder is aligned with the axis of the second metatarsal.

Data are acquired at a sampling frequency of 20 kHz and filtered as recommended in SAE J211/1. The lower leg is instrumented with a lower and upper tibial load cell. The impactor is instrumented with a single axis accelerometer.

For toe impacts at 6 m/s: the pendulum acceleration, the tibial force and the tibial bending moment should be within the corridors defined.

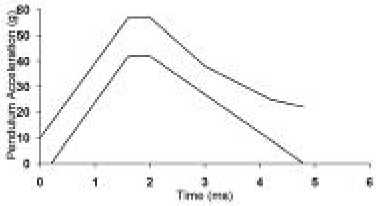

Figure 38. Pendulum Acceleration vs. Time (Toe Impact)

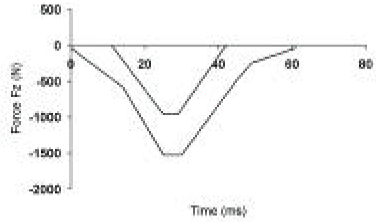

Figure 39. Tibial Force vs. Time (Toe Impact)

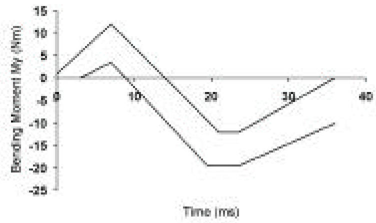

Figure 40. Bending moment vs. time (Toe Impact)

_Lower leg test 2 is a pendulum test to the heel, impact speed 4 m/s. The impactor comprises of a horizontal cylinder (mass 1.25 kg, diameter 50 mm) and a lightweight support arm. It is supported by an adjustable scaffolding structure. The test procedure and the test set-up are equal to Lower Leg Test 1. Only the position and the speed of the impactor has been changed. The EEVC foot certification procedure defines a height of 62 mm for heel impacts for a foot length of 265 mm. The height at which the cylinder impacted the dummy foot, will be scaled according to overall dummy foot length. Laterally the center of the cylinder is aligned with the axis of the second metatarsal.

Data are acquired at a sampling frequency of 20 kHz and filtered as recommended in SAE J211/1. The lower leg is instrumented with a lower and upper tibial load cell. The impactor is instrumented with a single axis accelerometer.

For heel impacts at 4 m/s: the pendulum acceleration and the tibial force and pendulum force should be within the corridors defined.

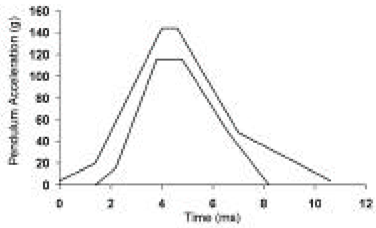

Figure 41. Pendulum acceleration vs time (Heel Impact)

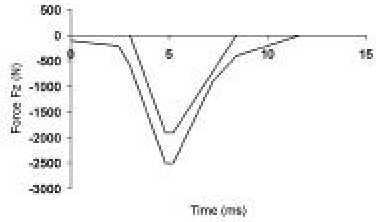

Figure 42. Tibial Force vs. Time (Heel Impact)