SAE International

SURFACE VEHICLE INFORMATION REPORT

J1460 FEB2013

Issued 1985-03
Stabilized 2013-02

Superseding J1460 MAR1985

Human Mechanical Response Characteristics

RATIONALE

The members of the SAE Human Biomechanics and Simulations Standards Steering Committee have reviewed J1460 and made a conscious decision to stabilize this Information Report. J1460/1 contains updated information on the impact responses of the abdomen and J1460/2 contains updated information on the responses of the neck to inertial loading of the head, Additional research into human impact responses to all body regions has been conducted since this Information Report was last revised. The design of the Hybrid III family of dummies was developed based on the data included in J1460. This Information Report has historical value.

STABILIZED NOTICE

This document has been declared "Stabilized" by the SAE Human Biomechanics and Simulations Standards Steering Committee and will no longer be subjected to periodic reviews for currency. Users are responsible for verifying references and continued suitability of technical requirements. Newer technology may exist.

1. Purpose—The purpose of this report is to provide a first-generation version of a standardized SAE document to define human mechanical response characteristics. It is intended for use by anthropomorphic test dummy designers and analytical modelers who need quantitative definitions of human mechanical behavior. This report describes how the human being responds to mechanical stimuli. It does not discuss criteria for assessing human injury potential. Human injury criteria are the subject of a separate SAE Information Report entitled, HS J885 APR80, "Human Tolerance to Impact Conditions as Related to Motor Vehicle Design."

2. Scope—While this report does not include a discussion of all of the available data defining human response or address all body areas, for those areas addressed it does utilize references generally judged by those in the field to be practical and meaningful guidelines for the development of human surrogates. This report is intended to be a "living" document that will be updated periodically.

A number of problems need to be addressed in defining human impact response characteristics. There is the problem of human response variability from subject to subject in volunteer tests. There is the problem of extrapolating such volunteer data which are obtained at low impact severities to higher impact severities using human cadaver response data obtained at injurious levels of impacts. Live animal experiments have been conducted over the years in an attempt to define human impact response and tolerance. The problem with using animal response data is the lack of geometric scaling techniques needed to interpret the data relative to the human size and shape. The last problem area is that the human form is not of unique size and proportion. There are significant geometric differences between and among adults and children and between males and females. The available literature dictates that treatment of this topic be constrained essentially to guidelines for average adult male responses, with the exception that thoracic response has also been quantified for 5th precentile female and 95th percentile male adults.

The scope of this first-generation report will be limited to discussions of the response characteristics of the head, face, neck, and thorax. While there are many additional regions of the body for which mechanical response characteristics have been quantified to some degree, only these four categories are addressed in this first-generation document. This limitation facilitated the issuance of this document. Moreover, these areas are thought to be of primary interest to expected readers. It is intended that any new data enabling more comprehensive treatments of the above topics be utilized in future updates of the report. Furthermore, additional body areas such as the abdomen and knee-thigh-hip complex are to be addressed.

3. Impact Response of the Human Head

3.1 Introduction—The impact response of the head will be described in terms of its acceleration responses to prescribed impact conditions and/or the interaction forces that occur between the head and the contact surface for the prescribed impacts. Both of these impact response parameters are dependent on the head's mass and mass distribution, the dynamic force-deformation characteristics of the skull and the soft flesh covering the skull, and the location and direction of the impact force. Determination of the head's acceleration response is particularly important since most head injury criteria are based on measured head accelerations.

3.2 Inertial Properties—Several sources of data for inertial properties of the human head have been reviewed (Refs. 1, 3, 4, 5, and 6). The recent study by Reynolds et al. (Ref. 1) is judged to be the singlemost appropriate information source for this report. While their subjects averaged somewhat below the 50th percentile in body mass, the data can be adjusted, and it will be shown that the adjusted results compare well with those of other studies.

Reynolds et al. evaluated six adult male cadavers having a mean body mass of 65.2 kg. They determined that the head mass is 6.1% of the total body mass. Furthermore, average mass moments of inertia for axes through the center-of-gravity of the head were as follows:

Axis	Moment of Inertia (kg · m^2)
Lateral (left-right)	$16.4(10)^{-3}$
Anterior-Posterior (front-to-back)	$17.4(10)^{-3}$
Superior-Inferior (top-to-bottom)	$20.3(10)^{-3}$

TABLE 1—HEAD INERTIAL DATA FOR 50TH PERCENTILE ADULT MALE

Head Mass—4.69 kg Head Moments of Inertia (about axes through c.g.)	
Axis	**Moment of Inertia (kg · m^2)**
Lateral (left-right)	$21.3(10)^{-3}$
Anterior-Posterior (front-to-back)	$22.6(10)^{-3}$
Superior-Inferior (top-to-bottom)	$26.3(10)^{-3}$

Exact definitions of the coordinate axes and head-neck sectioning plane utilized for specifying the above inertial properties may be obtained from Ref. 1.

The 1971–74 Health and Nutrition Examination Survey (Ref. 2)[1] indicates that the U. S. 50th percentile adult male has a nude body mass of 76.9 kg, an increase of 2.5 kg from the 1960–62 value (Ref. 3) of 74.4 kg.[2] Since head mass can be taken to be 6.1 percent of body mass, the six cadavers evaluated by Reynolds et al. have an estimated mean head mass of 3.98 kg, whereas that of the current 50th percentile adult male is 4.69 kg; an apparent difference of 0.71 kg.

The above-quoted inertial data of Reynolds et al. must be adjusted upward to account for the apparent mass disparity of 0.71 kg if values associated with a 50th percentile body mass are desired. The adjustments are derived in Appendix A, and the results are presented in Table 1. Table 1 compares favorably with the review of the literature by Hubbard and McLeod (Ref. 4), who recommended a head mass of 4.55 kg (3.0 percent lower) and a lateral-axis moment of inertia of $23.3(10)^{-3}$ kg · m^2 (9.4 percent higher) for the 50th percentile adult male. Furthermore, Beier (Ref. 5), from tests on 21 cadavers having a mean body mass of 74.0 kg (3.8 percent lower), has obtained a mean head mass of 4.3 kg (8.3 percent lower) and a lateral-axis moment of inertia of $22.1(10)^{-3}$ kg · m^2 (3.8 percent higher). Finally, Mertz (Ref. 6) estimated his volunteer to have had a head mass of 4.90 kg (4.5% higher) and a lateral-axis moment of inertia of $22.6(10)^{-3}$ kg · m^2 (6.1% higher).

3.3 Head Impact Response—Two sets of data are available from Hodgson and Thomas (Refs. 7 and 8) who performed a series of embalmed human cadaver drop tests during which the head impacted a flat, rigid surface mounted on a force transducer. While some of these tests involved the use of decapitated heads, others involved complete cadavers that were strapped to a light-weight pallet which was supported by a pivot point at the feet. The pallet was released at various angles from the horizontal and allowed to drop freely under the influence of gravity, causing the head to impact the rigid surface. An equivalent free-fall drop height was computed from the head velocities[3] measured just prior to impact. The cadaver responses were evaluated for three impact locations: frontal, lateral, and rear (i.e., frontal, parietal, and occipital.) While separate response definitions for each of the three individual impact directions are of interest for a comprehensive definition of head response, the paucity of data and their scatter preclude such detail. Qualitatively, the available data did not indicate any dramatic dependence of response on impact location. A pragmatic decision was made that a response definition based on the pooled data is preferable to no definition at all. The results of these tests are summarized in Fig. 1 which illustrates the responses in terms of peak force[4] and peak acceleration as a function of head drop height. Data documenting the temporal characteristics of the pulses are not available. Due to the scatter[5] in the data and the quantity of data points, it

[1] This provides the most current data for the USA population.

[2] Approximate clothing masses of 0.2 kg and 0.9 kg have been subtracted from the 1971–74 and 1960–62 data, respectively, as recommended in the respective publications for approximating the nude body mass.

[3] Velocities reported in Refs. 7 and 8 for pallet drop tests were determined to be erroneous. In cooperation with this Task Force, Dr. Hodgson made additional velocity measurements using a Hybrid III dummy. (See attachment 4.3 of the September 15, 1982 minutes of the SAE Human Biomechanics and Simulation Subcommittee.) This information was found to be consistent with analytical predictions, and was used to adjust the originally-reported values. Equivalent free-fall drop heights were then computed from these velocities.

[4] In consultation with Dr. Hodgson, the peak forces in Ref. 7 have been adjusted upward by 19% to account for an original data reduction procedure thought to have been erroneous.

[5] The spread in the data probably reflects the effects of 1) different head sizes and masses of the cadaver subjects, and 2) the changes in skin characteristics that occur when the same site on a cadaver's head is impacted more than once.

is not practical to plot each individual test result. Furthermore, attempts to fit meaningful regression curves proved to be futile. Instead, the results have been reduced to 4 response regions, A through D in Fig. 1. Regions A and B provide "windows" of peak force versus drop height, whereas regions C and D are peak acceleration versus drop height windows. Regions A and C are based on tests where skull fracture did not occur; there were 50 tests on 7 cadavers, with equivalent drop heights ranging from 76 to 188 mm. The windows represent mean values ± ½ standard deviation, both for drop height and for force and acceleration. Regions B and D are based on tests where fracture did occur; there were 15 tests on 15 cadavers, with equivalent drop heights ranging from 330 to 1060 mm. The window in each of these cases represents the linear regression line plus and minus approximately one-half standard deviation of force or acceleration. The alignment of the head relative to the force transducer, the dynamic characteristics of the force transducer, and the channel class of the instrumentation utilized in these experiments are described in Appendix B. A detailed documentation of the analyses leading to the results presented in Fig. 1 was in preparation at the time this document was written. It is intended that such documentation be included in an updated version.

In the biomechanical study summarized above, force data and acceleration data are presented. Specification of performance in terms of force has the practical advantage that anthropomorphic dummy heads can be tested without the necessity of instrumentation mounted on the head. A disadvantage, however, stems from the fact that dummy heads are normally instrumented with accelerometers located at the center-of-gravity, these accelerometers being used to assess the severity of a blow from the standpoint of injury. Therefore fidelity of head center-of-gravity acceleration response is important. Unfortunately, all cadaver head accelerations have been monitored by affixing accelerometers to the head externally. Thus, cadaver and dummy head accelerations may not be directly comparable. The acceleration data of Hodgson and Thomas presented in this report involve laterally placed accelerometers for frontal and occipital impacts and occipitally placed accelerometers for lateral impacts.

It is recommended that in future head impact response studies, the acceleration of the center-of-gravity of the head be reported. This will require instrumenting the head with an appropriate array of accelerometers such that the acceleration of the center-of-gravity can be calculated. Also, it is recommended that only a single impact be delivered to a given impact site for a given cadaver head. Multiple impacts damage the underlying soft tissue, resulting in progressively increasing peak force and acceleration responses which confound interpretation of the data.

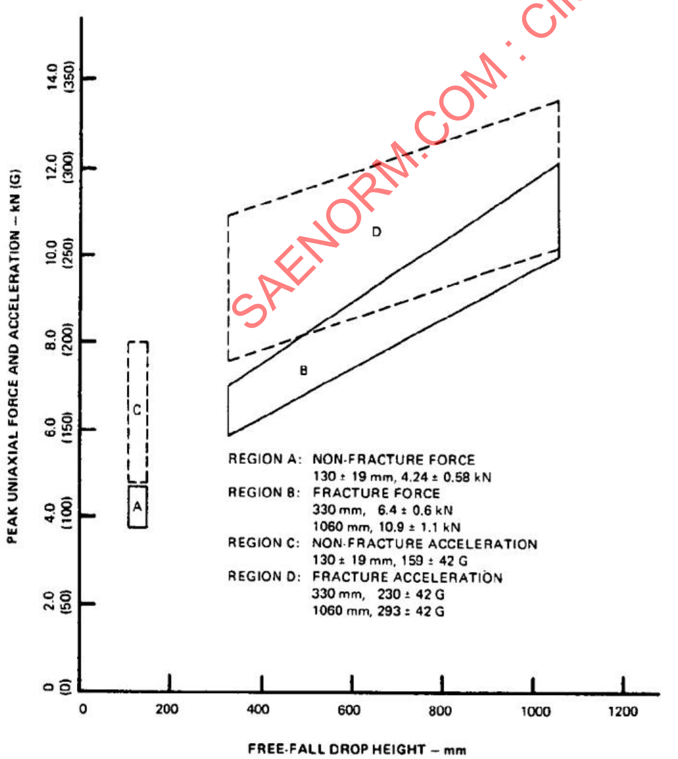

FIG. 1—RESPONSE OF THE HEAD FOR IMPACT TO A HARD, FLAT RIGID SURFACE

3.4 Facial Impact Response—The influence of facial structures on head impact response has received little attention. The primary concern in head injury research has been to prevent serious brain injury. Injury to the face, while presenting the problem of possible disfigurement, has not been considered as serious in nature as that of brain injury. In addition to producing injury to the facial structures through damage to the soft and hard tissues of the face, these same failure mechanisms can have an effect on the loading that the brain will receive.

The mechanical response of the face depends upon the properties of the overlying soft tissues and the 14 facial bones. While biomechanical studies have been conducted for the zygoma (cheek bone), the mandible (lower jaw bone), and the maxilla (upper jaw bone), the primary thrust was to attempt to establish tolerable force values. As such, the majority of the experiments were not structured to provide impact response directly in terms of force-displacement characteristics.

A series of sub-fracture experiments on a single, instrumented, embalmed cadaver were conducted by Hodgson et al. (Ref. 9). Static loads were applied to the zygoma or to the zygomatic arch and the resulting deflections were recorded for both loading sites with loads up to 445 N for the arch and 890 N for the zygoma. The resulting spring rates were found to be 1734 N/mm for the zygomatic arch and 4939 N/mm for the zygoma. Generally similar force-strain relationships were found for both the static and dynamic loads.

4. Dynamic Response of the Human Neck

4.1 Introduction—In the automotive crash environment, direct impact to the neck is relatively uncommon. Accordingly, this discussion of neck response will be limited to the effects of inertial loading due to head acceleration. Such a response characteristic is important in view of the fact that it governs the head trajectory during a vehicle collision. Faithful simulation of neck response and head and neck geometry is of obvious concern in the prediction of the nature of human head contact with a vehicle interior based on crash tests using a human surrogate.

While several studies of neck response have been conducted during the past decade, including Ewing et al. (Refs. 10 and 11), Mertz et al. (Ref. 12), Foust et al. (Ref. 13), and Patrick and Chou (Ref. 14), only the data from Mertz et al. appear to be appropriate for use in this document. The other studies have produced data which were either not fully analyzed and condensed into a concise performance specification, not produced under test conditions representative of a crash environment, or not sufficient to establish a definitive trend. By excluding the studies mentioned above, this report is necessarily restricted to a definition of neck response in forward flexion and extension. There is clearly a need for additional response data; for example, performance specifications for lateral flexion. It is hoped that subsequent updates of this report will contain more complete information as it becomes available from analyses of existing data or new data.

4.2 Performance Guidelines for Forward Flexion and Extension—The guidelines described below are based on the response of the human head-neck-torso system, and assume the use of a geometrically and inertially accurate representation of the human head. The neck, when subjected to the dynamic tests specified in paragraph 4.3, should have the following response:

1. The moment-angle relationships during neck loading[6] should lie within the corridors shown in Figs. 2 and 3 for flexion and extension, respectively.

2. The quotient of the area between the loading and unloading curves and the area under the loading curve should be 0.3 to 0.5 for flexion and 0.4 to 0.6 for extension.

4.3 Test Conditions—For purposes of these tests, the torso can be that of a dummy tightly strapped to a rigid seat assembly, or it can be simulated by a rigid structure that does not rotate during the test. The longitudinal axis of the neck at the neck-torso junction should be oriented at an angle of 25 deg rearward with respect to the vertical direction. The anterior-posterior axis of the head should be horizontal. This initial head position should be maintained until the onset of the prescribed loading.

The kinematic requirements for the base of the neck in these tests are as follows:

FLEXION—The base of the neck should experience a horizontal change in velocity of 9.75 ± 1.5 m/s producing flexion of the neck. The maximum moment produced at the occipital condyles should be at least 163 N · m. Foster et al. (Ref. 15) successfully met these requirements by utilizing an essentially square-wave deceleration pulse from 10.0 m/s with a stopping distance of 254 mm.

EXTENSION—The base of the neck should experience a horizontal change in velocity of 6.71 ± 1.5 m/s producing extension of the neck.

[6] That portion of the event where the moment is increasing.

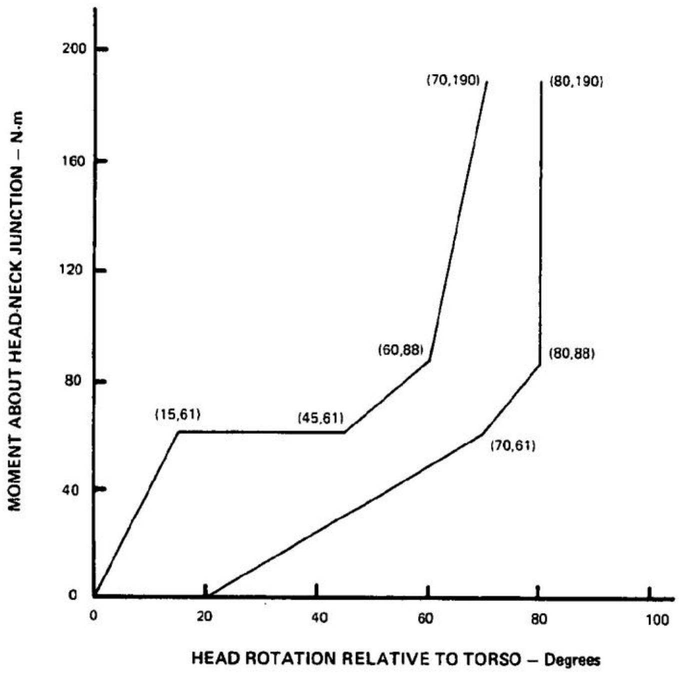

FIG. 2—NECK FLEXION (FORWARD BENDING) LOADING CORRIDOR

The maximum moment produced at the occipital condyles should be at least 54 N · m. Foster et al. (Ref. 15) successfully met these conditions utilizing an essentially square-wave deceleration pulse from a velocity of 7.2 m/s and a stopping distance of 432 mm.

5. Dynamic Response of the Human Thorax

5.1 Introduction—There is a large quantity of data that can be used to define the response of the human thorax to mechanical loading. The chest has been loaded by various means from the most simple, quasi-static system to complex interactive dynamic loading, such as an air bag deployment. Subjects in such tests have ranged from various animals to human cadavers and volunteers.

Although these studies have made substantial contributions to the understanding of thoracic response and injury, they are for the most part too general to be of use for explicit definition of the mechanical response of the chest.

Two of these studies were conducted with a high level of control over the impact environment. Human unembalmed cadavers were used as test subjects in these studies. Data from one basic study have been reported by Kroell et al. (Refs. 16 and 17) and Nahum et al. (Refs. 18 and 19), and summarized by Kroell (Ref. 20). A second, similar study was conducted by Stalnaker et al. (Ref. 21).

The response definition recommendations herein are based on the data from references 17–22. It is important to point out that in these studies the load data were adjusted upward based on limited volunteer testing to account for the lack of muscle tone in the cadaver subjects. The actual recommendations come from an analysis of these studies done by Neathery (Ref. 25) and are congruent with those of Stalnaker (Ref. 21). The above studies are restrictive in that both the loading environment and the resulting response are simplistic, consisting of well-controlled blunt frontal impact and response in the anterior-posterior (front-to-back) direction only. Consequently, the limitations and cautions that accompany the recommendations should be taken seriously.

Since the time of publication of the above-mentioned references, several events have occurred that make the recommendations herein appropriate. There was an early acceptance of the results of these data by those attempting to develop crash test dummies with improved biofidelity. Typical efforts are those by Lobdell et al. (Ref. 22), Neathery and Lobdell (Ref. 23), McElhaney et al. (Ref. 24), and Foster et al. (Ref. 15). Reports on the development projects have led to an international awareness of the test procedure and the general results. These development projects have led investigators to thoroughly examine and analyze these data, resulting in a codification and scaling of results to different sizes of the population (Ref. 25). The final step, which has led to a universal awareness of and capability for the test conditions in the automotive safety test community, was the inclusion of the test procedure (without humanlike response requirements) in the specifications for certifying the Part 572 dummy in Federal Safety Standards (Ref. 26).

More recent studies by Melvin et al. (Ref. 27) and Tarriere et al. (Ref. 28) have been pursued to define thoracic response to lateral and oblique impacts. The results of these studies are under review by the scientific community at the time of preparation of this document. It is intended that appropriate information be included in an updated version.

5.2 Mechanical Response Guidelines for Frontal Impact—Biomechanical response guidelines for the thorax may be specified using the test configuration developed for collecting the human data base. These guidelines may be applied to anthropomorphic dummies, computer simulations, and other analogs of the human. For dummy evaluation, the test configuration is very nearly a replication of the original environment and is also the same as that required by NHTSA for Part 572 dummy certification (Ref. 26). For other than dummy analogs, test conditions must be developed by the user while holding to the parameters specified in this document.

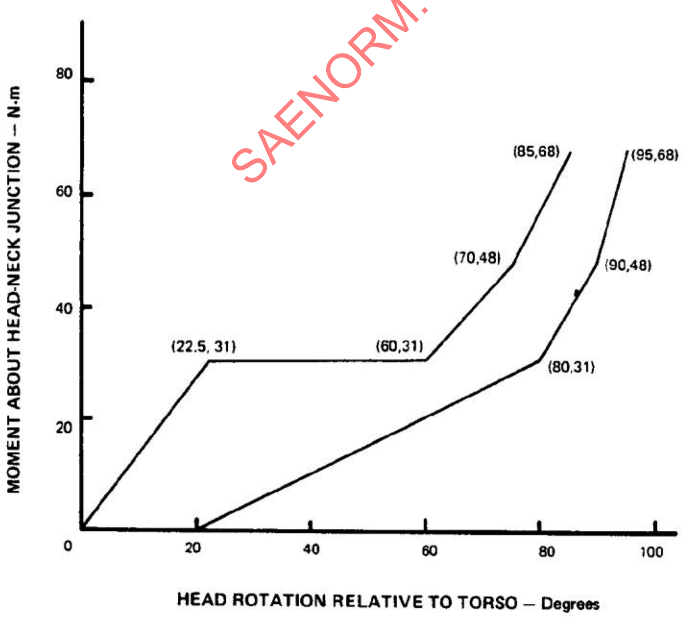

FIG. 3—NECK EXTENSION (REARWARD BENDING) LOADING CORRIDOR

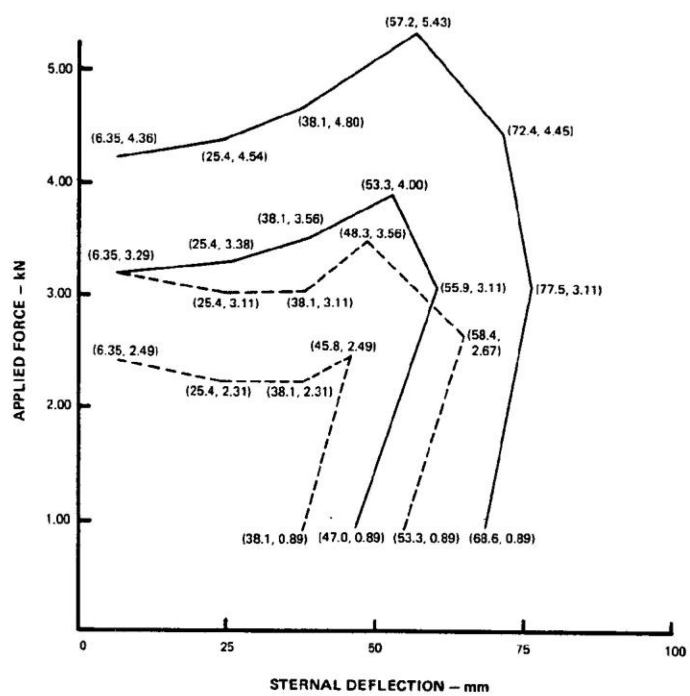

FIG. 4—THORACIC RESPONSE CORRIDORS FOR 5TH PERCENTILE FEMALE

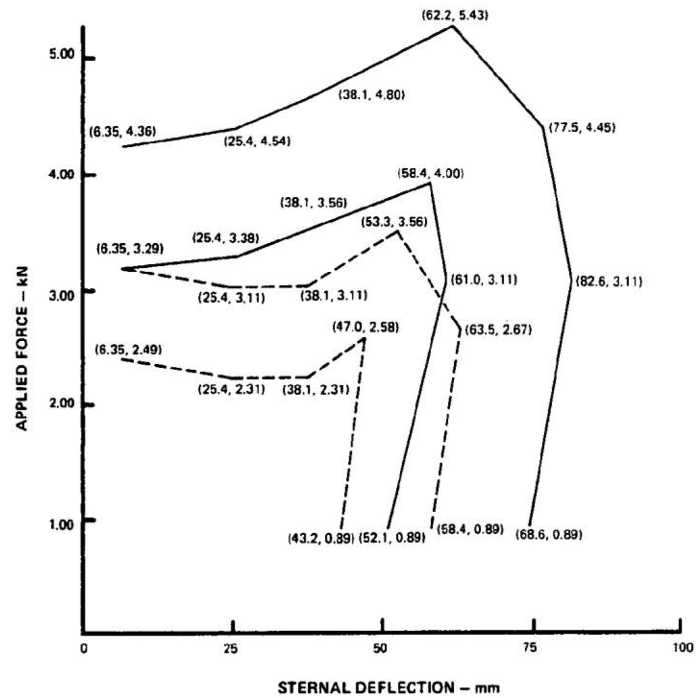

FIG. 5—THORACIC RESPONSE CORRIDORS FOR 50TH PERCENTILE MALE

Satisfying these performance guidelines is necessary to assure biomechanical fidelity of the chest to blunt-frontal, midsagittal impacts. Meeting these guidelines does not assure biofidelity of response for other loading conditions. Specifically, it has not been demonstrated that these performance criteria are appropriate for frontal loading due to belts or air bags, or for time durations outside of the regime covered by the test conditions. It is, however, assumed that in satisfying criteria at two dynamic conditions, there is some potential for extrapolation of the results. The criteria do not address oblique and/or lateral loading of the thorax.

When struck by an impactor as described below, the impact force plotted against the sternal deflection (relative to the spine) should fall within the appropriate corridor shown in Figs. 4, 5, and 6 depending upon

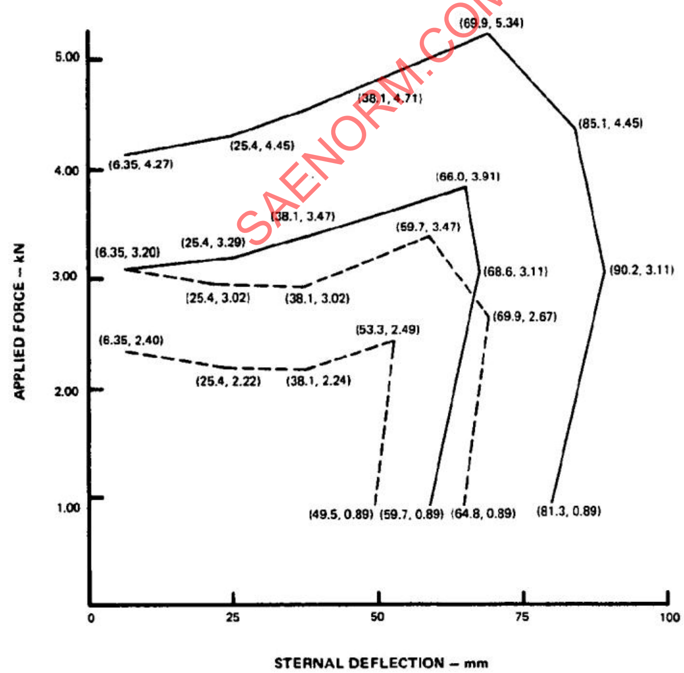

FIG. 6—THORACIC RESPONSE CORRIDORS FOR 95TH PERCENTILE MALE

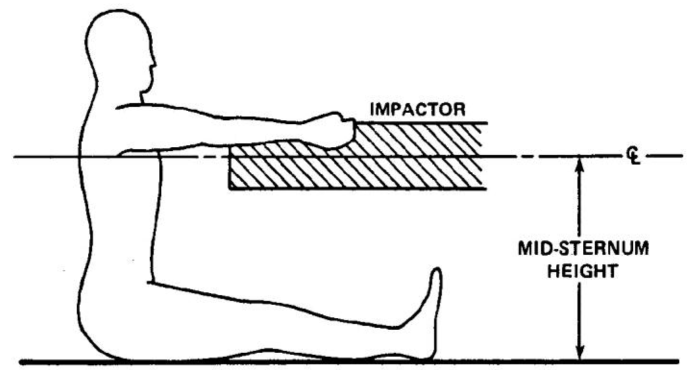

FIG. 7—TYPICAL TEST SET-UP FOR THORACIC RESPONSE TEST

subject size and probe velocity. The figures are for 5th percentile female, and 50th and 95th percentile male sizes, respectively. Both the 4.27 m/s and the 6.71 m/s corridors should be satisfied.

5.3 Test Conditions—A typical test set-up is shown in Fig. 7. Details are as follows:

1. Seat the subject in the upright position on a flat, horizontal surface without back support and extend the arms and legs horizontally forward and parallel to the midsagittal plane.
2. Adjust the subject so that the surface of the thorax in proximity of the projected longitudinal centerline of the impactor is vertical.
3. Adjust the longitudinal centerline of the impactor to a vertical height equivalent to mid-sternum.
4. Orient the impactor so that at impact its longitudinal centerline is horizontal and in the midsagittal plane of the subject.
5. Impact the thorax with the impactor moving horizontally at the specified velocity. The impact velocities should be 4.27 ± 0.1 m/s for the lower corridor and 6.71 ± 0.1 m/s for the upper corridor.
6. Record the time histories of the force applied to the chest by the impactor and the sternal deflection of the subject. Cross-plot these traces to obtain force vs. deflection. Processed data should conform to the requirements of SAE Channel Class 180.

5.4 Impactor—The impactor should have a cylindrical end 152 ± 1 mm in diameter, a flat face perpendicular to the longitudinal axis, and an edge radius of 12.7 ± 0.5 mm. Its mass, including all instrumentation, should be 23.4 ± 0.1 kg.

6. References

1. Reynolds, H. M.; Clauser, C. E.; McConville, J.; Chandler, R.; and Young, J. W., "Mass Distribution Properties of the Male Cadaver," SAE Paper No. 750424, 1975.
2. Abraham, S.; Johnson, C. L.; and Najjar, M. F., "Height and Weight of Adults 18–74 Years of Age in the United States," *Advancedata*, No. 3, November 19, 1976, from Vital and Health Statistics of the National Center for Health Statistics, U. S. Department of Health, Education, and Welfare, Washington, DC.
3. Stoudt, H. W.; Damon, A.; and McFarland, R., "Weight, Height and Selected Body Dimensions of Adults—United States, 1960–1962," Public Health Service Publication No. 1000, Series 11, No. 8, June, 1965, from Vital and Health Statistics of the National Center for Health Statistics, U. S. Department of Health, Education, and Welfare, Washington, DC.
4. Hubbard, R. P. and McLeod, D. G., "Definition and Development of A Crash Dummy Head," SAE Paper No. 741193, *Proceedings 18th Stapp Car Crash Conference*, 1974.
5. Beier, G.; Schuller, E.; Schuck, M.; Ewing, C.; Becker, E.; and Thomas, D., "Center of Gravity and Moments of Inertia of Human Head." Proceedings of the Fifth International Research Committee on Biokinetics of Impact Conference, Birmingham (England), September, 1980.
6. Mertz, H. J., "The Kinematics and Kinetics of Whiplash," Ph.D. Dissertation, College of Engineering, Wayne State University, Detroit, MI, 1967.
7. Hodgson, V. R. and Thomas, L. M., "Breaking Strength of the Human Skull vs. Impact Surface Curvature," Final Report, DOT Contract No. HS-146-2-230, September, 1973.
8. Hodgson, V. R. and Thomas, L. M., "Head Impact Response," Vehicle Research Institute Report VRI 7.2, Society of Automotive Engineers, Inc., 1975.
9. Hodgson, V. R.; Nakamura, G. S.; and Talwalker, R. K., "Response of the Facial Structure to Impact." *Proc. 8th Stapp Car Crash Conf.*, (1964), Wayne State Univ. Press, pp. 229–250, 1966.