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THE ENGINEERING DESIGN, DEVELOPMENT, TESTING, AND EVALUATION OF AN ADVANCED ANTHROPOMORPHIC TEST DEVICE

PHASE 1: CONCEPT DEFINITION

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16. Abstract

This report summarizes the results of Phase 1, Concept Definition, of the AATD program and identifies the reasons such a new test device is needed. The following areas are reviewed: injury priority from accident data; biomechanical impact response and injury; instrumentation, data processing, and certification procedures; current dummy design and use; anthropometric data for biomechanical response simulation; and AATD technical characteristics, design concepts, and trauma assessment.

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OVERVIEW

Phase 1 of the AATD program has shown that there is a need to develop a new anthropomorphic test device that

- Has humanlike response in frontal through lateral as well as rollover impact directions,
- Interacts realistically with restraint systems and the vehicle interior,
- Is capable of making numerous meaningful measurements for injury assessment,
- Is durable and easily maintained, and
- Can be certified without disassembly.

New Data are Available. Because of the necessary time lag between design and production, current dummies are not based on currently available data. A greatly improved and expanded biomechanical data base has been developed in Phase 1. These data form the basis for defining impact response characteristics necessary to ensure that the AATD will perform in a humanlike manner.

A Clean-Sheet-of-Paper Approach is Needed. Past dummy development has been an evolutionary process. Such a process does not allow for significant departures from traditional design solutions, because each change must take into account its effect on other dummy components that remain unchanged. Although the best aspects of past dummy designs will be drawn from, past designs will not limit the innovative approaches open to AATD design and development. Only by taking a fresh look at all dummy design concepts can the desired results be achieved.

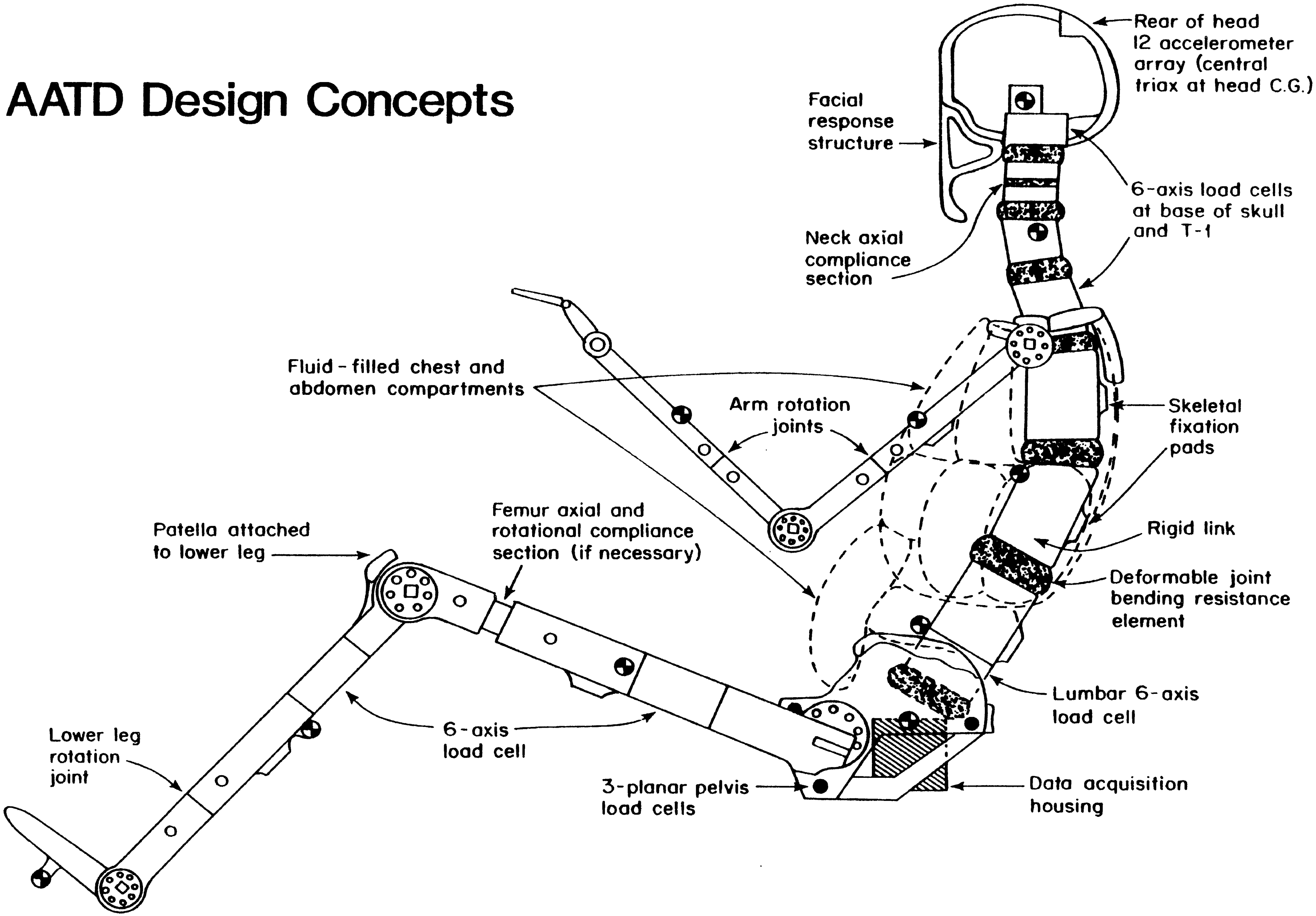
The Technology Exists. The technology is presently available to develop an AATD with greatly improved impact biofidelity, measurement capability, durability, maintainability, and certification ease. Design concepts, technical characteristics, and measurement requirements, schematically presented on the next page, have been developed in Phase 1 for all body regions. These will be further developed and result in a complete prototype AATD by the end of Phase 2.

The new dummy will have significant advantages over current ATDs.

- Its shape in the seated configuration will be based on actual mid-sized male seated anthropometry, which will contribute to more realistic interaction with the vehicle seat and other components.

- It will have biofidelity in frontal, oblique, lateral, and vertical impacts, to be achieved through such features as a deformable face, a multidirectional neck and chest, and a flexible thoracic spine. Thus only one dummy will be needed for crash testing purposes, and it will give realistic trajectory, contact-point, and loading results.

AATD Design Concepts



- Its humanlike response will allow many more meaningful engineering measurements to be made, which will in turn greatly enhance injury assessment capability.

- Materials will be durable, and designs will be rugged and repeatable but cost-effective to manufacture. Materials will be selected to minimize the sensitivity of the AATD to temperature variations.

- The certification process will be more efficient and repeatable between laboratories, to be achieved through the use of a whole-body certification fixture with a dedicated data processing system.

INJURY PRIORITY ANALYSIS (Task A)

This analysis of the National Accident Sampling System (NASS) data for 1980 and 1981 places the cost of individual or aggregated groups of injuries in perspective relative to the cost to society for all AIS 2-6 injuries to automobile occupants. The NASS files were first augmented to incorporate an impairment factor that goes beyond the Abbreviated Injury Scale (AIS) by taking into account the consequences of an injury as determined by two panels of physicians as well as a percentage impairment based on American Medical Association guidelines. Then a factor was generated from an economic cost model to account for expected lifetime earnings had an individual person not been injured. The impairment factors for the actual NASS injuries were multiplied by the expected earnings factors for individual injured persons to create an Injury Priority Rating (IPR). These IPRs were then aggregated by body region, direction of force, and delta V and expressed in terms of a percentage of total IPR. Tables A-1 through A-3 show these distributions, which indicate the relative contribution of each grouping of injuries to the total societal cost of injuries to automobile occupants.

TABLE A-1

IPR DISTRIBUTION BY BODY REGION

Body Region	Distribution (%)
Head	44.6
Face	10.5
Neck	5.1
Shoulder ..	0.3
Chest	18.9
Back	1.6
Abdomen ..	7.5
Pelvis	1.1
Thigh	2.1
Knee	1.6
Lower Leg .	1.0
Ankle/Foot .	0.6
Lower Limb	0.0
Upper Arm	1.3
Elbow	0.5
Forearm ..	1.3
Wrist/Hand	0.4
Upper Limb	0.3
Whole Body	0.9
Unknown ..	0.2
TOTAL ...	100.0

TABLE A-2

IPR DISTRIBUTION BY DIRECTION OF FORCE

Direction of Force	Distribution (%)
1 o'clock	4.8
2 o'clock	9.8
3 o'clock	3.5
4 o'clock	0.0
5 o'clock	0.1
6 o'clock	0.7
7 o'clock	0.3
8 o'clock	1.2
9 o'clock	3.3
10 o'clock	7.9
11 o'clock	5.0
12 o'clock	36.9
Non-Horizontal	16.1
Unknown	10.4
TOTAL	100.0
N	2262

TABLE A-3

IPR DISTRIBUTION BY DELTA V

Delta V	Distribution (%)
1–5 mph ..	0.1
6–10 mph .	0.6
11–15 mph	2.0
16–20 mph	4.3
21–25 mph	7.0
26–30 mph	4.9
31–35 mph	2.6
36–40 mph	1.5
41–45 mph	1.5
46–50 mph	0.0
51–55 mph	2.7
> 55 mph	3.8
Unknown .	68.9
TOTAL ..	100.0
N	2262

The primary conclusions from these and further bivariate analyses are as follows:

1. The combination of the head, face, and neck body regions accounts for 60 percent of the IPR to passenger car occupants.

2. The combination of the chest, back, and abdomen body regions accounts for 28 percent of the IPR to passenger car occupants.

3. Over one-third of driver IPR occurs from collisions with a 12 o'clock direction of force. A fifth results from collisions with non-horizontal directions of force.

4. Oblique side collisions account for more IPR than direct side collisions. This applies both to drivers and to right-front passengers. Thus, 9 o'clock collisions account for 4.3 percent of driver IPR, but 10 and 11 o'clock collisions account for 11.9 percent. Similarly 3 o'clock collisions account for 9.4 percent of IPR to right-front passengers; 1 and 2 o'clock collisions account for 19.2 percent.

5. Using only known values of delta V, 84 percent of the driver IPR with a 12 o'clock direction of force results from severe crashes, i.e., those with a delta V greater than 20 mph. For right-front passengers, the figure is 97 percent. However, it should also be noted that, for cases with known delta V, 81 percent of driver IPR and 77 percent of right-front passenger IPR was attributable to crashes with a delta V of 45 mph or less.

6. Again using only cases with known delta V, 66 percent of driver IPR for injuries to the head, face, and neck results from severe crashes. For injuries to the chest, back, and abdomen the comparable figure is 81 percent; for injuries to the upper extremities, 45 percent; and for injuries to the lower extremities, 93 percent. Thus one might conclude that, for drivers, serious injuries to the upper extremities are the most easy to prevent because a higher proportion of them occur in less severe crashes. Next would come the combination of the head, face, and neck, followed by the combination of the chest, back, and abdomen, and last the lower extremities.

7. Comparison of IPR with the earlier Harm model indicated that the two models were in complete agreement in assigning relative priority to the directions of force in the 1980 and 1981 NASS data. When ranking body regions, however, the IPR model gives higher priority to the head, face, and neck, and correspondingly less prominence to the chest, abdomen, and extremities. This is because of the relatively severe long-term consequences of injury to the head, face, or neck.

Because of limitations in the data, it was not always possible to depict, to the extent that was desired, the crash environment in which the IPR to the various body regions was incurred. In particular, the high rates of missing delta V meant that analysis of crash severity was often not possible. Another concern is with the comparatively small number of occupants in the NASS files that sustain serious injuries. The 1980 and 1981 NASS files combined have only some 2,262 injuries of severity AIS-2 or greater to passenger car occupants.[1] These injuries are sustained by 1,262 occupants. There are a total of 15,378 passenger car occupants in the combined 1980 and 1981 files. Thus 92 percent of the occupants sustain no injuries, injuries of AIS 1, or injuries of unknown severity.

One solution to the shortage of cases for analysis is to incorporate additional years of NASS. It is hoped that the 1982 NASS data can be added to the existing data structure

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[1]This figure includes some injuries originally coded with an AIS of 7.

so that the analyses reported here can be run with additional confidence and perhaps be extended to include such issues as contact point in more detail. Another solution would be to revise the threshold for the inclusion of cases in the NASS system or to sample at higher rates cases in which injuries greater than AIS 1 are sustained. Such a revised sampling scheme could be combined with reducing the amount of investigation carried out on cases with no injuries or minor injuries.

Finally, the reader should bear in mind that the model used here is not completely satisfactory. In particular, it does not take into account the fact that a single person may have sustained more than one injury. It is hoped to pursue the development of a multi-injury model in the future.

REVIEW OF BIOMECHANICAL IMPACT RESPONSE AND INJURY IN THE AUTOMOTIVE ENVIRONMENT (Task B)

This review includes literature through 1984 and is divided into chapters covering the following body regions: head, spine, thorax, abdomen, pelvis, and lower extremities. Each chapter includes information on anatomy; clinical injury experience; biomechanical response to impact; and injury mechanisms, tolerance, and criteria from laboratory studies. Each chapter also contains its own reference list and thus can stand alone as a review of the literature on that region of the body. Summaries of each chapter follow.

HEAD

The head is considered the most critical part of the body to protect from injury because of the irreversible nature of injury to the brain. In the Injury Priority Analysis, head injury constitutes nearly 45% of the total IPR. Facial injury, however, accounts for an additional 10.5%. The costly facial injuries are primarily lacerations to younger occupants. While these injuries are not likely to be life-threatening, the impairment to the individual from facial nerve damage and/or facial disfigurement as well as the need for reconstructive surgery make such injuries relatively costly to society.

A variety of mechanisms have been postulated for mechanical damage to the brain from impacts to the head. They include: (1) direct brain contusion from skull deformation at the point of contact; (2) indirect brain contusion produced by negative pressure on the side opposite the impact; (3) brain contusion from movements of the brain against rough and irregular interior skull surfaces; (4) brain and spinal cord deformation in response to pressure gradients and motions relative to the skull, resulting in stress in the tissues; and (5) subdural hematoma from movement of the brain relative to its dural envelope, resulting in tears of connecting blood vessels. The latter three mechanisms have also been postulated for mechanical damage resulting from head motions due to indirect impact.

The data presently available for defining the response of the head to impact are limited to rigid impacts and are predominantly based on embalmed cadaver tests. The data are adequate to define general response specifications for rigid impacts to the front and side of the head, in terms of peak contact force over a range of impact velocities from 1 to 8 m/s. The corresponding acceleration response data are limited to an impact velocity range of 1 to 5 m/s.

There is a need for additional studies to define the impact response of the human head using unembalmed cadavers with rigid impact surfaces and current acceleration measurement techniques. A repeatable and reproducible method for producing padded impacts also needs to be developed to allow cadaver studies to be conducted for padded head impact response definition.

The parameters of head motion that have been associated with the production of brain injury are translational acceleration, rotational acceleration, and rotational velocity. Of these, most attention has been given to translational acceleration in terms of developing head injury criteria. For direct impacts to the head, the Wayne State Tolerance Curve and the Japan Head Tolerance Curve, both based on head translational acceleration, are in close agreement. Injury criteria that have evolved from the tolerance curve approach would be expected to provide accurate assessment of injury potential during direct head impacts.

The Head Injury Criterion (HIC), based on the resultant translational acceleration of the center of gravity of the head, is the most commonly used method of evaluating head impact data. Statistical analysis of direct head impact cadaver test data has been used to define the relationship between HIC values and the probability of sustaining a particular level of injury, thus providing a continuous ability to interpret HIC values. A HIC level of 1000 was found to produce an expected 16 percent incidence of life-threatening brain injury to the adult population.

The validity of the HIC for long duration and non-contact head accelerations remains in question. Injury criteria based on head angular acceleration and angular velocity have been proposed for such situations, but they lack the extensive evaluation and review that has been given the HIC for short duration (less than 15 ms) head impacts. Mathematical models of the head hold promise for evolving into injury predictive models given proper development and evaluation. Simple models, such as the mean Strain Criterion (MSC), which are based on translational acceleration, have the potential for describing the dependence of the injury response on impact waveform and direction of impact. The application of the MSC to dummy head accelerations, however, remains to be developed. More sophisticated finite element models of the brain and skull have been developed, but their complexities and lack of validation have hampered their development into injury predictive models.

The response of facial structures to impact loads has been studied to a limited extent. The fracture and collapse of the facial bones during distributed loading significantly reduces the peak forces and resulting head accelerations in comparison to those produced by similar impact tests to the skull.

The tolerance of the facial bones to direct impact loading has been studied by a number of researchers, and fracture loads for individual bones and the whole face have been determined. The failure characteristics of facial soft tissues due to laceration from sharp edges have been studied, and rating systems for the assessment of the severity of the lacerations have been developed. There is a need, however, to study the mechanisms of lacerations to facial tissue due to blunt impact.

SPINE

The vertebral column is the principal load-bearing structure of the head and torso and provides a flexible protective pathway for the spinal cord. Injuries that affect the function of the spinal cord can result in death, quadriplegia, or paraplegia. Despite these potentially serious consequences, the actual incidence of such injuries is relatively low, and thus they contribute probably less than 6% to the total IPR. (This figure is uncertain because NASS does not code the spine directly but rather incorporates it into the neck and back regions.)

The static and dynamic response of the head/neck system to indirect inertial loading at low crash severities has been studied extensively in volunteers and, to a lesser extent, in cadavers. These studies have included frontal, lateral, and oblique impacts. Specifications for suitable neck linkage systems, ranges of motion, and joint resistance characteristics are available from the published literature. Direct crown loading experiments have also produced data on the superior-inferior compliance of the cervical spine in cadavers.

The static midsagittal bending response of the thoraco-lumbar spine has been studied in volunteers for flexion and extension. Specifications in terms of overall rotation ranges and bending resistance characteristics of the rotation of the thorax relative to the pelvis have been produced. The equivalent dynamic data are quite limited but do indicate the presence of upper thoracic spine mobility with values similar to those for lower spine mobility.

The status of knowledge on the tolerance of the neck to loading is limited. Of necessity, all volunteer data are below the injury threshold. Additionally, injury mechanisms can be quite different than those mechanisms controlling response. Most injury threshold data are either based on cadaver tests or on reconstructions of accidents with instrumented dummies. As such, the threshold values are subject to the limitations associated with the surrogate used to obtain the data. These data sources have been used to develop limiting tolerance values for neck bending moments in midsagittal flexion and extension, axial compressive and tensile neck forces, and neck shear forces. No efforts have been made at this time to develop limit values associated with combinations of the various forces and moments. Corresponding studies of the tolerance of the thoracolumbar spine are not available. The only tolerance studies done on the thoracolumbar spine are those related to vertical accelerations.

THORAX

The thorax houses most of the body's vital organs and is thus the next most critical region, after the head, to protect from injury. Injuries to the chest constitute nearly 19% of the cost to society of injuries sustained by automobile occupants, as calculated using the Injury Priority Analysis. The nature of thoracic injury, however, is such that there are few long-term disabilities. In general, the victim either dies soon after impact or recovers completely.

The most critical injuries are those to the internal organs. In most experimental studies using cadavers, however, injury rating has been based on skeletal damage. As thoracic skeletal deflection increases under dynamic loading, the force resisting the motion remains somewhat constant. Further deflection begins to produce rib fractures, which can be followed by the sudden appearance of internal soft tissue injuries as the skeletal structure collapses. It is necessary, therefore, to be conservative in defining thoracic injury criteria in terms of deflection levels related only to rib fracture because of the instability of the thoracic structure under such conditions. Applied load by itself is also inadequate as an injury criterion, because of its insensitivity to increasing deflection in the force-plateau region characteristic of dynamic thoracic response.

Another factor that must be considered in defining thoracic injury criteria is the fact that thoracic response to impact loading is highly rate-sensitive. Viscous and inertial forces dominate the initial response, and elastic forces become significant only as large deflections of the system occur. Some forms of pulmonary and cardiac injuries have been found to occur only in conditions of high impact velocities with very little chest deflection. The rate of thoracic deflection as well as the degree of deflection can both be important parameters in describing the injurious effects of an impact to the chest, and they should both be considered in the development of general thoracic injury criteria.

In terms of response, the sensitivity of the thoracic structure to the rate of loading makes it difficult to interpret the findings from different types of experiments without accounting for this variable. For instance, the strip loading produced by the shoulder belt may produce an apparent stiffness that is lower than that produced by a flat circular impactor, due to differences in shape and area of loading. The rate of loading in shoulder belt tests, however, is usually much lower than that of the typical impactor test, thereby confounding the interpretation of shoulder belt interactions with the thorax. Impactor mass is a variable that can also strongly affect the apparent response of the thorax and must be accounted for when comparing experimental results.

Flat circular impactor tests tend to produce a characteristic thoracic force-deflection response that consists of an initial linear region, followed by a plateau region of almost constant force, and finally, if the impact has sufficient severity, a third region of increasing stiffness. This general form of response has been shown to be true for both frontal and side impact and with volunteers as well as cadavers. Thoracic structural rate sensitivity appears to be responsible for much of the initial stiffness and for the subsequent plateau in force as the rate of loading decreases during the impact. However, the distribution of load by the flat impactor surface must play some role in determining the response, since shoulder belt loading does not appear to produce the plateau region, even when loading rates are taken into account. Such local loading effects are not, however, well documented.

Because of the complexities of thoracic response, simple elastic structural representations are inadequate to guide the designer of mechanical analogues of the thorax. Instead, representation by means of spring-mass-damper models and/or transfer function approaches are necessary to provide the designer with the proper insight into the relative contributions of elastic, viscous, and inertial forces to the overall system response.

The three-dimensional structure of the thoracic skeleton and its contents requires deformation descriptors that are global in nature to provide an omnidirectional description of response. In the cadaver, this has been accomplished to some degree by the use of arrays of accelerometers on the periphery of the thorax. Similar or alternative methods of global response measurement will be necessary in the AATD to ensure adequate capability to assess injury potential in different directions and under different types of loads and loading rates.

ABDOMEN

The abdomen includes the organs and viscera below the diaphragm and above the pelvic girdle. Although there is little bony structure to protect these organs from blunt impact, injuries to this region contribute only 7.5% to the total IPR. Like the thorax, the abdomen can be the site of injuries induced by restraint systems themselves, including belts and steering systems. As far as the crucial organs are concerned, the liver, spleen, and kidneys are most frequently injured, and these injuries tend to be the most serious and life-threatening.

Injury mechanisms in the abdomen are thought to be primarily the result of deformation or penetration of the abdominal contents along with significant force or pressure generation in the deformed organs. In addition, solid organs, such as the liver, may undergo severe damage due to pressure generation alone at high impact velocities. There is evidence to show that these organs are viscoelastic, that the rate of loading is a crucial factor in injury causation, and that a compressive stress of 300 kPa (43 psi) will cause a superficial liver injury. Regarding dynamic response of the abdomen, the problem is complicated by the fact that there is a variety of surface geometries and component materials that can impact the upper abdominal area in a vehicle crash environment. In side impacts, however, the surfaces such as doors and armrests are somewhat well-defined, and dynamic load-deflection response curves do exist to a limited extent for lateral impact. Much more research data are needed, however, before abdominal response to impact can be fully quantified.

PELVIS AND LOWER EXTREMITIES

The pelvis is a bony structure that transmits the weight of the torso to the lower extremities during normal locomotion and supports the torso in the seated position. In an automotive impact environment, it can sustain injury from both frontal and side impact, and, during aircraft ejection or vertical falls, it is called upon to take the entire inertial load from seat-to-head acceleration. Injuries to the pelvis, however, contribute only about 1% to the total IPR. This structure is important, therefore, primarily for its response during load transmission.

The lower extremities constitute approximately one-third of the body weight, and, during normal locomotion, are required to withstand large dynamic loads. Injuries to the lower extremities of automobile occupants are rarely fatal but require significantly longer periods of hospitalization and lost working days than injuries to other body regions at the same AIS level. Even so, injuries to this region constitute only a little more than 5% of the total IPR.

The frontal impact response of the knee/femur/pelvis complex during seated knee impacts has been studied extensively. This research includes information on the acceleration-time histories, force-time histories, impedance, and effective mass. Other studies have defined the geometry of engagement of the knee into crushable padding. Load–deflection data are also available for subluxation of the tibia with respect to the knee joint. Lateral response of the pelvis has been studied for both impactor and flat-wall impacts and has been described in terms of force-time histories and pelvic acceleration-time histories.

Injury tolerance data for the knee/femur/pelvis complex consists primarily of axial loads in the femur. Lateral loading tolerances for the pelvis are available in terms of forces and peak accelerations. For the femur, tolerance to lateral impact can be defined in terms of maximum bending moment as can the loading tolerance of the tibia in the transverse direction. There is also information on the strengths of the knee-joint ligamentous structures.

INSTRUMENTATION, DATA PROCESSING, AND CERTIFICATION PROCEDURES (Task C)

INSTRUMENTATION

State-of-the-art technology was reviewed with regard to the measurement of force, moment, linear and angular acceleration, pressure, and displacement as might be applied to the AATD. Innovative near-term developments were analyzed in this context, and instrument size, weight, power consumption, performance, and compatibility with the planned data acquisition system were considered.

Current transducer technology is adequate for most of the sensors used in today's test dummies. However, as part of the effort to develop a multidirectional dummy that can respond more like a human and to instrument it with sensors that can more accurately yield injury data, some deficiencies do exist.

Force and moment measurements can be made using piezoresistive (PR) or piezoelectric (PE) sensors. PRs have the advantage of being smaller than PEs and can thus be easily incorporated into an AATD design. On the other hand, PEs are more well suited to multidirectional dynamic measurement. Their miniaturization, however, is not likely in the near future.

For linear acceleration, PR accelerometers are routinely used because a shunt calibration can be performed just prior to testing. There is some question, however, whether their frequency response is of sufficient accuracy to allow angular acceleration to be determined using multiple linear-accelerometer arrays. Although PE accelerometers are also available, their disadvantages with regard to pre-test calibration and compatibility with existing systems eliminate them as a choice at this time. Presently, angular accelerometers are not sufficiently developed to be considered for this type of application.

Pressure transducers of various types are available for the different requirements in the AATD. They can be used to indicate impact severity in the compartmented chest being considered. Direct measurement of the flow between compartments can also be made with the various flowmeters developed for medical applications. However, differential pressure measured across an orifice, whose size could be readily varied, would probably be more versatile and reliable.

DATA PROCESSING

The objective of this work was to review the data acquisition and processing techniques that are currently used by crash and sled testing facilities to measure anthropomorphic dummy response data during impact testing.

A number of recommendations are given regarding the development of an advanced dummy instrumentation system:

1. Based on the projected data channel requirements of from 72 to 100 data channels on-board the advanced dummy, and based on the present size of state-of-the-art instrumentation, the development of an on-dummy instrumentation system is

recommended. This system should be designed to include integral memory but with capability to conveniently expand record times through use of external memory.

The on-dummy instrumentation should be developed as a microprocessor based system to perform on-dummy analysis and control functions as well as to allow transfer of measured data to an external computer based test set. The on-dummy microprocessor would also function to perform self tests of individual data channels under both internal and external control.

2. A microprocessor based external test set should be developed as an integral part of the advanced dummy instrumentation system. The test set should be designed to allow calibration signals to be injected into individual data channels on-board the dummy. The test set would also function to perform analysis of data channel response signals. The response signals may result from injected calibration signals or actual dummy responses generated during dummy certification tests. The test set would allow for rapid turnaround of results to provide pass/fail indications of channel performance immediately after a verification test was performed.

3. The advanced dummy instrumentation should be developed to meet the requirements for data systems as outlined in ISO 6487, *Road Vehicles—Techniques of Measurement in Impact Test—Instrumentation.* In specialized cases, as may apply to the nine component head accelerometer array, increased accuracy may be required. The recommended performance for the advanced dummy instrumentation is summarized in Table C−1.

4. Procedures should be developed for use in calibrating accelerometers and other sensors. ISO 6487 should serve as a basis for the procedures. At a minimum, the following parameters should be calibrated at six-month intervals.

- Amplitude linearity at a fixed frequency
- Amplitude response versus frequency
- Phase response

The advanced dummy instrumentation should be designed to meet specified environmental performance criteria. Recommended criteria and suggested limits are summarized in Table C−2.

TABLE C-1

ADVANCED DUMMY INSTRUMENTATION REQUIREMENTS

Amplitude Linearity	2.5%
Amplitude Resolution	12 bits (0.02%)
Time Linearity	1%
Time Synchronization	0.1 ms
Time Zero Offset	0.1 ms
Sample Rate	8000 Hz
Record Time (per channel)	500 ms (extendable to 5 seconds)
Channel Capacity	72 (expandable to 100)

TABLE C-2

ENVIRONMENTAL SPECIFICATIONS

Parameter	Design Limits
High Temperature	180°F.
Low Temperature	−10°F.
Temperature Shock	170°F.
Humidity	85%
Acceleration (linear)	10 G
Vibration	1 G RMS Random
Shock	500 G, 0.5 ms
EMI/RFI	Standard Industrial

CERTIFICATION TEST PROCEDURES

The objective of this effort was to review test procedures that are currently employed to certify the Part 572 anthropomorphic dummy for use in crash tests and recommend approaches that appear promising for use with the advanced dummy. These certification tests are intended to ensure that dummy responses to impact stimuli are both repeatable and reproducible. A brief review of Hybrid III dummy certification procedures was also conducted.

A number of recommendations have been developed as a result of this review:

1. Routine certification testing should be done on a completely assembled advanced anthropomorphic dummy. Test procedures should involve dynamic exposure of the dummy to levels that are consistent with the automobile crash environment. That is, certification testing should mimic the in-use environment to the maximum extent possible. Test procedures and equipment must be developed that will allow efficient and rapid testing.

2. On-dummy instrumentation should be used in certification testing. It is recognized that additional electronic measurements may be necessary as a part of new test procedures; however, where possible, the sensors, instrumentation, etc., that are a part of

the dummy should be utilized to provide certification test data. As this instrumentation is part of the dummy, it is reasonable that it should also be checked as a part of the certification process. However, a complete calibration of sensors is not viewed as necessary each time a dummy is certified. Rather, a calibration interval would be defined at which time all instrumentation would undergo calibration according to established procedures.

3. Performance criteria should include injury measures as well as engineering measures. That is, if HIC is used as an injury measure for the head, then a test procedure which results in a comparison of a HIC with established error bounds should be utilized. This approach results in a better knowledge of error limits on the primary dummy outputs than is currently available. It should, however, be noted that measures of response in addition to injury measures may be desirable and even necessary for identification of sources of unacceptable component response.

4. Dummy disassembly and component level testing should be undertaken only if whole body testing indicates a problem in meeting response limits. That is, component level testing should be undertaken as a diagnostic aide in determining a specific mechanical item in need of repair or adjustment.

REVIEW OF DUMMY DESIGN AND USE (Task D)

DUMMY USER SURVEY RESULTS

A survey of users of the Part 572 and Hybrid III dummies was taken to determine the problems these users have encountered and their preferences for changes and improvements in dummy design. General topics included mechanical design, serviceability and maintenance, durability, certification, repeatability and reproducibility, and ease of use. The survey questionnaire was made available to a wide audience through NHTSA, SAE, ISO, and various individual contacts.

Responses were received from thirty-eight individuals representing twenty-nine organizations. The affiliation of the thirty-eight respondents can be categorized as follows:

7 U.S. vehicle industry
9 Foreign vehicle industry
12 U.S. government
4 Foreign government
2 Dummy manufacturing
4 Independent research

Throughout the responses to this survey, a strong emphasis was placed on the need for a durable, stable, and repeatable test device, even at the expense of biofidelity. The lack of enthusiasm for an omnidirectional dummy, expressed by several respondents early in the survey, seemed to be based on an assumption that such a dummy could not be made to be as repeatable and reliable as a unidirectional test device, while retaining a suitable simplicity of design. Comments in later sections of the survey clearly indicated that considerable time and effort was required to prepare the Part 572 dummy for testing, and that between-test repair, replacement, adjustment, and recalibration were frequent necessities. As the survey proceeded from theoretical design to more hands-on issues, the number of respondents decreased, but the conviction and level of detail of responses increased.

Respondents were generous with their advice as to how life with an anthropomorphic test device could be made easier. Designed-in means were needed for holding on to various parts of the dummy for transporting, positioning, and storage and for holding them fixed in space for certification tests. Also needed were visible indicators of the dummy's internal structural configuration and segment centers of mass. Joints were singled out as the assemblies in particular need of redesign. In addition, it was made clear that the performance characteristics of a dummy must be built up from the smallest components, but that performance checks of components in their assembled state were also necessary.

After reviewing the responses of this international group of dummy users, the term "advanced" in AATD begins to take on a broader meaning. Not only is there an opportunity here to advance the state of the art with regard to humanlike response and innovative instrumentation techniques, but there is a necessity to make significant design and materials improvements that will result in a more durable, repeatable, and trouble-free dummy.

DUMMY DESIGN, MANUFACTURING, AND COST CONSIDERATIONS

The Part 572 and Hybrid III dummies were reviewed in detail from the standpoint of design features, manufacturing techniques, and associated production cost effectiveness. Many of the problems identified by dummy users relate to design concepts and manufacturing processes that need to be improved for cost purposes as well. The review was done at the component/assembly level, but general recommendations regarding dimensioning, tolerancing, and the use of standard-size materials were also made.

By far the most important features of a part or assembly drawing are the dimensions. Their nature and format should even be considered in the design stage. Dimensions should be placed in an orderly and uncrowded arrangement on the drawing for ease of use, and their relationship to one another should reflect the part's engineering intent. In order to permit a variety of manufacturing methods, a drawing should only specify the desired result and not the process to be used in obtaining that result.

The purpose of manufacturing tolerances is to assure that variations in the manufacturing process are controlled, but they should also permit the greatest economy of production consistent with the functional requirements of the part or assembly being produced. The accumulation of tolerances and dimension limits is a potential problem. Because dimensions can vary from the extreme high to the extreme low on related parts, it is possible that an undesirable condition will be created upon assembly. The likelihood of such a condition can often be reduced if parts are dimensioned such that the minimum number of dimensions and thus tolerances are involved in each aggregation.

Stock sizes for finished parts should be selected whenever possible to reduce material, tooling, and machining costs. Although it is generally easier to hold an external diameter to a closer tolerance than an internal one, if a shaft can be made from standard stock without further machining, it may be more economical to use that stock and specially machine the sleeve or housing.

The Hybrid III was considered particularly well documented. For most parts and assemblies the drawings are clear, and the parts list and parts list index are a great service to test dummy manufacturers as well as users. The head, neck, and lumbar spine are of simple and functional design. The clavicular, shoulder, elbow, and knee joint designs represent a significant improvement over the clevis-type joints on the Part 572 dummy. Other features representing advances in the state of the art with regard to dummy design are the six-axis neck transducer, the adjustable neck base, the designed-in chest deflection measuring device, the systems for mounting a mechanical pelvic angle gage and a chest target angle gage, and the removable insert in the knee impact area. Manufacturing cost efficiencies could be realized, however, by simplification of some of the complex shapes as well as by relaxing some of the dimensional tolerances and surface roughness value requirements, which would not affect the test dummy weight distribution or its dynamic performance.

REPEATABILITY AND REPRODUCIBILITY

The objective of this work was to identify appropriate techniques for assessing repeatability and reproducibility where possible, and to identify areas requiring further development in Phase 2. Repeatability refers to the variance of replicate tests with the same test device, while reproducibility refers to the variance arising from different test devices. Although reproducibility may be a primary design consideration, the level of repeatability in both component and actual testing represents a practical limit on the levels of reproducibility.

Specification of dummy design objectives will be influenced by the magnitude of sources of error *apart* from the dummy, as well as sources of error inherent to the dummy. In a discussion of error, it is important to distinguish random errors from bias, or systematic, error. Although any error is undesirable, random error is the preferable of the two, since it has zero mean by definition. Consequently, statistical techniques can identify and measure random error, and estimates of the desired quantities can be made as accurate as one wants (if the necessary sample sizes are not impractical). However, bias error nearly always causes serious problems since, by its nature, it is not distinguishable from the desired response.

An outline of sources of error is given below in relation to the measurement of dummy repeatability and reproducibility.

A. Repeatability

1. Variation in the Test
   - Type of test (vehicle, sled)
   - Type of seat (hard, soft)
   - Initial speed
   - Deceleration pulse
   - Signal processing
   - Positioning

2. Variation in the Dummy
   - Variation in the dynamic response (due to joint friction changes, etc.)
   - Transducer errors
   - Permanent damage, or deterioration

B. Reproducibility
   - Systematic differences in response from one dummy to another

An obvious point is that the error associated with repeatability will include any inherent variation in the test itself. Consequently, when measuring repeatability one would like to use the most repeatable test procedure possible as long as the dummy is exercised in a manner that is representative of its intended use.

Analysis-of-variance techniques are the methods of choice for evaluating the repeatability and reproducibility of time-independent measures. The techniques separate "within" and "between" variability. "Within" refers to the variability of individual dummies over replicate tests, and "between" refers to the variability that arises from the use of different dummies in replicate tests. The null hypothesis is that the variability from one dummy to another is not greater than the variability of an individual dummy. Sample size can be determined to identify a specified level of between variability given the magnitude of the within variability.

Time-dependent measures may compare either the magnitude or the phase relationship of the transducer time-history. Currently, the most attractive measure that has been proposed is the Normalized Integral Square Error (NISE). This measure is derived from the autocorrelation functions. It is relatively easy to compute, and has the advantage of partitioning the error into that arising from phase differences, magnitude differences, and a remaining error that may be interpreted as waveform differences.

Donnelly, Morgan, and Eppinger have developed limits of acceptability for the NISE derived from the comparison of a single pair of time-histories based on an overall percent error criterion. However, methods for statistical hypothesis testing for this measure have not been developed. Such methods are needed to compare the distribution of the NISE obtained from several replicate runs as would be needed to assess repeatability. This work would seem to be a logical task for Phase 2.

ANTHROPOMETRIC DATA AND BIOMECHANICAL RESPONSE SIMULATION FOR AATD DESIGN

A summary is presented of the anthropometric data available for use in developing design specifications for an AATD. The use of these data is also demonstrated in a limited simulation of dummy response using the CAL-3D crash victim simulation code. The parameters of particular interest are those associated with spinal flexibility and shoulder mobility, which are shown to play a major role in controlling torso and head motions and applied forces.

Anthropometric Data. Several data resources are available from the recently completed dummy anthropometry project on Contract no. DTNH22-80-C-07502. These include mass and inertial properties, body surface shape, seated posture, estimations for the location of the bony structure, joint locations, and range of motion at the various joint structures. These data are all static and are for a mid-sized male driver in an average vehicle-seated posture. Mass and inertial properties are given for a traditional linkage and segmentation of the body in the static seated posture. The mechanical term "linkage" implies that the available data are most applicable to dummies or simulations based primarily on rigid-body mechanical models. The dummy response simulation performed here uses a lumped-mass chain-linkage dynamic simulation software package consisting of rigid masses connected at joint structures.

Details are given in the report of the body linkages used as well as data sources, however limited, and methods for determining the segmentation of the shoulder, neck, thoracolumbar spine, and thorax. Of particular importance are (1) the division of the traditionally rigid thorax into one rib-cage, three spinal, and two shoulder segments and (2) the approach of coupling the rib cage to the spine to achieve mobility and rotation of the thorax with respect to the spine.

Mobility and range-of-motion data are in fair supply only for static mobility and voluntary motion. In other words, it is possible to estimate how far the segments can move with respect to each other before subjects say "ouch" or before outside forcing agents must be supplied to produce further motions. Similar quantitative data for dynamic mobility (dynamic motions voluntarily made or those that occur under the influence of an outside forcing agent, such as a deceleration device) are virtually non-existent, except for the neck and, perhaps to a more limited extent, for the thoracic and lumbar spines. Dynamic torque data are not yet available for the thoracic and lumbar regions.

Other data necessary for simulating a dummy in a crash environment include force-deformation characteristics of various body regions and data on the crash event itself. Data sources included project literature reviews, accident data analyses, and vehicle specification packages.

Regarding the status of the anthropometry data base, it was concluded that the mass and postural data are available, but body segmentation needs to be refined to include spinal and shoulder girdle masses. The data base on joints also needs to be expanded to provide torque on joints versus relative angle of the segments. The most pressing need is for additional information on spinal flexibility and torsion properties.

Biomechanical Response Simulation. The purpose of this analysis was to study the effects of spinal and shoulder flexibility on whole-body response. Degrees of freedom not found in current dummies were added to the linkage model. Comparative results for stiff and flexible joints between the thorax components are presented for belted and unbelted occupants.

Figure D-1 shows the difference in head excursion for three-point belted occupants with a flexible three-link thoracic spine versus a stiffer linkage similar to that used in current dummies. The positions are those at 100 ms, and the more restricted forward head excursion with the stiff joints is apparent. For the unbelted case, there was a marked reduction in the force of interaction between the thorax and the steering wheel when the shoulder masses were uncoupled from the thorax, even when the spine remained stiff.

It is clear from these exercises that crash victim simulation software can be used to study the effects of changing dummy design parameters. Specifically, it was shown that the addition of thoracic spinal flexibility has major effects on the crash victim motion. This is particularly noticeable in head excursion increases of several centimeters for belted occupants. In addition, it was shown that the uncoupling of the shoulder masses from the thorax has a major effect on the interactions of a crash victim with vehicle structures forward of the occupant, such as the steering column and the instrument panel.

ATD CRITIQUE

A review of current ATD designs was made from the standpoint of biofidelity, measurement capability, directionality, and impact testing performance. Although a wide variety of currently available test devices was reviewed, primary emphasis was given to the Hybrid III and the Side Impact Dummy (SID), since they represent test devices currently in use in research and developmental testing in this country.

The Hybrid III was developed over ten years ago, and its design is based on biomechanical knowledge available at that time. It is a frontal-loading-only ATD whose design represents an evolutionary improvement over conventional ATD design. For example, the rib cage uses the same design as the previous Hybrid II, but with altered structural stiffness to produce a humanlike impact response to mid-sternal moving-mass impactors. The Hybrid III also possesses humanlike hard-impact forehead response and midsagittal neck bending response. As presently configured, the Hybrid III has the greatest measurement capability of any ATD, with 44 data channels.

The SID was developed more recently than the Hybrid III and represents a modification of the chest region only in an otherwise standard (non-biomechanical) ATD. The SID was developed to provide lateral chest response biofidelity under rigid and padded impacts. The shoulder response is included in the chest response, and as a result the design has no separate shoulder structure. The remainder of the SID structures are standard Part 572. Except for additional chest wall accelerations and lateral chest displacement, the SID has the same measurement capabilities as the Part 572 ATD.

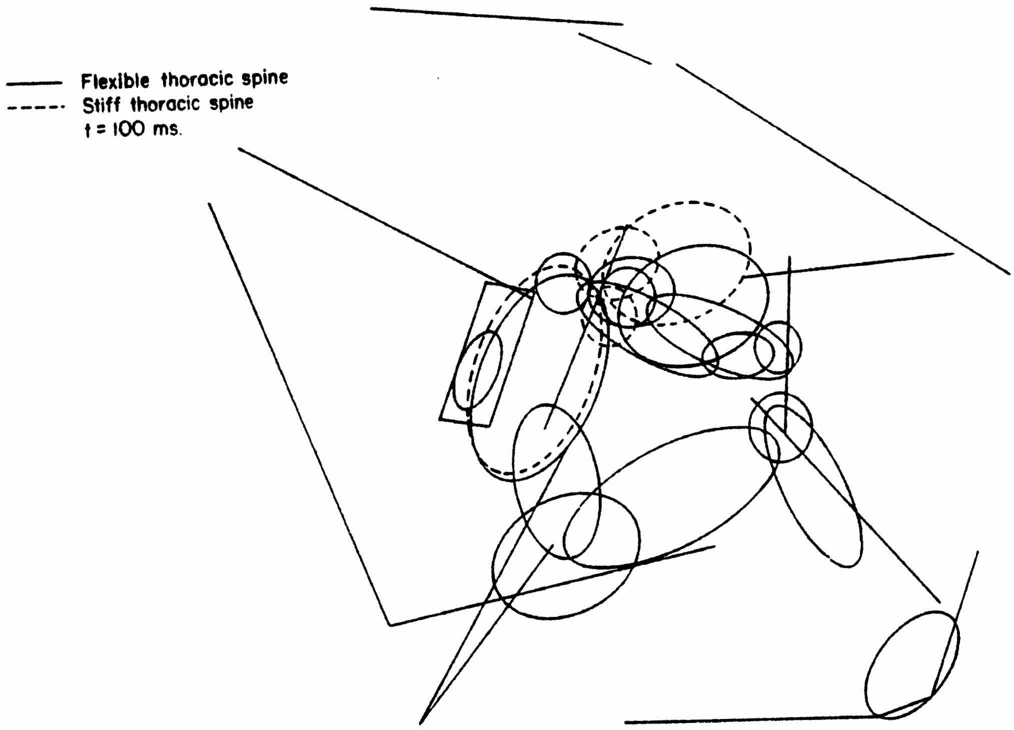

FIGURE D-1. Effect of Thoracic Spine Flexibility. Belted Case.

Both the Hybrid III and the SID are examples of ATDs intended for use in restricted test conditions and/or directions, and as such they have only limited biofidelity in the principal directions and none in other directions. One of the most serious deficiencies of both ATDs centers around the designs of the thoraxes, both the rib cages and the spines. The IPR analysis has indicated the great importance of the head and chest as primary sources of injury, disability, and death of unrestrained occupants. The development of effective countermeasures to minimize injury to these regions depends strongly upon having an ATD that produces realistic responses in terms of trajectories, contact-points, and loadings. The combination of rigid thoracic spines with present neck designs (including that of Hybrid III) and inadequate thoracic rib-cage conformability are producing head contacts and chest/steering-system interactions that are quite unlike those in real-world crashes. All present ATD designs are quite inadequate in this respect in the crucial head/torso regions for frontal, lateral, and oblique impacts.

The lack of realistic concentrated load response in the Hybrid III and SID chests is compounded for the case of shoulder-belt loading. Neither chest exhibits humanlike stiffnesses at the lower loading rates associated with belt restraint systems. This, again, will have an influence on both chest deformations and on head trajectories.

To illustrate the improvement in overall effectiveness of an AATD with omnidirectional response and measurement capability in the frontal-to-lateral range, an analysis was made of the Hybrid III, SID, and AATD based solely on measurement capability and directionality. Direct-impact biofidelity was not taken into account. The IPR distribution for each body region (Table A−1) was multiplied by the measurement capability of each ATD for each region, based on a rating of complete (100%) to partial (75% and 50%) to none (0%). This product was then apportioned according to the distribution of IPR to these body regions by direction of impact force. This process gave an estimate of the proportion of the IPR addressed by each ATD in a given direction. An overall effectiveness of an ATD could then be estimated by summing these IPR proportions for the range of directions from frontal to lateral.

For the left-front unrestrained driver position, it was judged that the combination of Hybrid III and SID ATDs could address directions 11, 12, and 1 o'clock (Hybrid III) as well as 9, 10, and 3 o'clock (SID). The potential gap in the oblique direction between 10 and 11 o'clock is ignored to give the Hybrid-III/SID the benefit of the doubt. The 2 o'clock PDOF, however, is not included because SID is judged to be kinematically unreliable in far-side oblique impacts. In comparison, the AATD will be able to continuously address the full range of PDOF from 9 through 3 o'clock in a clockwise direction. In addition, the more extensive measurement capability to be available on the AATD will provide a higher level of injury assessment capability in those directions. Based on the above considerations, the AATD was found to address 90.5% of the total IPR (see Figure D−2), while the combination of Hybrid III and SID addressed only 43.4%. The AATD would thus be twice as effective as the combination of the two present ATDs.

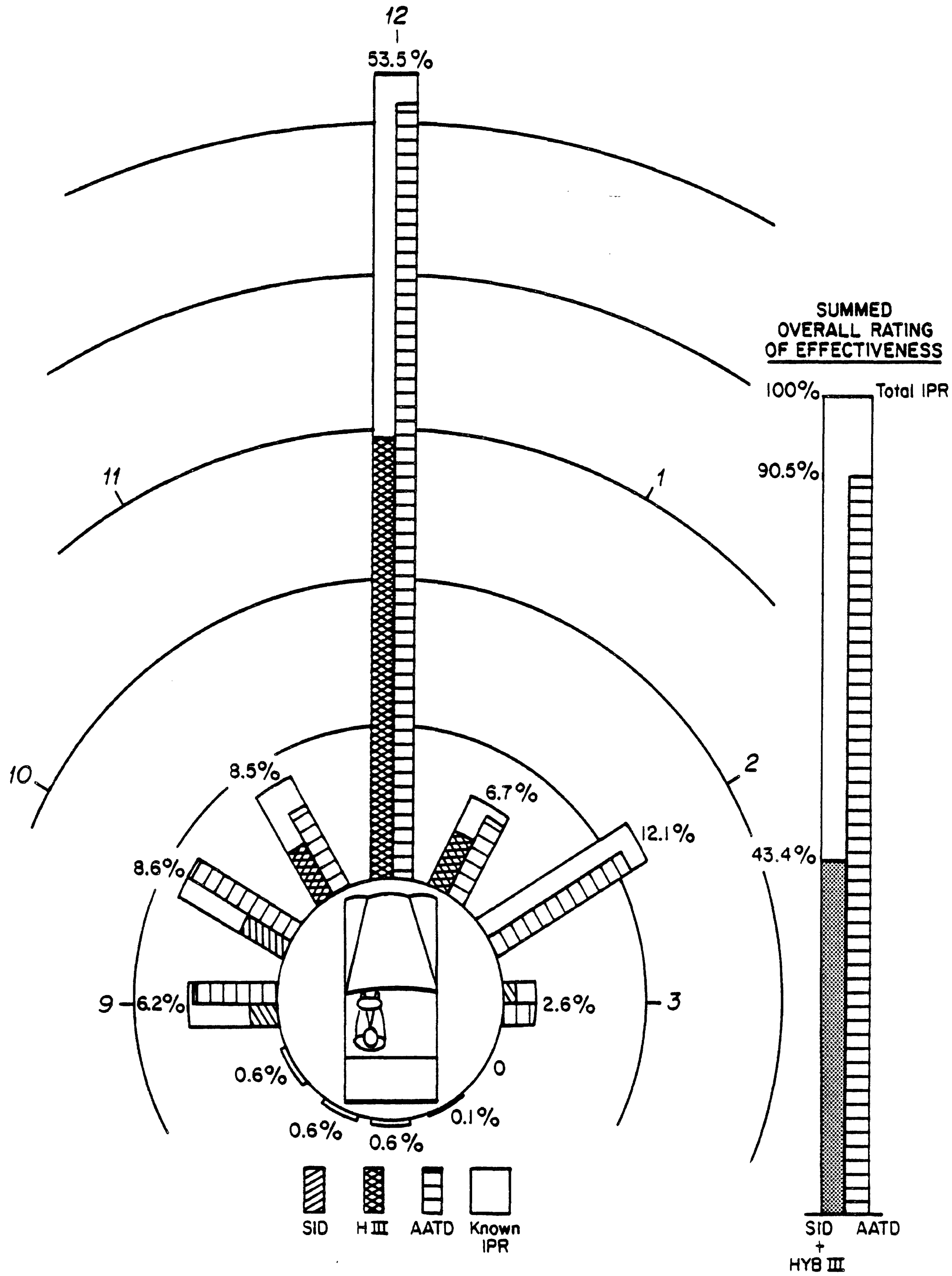

FIGURE D-2. Estimated Effectiveness of Hybrid III, SID, and AATD for Unrestrained Left-Front Occupants by Principal Direction of Force. (Percentages denote proportion of horizontal force IPR in each direction.)

AATD TECHNICAL CHARACTERISTICS, DESIGN CONCEPTS, AND TRAUMA ASSESSMENT CRITERIA (Task E & F)

This task brings together the results of our various reviews and analyses of accident data, biomechanical response and injury data, anthropometric data, and current ATD design, instrumentation, data processing, and certification procedures in order to establish technical characteristics for the AATD, propose injury criteria, and develop design concepts that will achieve program goals. In addition to the task activities summarized in previous sections, an important additional activity was in progress that provided necessary data for the current task. This was the analysis of biomechanical data, which is briefly described here.

BIOMECHANICAL DATA BASE

Our ability to generate response corridors for specific body regions, but particularly for the chest, depended on the existence of a large data base whose signals could be analyzed in a uniform manner. Both previously unpublished data and those published data that warranted reanalysis were identified, consolidated, categorized, and recoded or given additional coding as appropriate. A total of 1,190 tests were initially identified by test number and source as candidates for inclusion in the data base. However, we were able to obtain and include adequate data for only 221 tests, consisting of the following:

107 from UMTRI
55 from Heidelberg University
41 from ONSER
12 from WSU
4 from APR
2 from Calspan

These included the following test configurations:

110 pendulum impacts
52 three-point-harness tests
45 lateral sled tests
14 airbag tests

To augment these data, high-speed movies from 13 UMTRI thoracic tests were analyzed, and film readings were reformatted as "displacement" signals and incorporated with their corresponding sensor data. Finally, 10 Heidelberg tests containing 9 accelerometer signals were converted to the standard anatomical reference frame.

Various parameters were used as the basis for classification of the tests. These included (1) restraint type or impact surface, (2) severity of impact, in terms of impact velocity, (3) injury level, in terms of AIS rating, and (4) the subject size and condition. Test signals were further subdivided by body area, including the head, chest, spine, shoulder, and lower extremities. Once these groups were established, spectral analysis of each group was conducted in order to determine an appropriate filter to use for pre-processing. A procedure was developed to extract filter characteristics from the available pool of signals, and filter corners and slopes were obtained for the near side and far side of impact for the chest and head. The area determined to have the highest priority for response characterization was the thorax, and thus emphasis was placed on generating corridors for rib and sternal accelerations. Summary data plots of chest response for a variety of test conditions (shoulder belt, airbag, rigid wall, and rigid disc) were prepared, the data having been filtered and mass-scaled using the Livi Index. These are the basis for the design specifications of the chest.

AATD TECHNICAL CHARACTERISTICS AND DESIGN CONCEPTS

The AATD will be designed to provide omnidirectional response in the range of $\pm 90°$ from the front in the horizontal plane. This accounts for virtually all the horizontal collision IPR and 82% of all IPR with known principal direction of force (PDOF). The exposure severity levels associated with cumulative values of 85% of the IPR are delta Vs of 50 mph for frontal impacts and 30 mph for lateral impacts. It is expected that the AATD would be used unrestrained in such severe environments only if the vehicle structures and interiors to be tested incorporated advanced crashworthiness technology. Thus these delta V levels are taken to be upper limit exposure levels for the velocity of interaction with protective interior systems. The AATD will also be designed for upper body vertical (superior-inferior) impact associated with non-horizontal collisions. Such collisions account for 18% of all IPR with known PDOF.

Measurements to be made by the AATD are based on an analysis of ideal versus feasible measures. This analysis identified measures that would be desirable for injury assessment, given that an ideal test device could be developed that could reproduce all the anatomy and biomechanical responses of the human body. These were tempered by the state-of-the-art of measurement technology, and measurements were determined that would be consistent with the AATD design concepts for each body region. These measures are included in the following summaries of the desired technical characteristics and associated design concepts for the AATD.

Head. The head will have biofidelity of response for front, side, and top rigid impacts as well as facial impact response biofidelity. Head instrumentation will consist of an array of twelve linear accelerometers to provide complete three-dimensional impact motion measurement as well as direct head center-of-gravity translational acceleration measurement. Certification testing procedures for head response will not require disassembly of the AATD.

The head will have a featureless face to aid in producing repeatable impact response. In addition, the facial structure will be designed to produce realistic head acceleration responses when the head is impacted in the face by a rigid mass. This face will be a durable structural element that can, however, be removed and replaced with an optional frangible insert for determination of facial bone fracture. The overlying soft tissue simulation of the face would similarly be replaceable with a lacerable option.

The skull of the AATD will be of cast aluminum with the front, side, and top sized to produce, in conjunction with the scalp material, biofidelity of rigid impact response. The base and rear structure of the skull will be separate from the front/side/top structure and will be designed as a mounting structure for a 12-accelerometer motion measurement array made up of four triaxial accelerometer units. As such, the base and rear structure will be very stiff to preserve the accelerometer alignment during rigid impacts to the head. The array will be configured to put one of the triaxial units at the head center-of-gravity for direct measurement of translational acceleration at that point and for subsequent HIC determination. The base of the skull will also serve as a mounting structure for a six-axis neck load cell.

Spine. The spine will be designed as a total system from the base of the head to the top of the pelvis. Spinal flexibility will be provided for the cervical, thoracic, and lumbar spine segments. The neck section of the spine will have biofidelity of response for frontal, lateral, and oblique indirect impacts. The specifications for these responses are based on analysis of human volunteer sled tests. These data provide static neck joint stiffness characteristics in bending (flexion, extension, and lateral), axial loading (tension and compression) and torsion, as well as dynamic head/neck junction bending moment/ head-to-torso angle responses for flexion, extension, and lateral loading and accompanying head trajectory requirements. An additional requirement will be placed on matching the superior-inferior response of the head/neck system due to impacts to the top of the head. The mean force-time and acceleration-time responses of the head and spine, based on cadaver impacts, are the basis for this response requirement.

The thoracic and lumbar spine system will have biofidelity of midsagittal flexion and extension bending response based on static tests on human volunteers. The spine system will provide for adjustment of the initial spinal configuration. Six-axis load measurement (three moments and three forces) will be made at the base of the head, the base of the neck, and the base of the spine to aid in the interpretation of body loading from advanced restraint techniques.

The spinal structure will consist of a series of rigid links connected by joints at selected anatomical points. These joints will have omnidirectional motion capability and will be located at the head/neck junction, base of the neck (C7/T1), upper thorax (T4/T5), middle thorax (T8/T9), lower thorax (T12/L1), and base of the spine (L5/S1). Range-of-motion requirements in the cervical spine (75° flexion, 90° extension) require the introduction of a third neck joint near the middle of the cervical spine (C4/C5). This joint will also incorporate provisions for neck torsional stiffness and axial stiffness control.

The spine segment between T1 and T4 will provide the structural attachment for the shoulder, and the thoracic structure will be attached to the T5-T8 and T9-T12 segments. Provisions will be made at each joint to allow adjustment of the initial configuration through the use of wedge-shaped rigid blocks that can be inserted and fastened in place.

Shoulder. The shoulder will have a clavicle structure to carry shoulder belt and steering wheel impact loads but will also offer low lateral load resistance for side impact. The lateral stiffness and range of deflection of the shoulder will be matched to mean cadaver response. The shoulder linkage will be designed to reach its lateral deflection limit slightly after the chest reaches its lateral deflection limit.

Thorax. The thorax will be designed to provide biofidelity of response to frontal impacts for rigid disc, shoulder belt, and airbag loading conditions. Side impact response biofidelity will be required for rigid disc and rigid wall impacts. The response requirements will be met for different rates of loading as well as for different types of loading. The response specifications are based on volunteer tests at lower impact severities and on cadaver tests at the higher levels. Mean impact force-time and acceleration-time corridors for all types of direct thoracic impacts and force-deflection corridors for rigid disc impacts are the primary specifications. They will be supplemented with static load-deflection data on volunteers. Thoracic cage deformations will be measured at multiple locations to allow the assessment of global chest deformation and deformation rates.

The thorax structure will have a thin, flexible, monolithic shell to define its shape and to provide load distribution. This shell is not intended to carry significant load and therefore will not have an influence on thoracic response. The response elements in the thorax will be an array of fluid-filled bag compartments within the shell. Each bag will be constructed of cord-reinforced rubber and will represent a flexible, constant volume reservoir. There will be five bags on each of two levels, compartmentalizing the chest into upper and lower levels with frontal, lateral, and oblique sections. Each bag will communicate with a common gas-pressure-controlled reservoir (accumulator) through a single orifice into that reservoir. One-way flow-control valves will be provided in the passages from the individual bags to the reservoir to prevent flow from one bag to another instead of into the accumulator. The bag volumes, the accumulator fluid volume and gas volume, the accumulator initial gas pressure, and the orifice size will be adjusted to produce the desired thorax response. The compartmentalization of the chest will allow positional variation of local response in order to better match human response to asymmetric loads such as those from shoulder belts.

Each bag will have a pressure transducer mounted in it as will the gas section of the accumulator. The bag pressure transducers will indicate the region of loading and the loading rate (related to flow rate) from the resistance to flow. The central accumulator gas pressure will indicate total global deformation of the chest independent of the region of loading. The use of well-defined materials, such as silicone fluids and nitrogen gas, to control response will minimize repeatability, reproducibility, and temperature sensitivity problems.

Abdomen. The abdomen will be a deformable system with dynamic response biofidelity based on the mean rigid armrest load-deflection response of laboratory-impacted cadavers. The deformations of the abdomen will be measured at multiple locations to indicate the degree of abdominal intrusion in frontal and lateral loading. The abdomen will be of similar general design as the thorax but with only three fluid-filled compartments, lower control pressures, and a softer outer covering.

Pelvis. The pelvis and its associated covering will be designed to exhibit rigid impact response biofidelity, based on mean force-time and acceleration-time corridors for cadavers in lateral impacts, and will have humanlike mass distribution. It will also be instrumented to sense lateral loading. The pelvic structure will be designed as a simple geometric shape for the purposes of (1) providing a lightweight, rigid structure for tying the extremities and torso together, (2) allowing lateral pelvic and hip joint load measurement, and (3) providing a protected space for housing a future data acquisition system. The anatomical details of critical points for proper restraint system engagement and interaction, such as iliac bones and ischial tuberosities, will be provided for only where needed. The mass of the pelvic structure and the accompanying soft tissue mass and stiffness will be matched to provide lateral impact biofidelity.

Lower Extremities. The legs will have knee/femur/pelvis impact biofidelity in rigid knee impacts and will have humanlike skeletal mass distribution. Soft tissue coupling to the skeletal structure will also be humanlike in response, and the knee structure will have realistic geometry. Six-axis load measurements will be made in the femoral shafts, and multiaxial loads will be measured in the lower legs. The extremity joints will be single-axis planar joints or combinations of such joints to achieve appropriate degrees of freedom. The ranges of motion of the joints will be humanlike, and the resistance characteristics will be adjustable to achieve resistance ranges from 0 to 2 G. The resistive torque-angle response will be humanlike as will the joint stop characteristics.

Flesh. The flesh will be developed in conjunction with the underlying skeletal structures to ensure proper overall system response, including tissue shear mobility in critical loading areas such as the buttocks and upper pelvis. The durability and stability of the flesh materials will be given special consideration.

TRAUMA ASSESSMENT CRITERIA

An analysis of existing injury criteria has been made to identify those mechanical parameters that best indicate the potential for injury. This information will help guide the design of the AATD so that measurement capabilities will be provided that relate to injury assessment. Additional analysis has been conducted aimed at developing new or modified trauma assessment criteria (TAC), using measurement capabilities made possible by new AATD design concepts. As part of Phase 1, work was also performed to develop improved TAC for the head and chest. The following TAC limit values are recommended for each body region. These values represent response levels below which significant injury to that body region is unlikely to occur.

Head. The head of the AATD will be designed to measure the three-dimensional motion of the head during an impact. Complete information will be produced for both translational and angular motions. The influences of rotational accelerations and velocities on brain injury have been postulated, but injury threshold values for the human are not well established. The combined effects of rotational motions and translational motions were investigated as part of the Phase 1 work, but there are insufficient data at this time to allow establishment of combined limit values.

The best choice for a head trauma assessment criterion would appear to be the HIC method, but with a limit on the time interval over which it is calculated. This limit is important because the biomechanical basis for the HIC method is direct head impact. Thus we recommend a value of

$$\text{HIC} = 1000, \quad \text{for } (t_2 - t_1) \leq 15 \text{ ms}$$

Spine. The spine will have six-axis load measurement capability at the head/spine junction, at the base of the cervical spine (C7/T1), and at the base of the entire spine (L5/S1). There are limited data on injury threshold values for individual load directions at the head/spine junction, but not for the combined effects of such loads. TAC values for individual loads to the top of the spine (C1), to the base of the cervical spine, and the base of the spine are given in Table E-1.

TABLE E-1

SPINAL INJURY THRESHOLDS

	Top (C1)	C7/T1	Base (L5/S1)
Bending Moment			
Flexion	190 N-m	380 N-m	1235 N-m
Extension	57 N-m	114 N-m	370 N-m
Lateral	Between the above	Between the above	Between the above
Force			
Duration >45 ms	1.1 kN	1.1 kN	3.8 kN
Duration < 45 ms	Increase to limit of	Increase to limit of	Increase to limit of
Tension	3.3 kN at 1 ms	3.3 kN at 1 ms	12.7 kN at 1 ms
Compression	4.0 kN at 1 ms	4.0 kN at 1 ms	7.0 kN at 1 ms
Shear	3.1 kN at 1 ms	3.1 kN at 1 ms	10.7 kN at 1 ms

Thorax. The AATD thorax design will have a totally new way of determining deformations and deformation rates associated with chest loading. Thus, methods for relating these measurements to injury criteria will be developed in conjunction with the development of the chest concept. Since the thorax will have biofidelity of impact response, it is possible to use presently available thoracic deflection criteria and critical values of deflection rates at specific points under specific impact conditions as starting points.

The presently available data indicate the following range of TAC limit values for the thorax, with oblique loading limits being appropriately in between:

Frontal deflection at mid-sternum	7.5 cm	(3.0 in)
Lateral deflection at nipple level	6.0 cm	(2.4 in)
VC_{max} (frontal through lateral)	0.9 m/s	(3.0 ft/s)

Abdomen. The abdomen of the AATD will have the capability of measuring both global deformations and rates of deformation. This will be done through pressure measurements in the fluid and gas sections of the abdominal system. Local fluid pressures can also be related to injury potential. The presently available data indicate the following TAC limit values for the abdomen for frontal through lateral loading:

Deflection	7.5 cm	(3.0 in)
Dynamic pressure	265 kPa	(38.4 psi)
V*C	1.65 m/s	(5.4 ft/s)

Pelvis. The pelvis will have the capability of measuring lateral impact loads at three locations. The TAC limit value for the pelvis is 5 kN for any individual load-cell location, with a total distributed load of 15 kN.

Lower Extremities. The AATD will have six-axis load measurement capability in the femoral shaft and the tibial shaft. The TAC limit values for loading to the upper and lower legs are as follows:

Femur		
Resultant bending moment	336 N·m	(270 ft·lb)
Axial compression	10 kN	(2250 lb)
Tibia		
Resultant bending moment	244 N·m	(180 ft·lb)
Resultant shear load	4 kN	(900 lb)
Knee sheer displacement	15 mm	(0.6 in)