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HUMAN CRASHWORTHINESS AND CRASH LOAD LIMITS

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SUMMARY

The assessment of the hazard to a crewmember in a potential aircraft crash requires information about human tolerance to mechanical forces, a method for the identification and evaluation of the contributing factors to potential injury in a crash and a means of estimating the environmental forces on an occupant and the resultant responses of the occupant. Results of research in the US Air Force in these three areas are discussed. Specifically, 1) the rationale for and formulation of a dynamic response six degree-of-freedom whole body impact tolerance specification, 2) a detailed head-spine structural mechanics and a vehicle occupant gross motion rigid body dynamics model and their areas of applicability in crash analysis, and 3) features of the newly developed US Air Force Advanced Dynamic Anthropomorphic Manikin (ADAM) are discussed. It is suggested that a comprehensive method for injury risk assessment for any system must consider all three areas.

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LIST OF SYMBOLS

$\ddot{\delta}$	- acceleration of the dynamic response model mass relative to acceleration input point.
$\dot{\delta}$	- relative velocity between the input point and the model mass.
δ	- compression of the model spring.
ζ	- damping coefficient ratio.
DR	- dynamic response of the model.
ω_n	- undamped natural frequency of the model.
$\ddot{z}$	- seat acceleration component along the pertinent axis.
g	- acceleration due to gravity.
DRX	- dynamic response computed from the X axis acceleration component.
DRY	- dynamic response computed from the Y axis acceleration component.
DRZ	- dynamic response computed from the Z axis acceleration component.

I. INTRODUCTION

The ultimate effectiveness of the crashworthiness of an aircraft is its ability to protect the aircraft occupant. Various measures of an aircraft's structural integrity can be made, but the final measure is the state of the crewmember after the crash. This state primarily depends on two factors: the level of exposure to mechanical forces experienced by the crewmember and the crewmember's susceptibility to injury due to such exposure. The first requires the specifications of such conditions as whole body accelerations and localized impact forces on the body and the second requires the specification of the criteria for tolerance to such accelerations and forces.

In this report some of the latest advances in methods for injury potential assessment developed by the US Air Force are discussed. These include criteria for whole body tolerance to acceleration, analytical methods for predicting human body responses to various mechanical force exposures and the development of a highly sophisticated manikin.

II. SIX-DEGREE-OF-FREEDOM ACCELERATION EXPOSURE LIMITS

The current method for assessing the risk hazard associated with whole body acceleration and which takes into account the dynamic character of the body's response was developed by Stech (Ref. 1) and is known as the Dynamic Response Index (DRI). It idealizes the human response as that of a mass supported by parallel elastic spring and damper elements which respectively represent the upper body mass and the lumbar/thoracic spine. This model was originally developed to provide an injury risk assessment for aircrew members being ejected from aircraft and its applicability was strictly limited to longitudinal spinal accelerations. While this model was developed to address ejection problems, it also was applicable to other situations where the body primarily experienced abrupt vertical accelerations as, for example, in helicopter crashes.

This model has recently been generalized to also include fore-aft and lateral responses as well as an added rotational tolerance mechanism (Ref. 2). The approach taken has included the following assumptions and steps:

1. Use of the same dynamic model for all three orthogonal body axes.

2. Assignment of an injury-risk level for each axis.

3. Assumption of independent responses along each orthogonal axis.

4. Evaluation of the injury risk level assumptions using existing data from impact tests including ones with acceleration vectors off the orthogonal axes, ejection seat test data, and studies with mathematical models.

5. Assignment of angular acceleration limits based on the effects of their translational acceleration components.

The equations that describe the dynamic response along each major axis are given in the following equations:

$$\ddot{\delta} + 2\zeta\omega_n\dot{\delta} + \omega_n^2\delta = \ddot{z}$$

and

$$DR(t) = \frac{\omega_n^2\delta(t)}{g}$$

The axes are taken so that the +Z acceleration acts from foot to head and a +X acceleration acts from back to front.

Each of the dynamic response models, other than for the +Z axis, were developed by the same procedure. First, the experimental acceleration-time histories from tests with volunteer subjects were approximated with a half-sine pulse where feasible. The test data, which were measured on the test fixtures that transmitted the acceleration to the subjects in whole-body impact tests, were obtained from numerous reports published by US Air Force and Navy investigators and Department of Defense contractors. The approximations were established by fitting the peak acceleration and the time to the acceleration peak (rise time) with a half-sine pulse. This procedure yielded relatively good fits for the majority of the data. However, the fit was not good where the experimental acceleration pulse shape was actually more trapezoidal, as in some of Stapp's early tests (Ref. 3) or where the acceleration-time history was irregular. In such instances, the procedure used was to fit only the initial portion of the pulse; this approach provided a conservative estimate since the energy of the fitted half-sine pulse was always less than that contained in the actual data. Second, a model response curve was calculated which was descriptive of the higher acceleration data points where, in many cases, subjective tolerance limits of injuries had been identified by the original investigators. The curve was derived by computing the peak response of a single-degree-of-freedom model to half-sine acceleration pulses of varying durations. To select the natural frequency and the damping coefficient ratio for each axis, the natural frequency and the damping coefficient ratio of the model were adjusted until the shape of the peak response to the half-sine acceleration pulse matched available human response data. The results of no-injurious acceleration exposures of volunteer subjects were also considered to verify the frequency response and damping characteristics of the model. Verification was accomplished by study of the relationships between the acceleration input conditions and the measured responses of the test subjects, e.g., acceleration of body segments, displacement of body segments, restraint harness loads, and forces measured between the seat structure and the test subject.

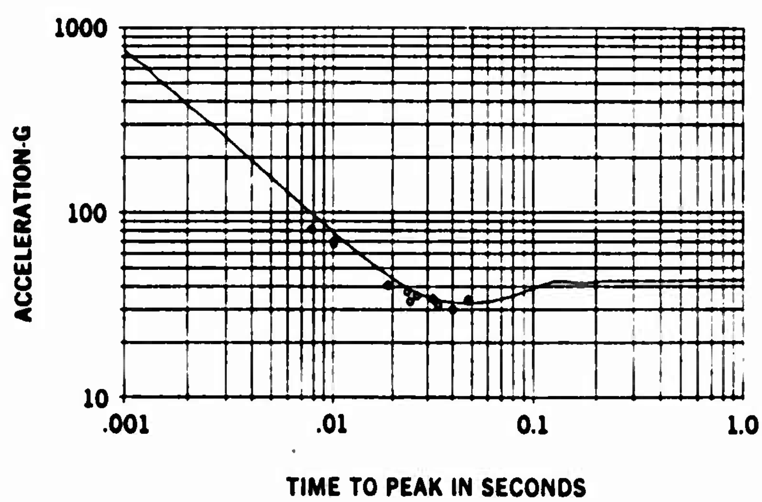

Figure 1 Model Response Curve for +X Axis Half-Sine Acceleration Pulses

Figure 1 shows the model response curve initially fitted to data collected from experiments conducted with the acceleration vector directed primarily along the +X axis. Tests resulting in injury (spinal fractures) or potentially serious sequelae (cardiovascular shock) are designated by the black symbols. The curve was derived from the responses of a mathematical model with a natural frequency of 62.8 rad/sec and a damping coefficient ratio of 0.2. Each of the models that have been developed presumes a specific restraining system consisting of a lap belt, crotch strap, and double shoulder strap configuration.

Figure 2 shows the derived curve and data points for -X axis impacts. The available data points do not demonsrate that the human body can tolerate higher acceleration levels as the duration of the acceleration pulse is decreased. However, this appeared to be a reasonable approximation on the basis of data from tests with animal subjects and analysis of physical responses of volunteer test subjects. A further refinement of the model coefficients was made based on transfer function calculation relating test seat and occupant chest accelerations. The results of 11 tests conducted at a level of 10G (impact velocity of 30.5 ft/sec) were analyzed. The mean natural frequency was found to be 64.2 rad/sec (SD = 6.0) and the mean damping coefficient was 0.23 (SD = 0.07).

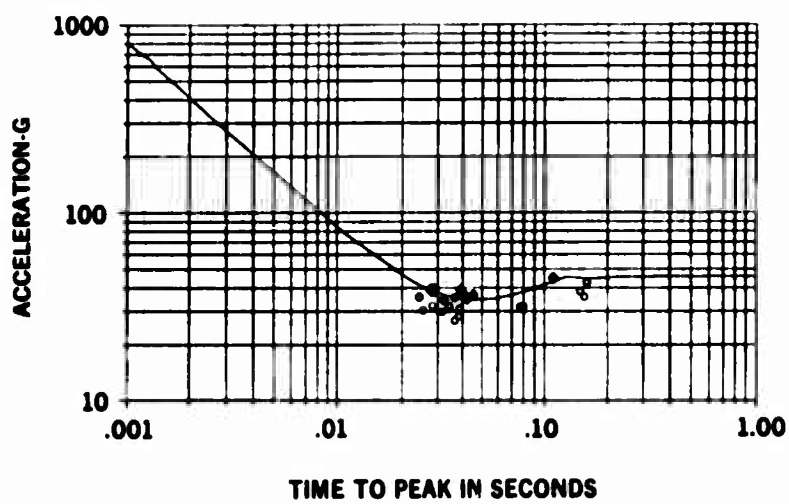

Figure 2 -X Axis Acceleration Response Curve

The injury-risk levels for the +Z axis were assigned by using the 50, 5, and 0.5-percent probability of spinal injury from the injury probability distribution for the DRI reported in reference 4. These injury risk levels are characterized as high, moderate, and low with respective DR values of 22.8, 18 and 15.2, as shown in Figure 3. A 50-percent probability of injury was selected as the highest spinal injury rate

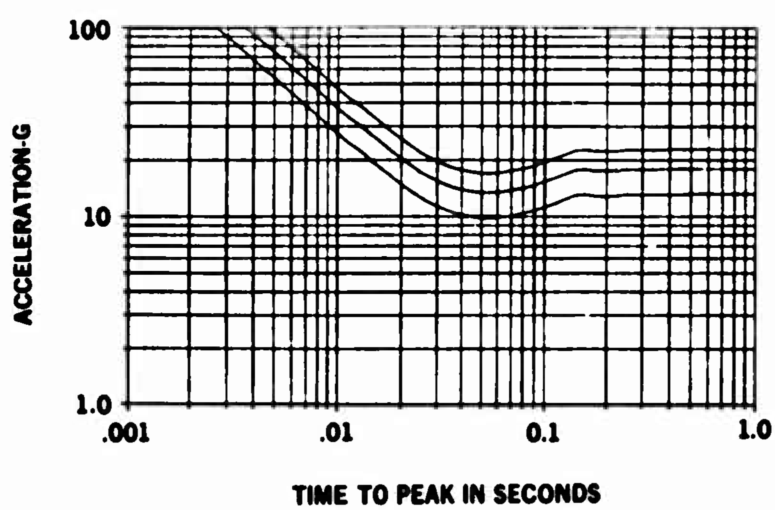

Figure 3 Injury Risk Levels for +Z Axis Half-Sine Acceleration Pulses

acceptable in the system design for two reasons. First, it is the highest spinal injury rate that has been observed for any USAF ejection seat. Second, this level was judged to be the maximum allowable considering that multiple exposures would be likely subsequent to the catapult acceleration, i.e., rocket acceleration, parachute-opening shock, and ground landing impact. The moderate-risk level corresponds to the level used in current USAF ejection-seat design (ref. 5) and is at a mid-point between the high and low levels. The low-risk level corresponds to acceleration conditions used routinely without incident in tests with volunteers conducted by the USAF.

Figure 4 illustrates the injury-risk levels assigned for the -X axis. The DR limit values are 46, 35, and 28.

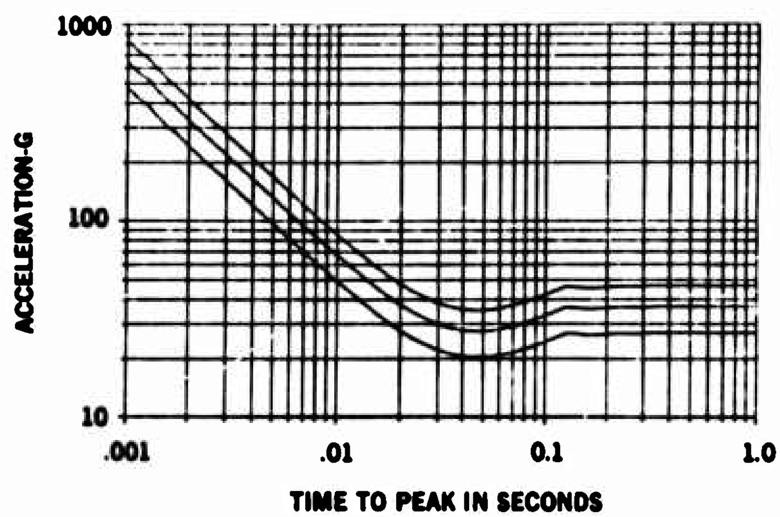

Figure 4 Injury Risk Levels for -X Axis Half-Sine Acceleration Pulses

The methodology used to establish the risk levels produced higher statistical confidence in its application of the +/-X and +Z axes than in its application of the +/-Y and -Z axes since more data are available to define the higher-risk levels. The data used to define the risk levels for the Y axis did not permit the assignment of high-risk levels with an adequate degree of confidence since clear evidence of injury has not been observed under laboratory conditions. The Y-axis model was initially assigned the same coefficients as the X-axis model (ref. 6). But the injury risk levels were lowered to correspond to the levels judged reasonable on the basis of available human test data. A transfer function analysis technique has been used to provide coefficients for the Y-axis model. Using data from 0-G impact tests with 13 volunteers, a mean natural frequency of 50.0 rad/sec (SD = 1.7) was derived. The damping coefficient was 0.09 (SD = 0.04). DR limit values, which have been estimated, are 22, 17, and 14.

The human test data available for the -Z axis, are also limited; however, the acceleration-time histories that have been used in non-injurious tests with volunteers span a relatively large range of time durations. The low-risk level was assigned on the basis of injury-free laboratory tests with volunteer subjects. The moderate level was selected on the basis of previous downward-ejection catapult acceleration limits, and the upper bounds of the available data from tests with heavily restrained subjects were used to establish the high-risk limits although injuries were not observed. The available data were not sufficient to do more than provide a rough approximation of the frequency response range of a model that would be descriptive of human dynamic response to -Z axis acceleration inputs. Since the +Z axis model was in that range, the natural frequency and damping coefficient for the +Z axis model were selected for the -Z axis acceleration limit model. The DR limit values which were selected are 15, 12, and 9.

While the individual orthogonal axes responses were assumed to be independent, a combined risk assessment in the form of an ellipsoidal envelope is proposed. It can be expressed in the following form:

$$\left[\left(\frac{DRX(t)}{DRX_L}\right)^2 + \left(\frac{DRY(t)}{DRY_L}\right)^2 + \left(\frac{DRZ(t)}{DRZ_L}\right)^2\right]^{1/2} \leq 1.0$$

where the suffix L denotes the limiting value for the assigned risk value.

The task of providing criteria for bounding allowable angular acceleration has been a problem that has required an assessment from first principles since no precedent exists. There is very little data available on human tolerance to angular acceleration and velocity. Translational accelerations and angular rates have been measured in only one test where a volunteer was exposed to the combined translational and high-level angular accelerations that may be associated with ejection seat operation (ref. 7). The approach selected to limit the angular acceleration is based on the hypothesis that the injuries associated with angular acceleration are directly related to local linear accelerations. This hypothesis has some support based on the experimental findings of Taraov (ref. 8) and Weiss et al (ref. 9). Thus, the tangential and, to a greater degree, the centripetal acceleration must be considered. Payne has recommended that assessment of the effects of angular acceleration be accomplished by applying to the injury models the net linear acceleration due to whole body translational acceleration plus the local linear acceleration due to body rotation at a body center point. In the application of this method a body center point that is 18.2 in. up and 3.4 in. forward from the seat reference point (the intersection of the planes of the seat, seatback, and the mid-sagittal plane of the seat occupant) was chosen.

III. PREDICTIVE MODELS

The most desirable form for tolerance criteria is one that is specified in terms of external to the body variables. For example, the acceleration of the seat, the impact velocity of an aircraft or the height of a fall can be used to estimate the likelihood of an injury. Unfortunately most injuries associated with aircraft operations do not lend themselves to such simple approaches because the exposure conditions are usually more complicated and better resolution between exposure conditions and injury consequences is needed.

This may be illustrated by considering the case of a helicopter crash in which the net crash deceleration and ground impact velocity may be reasonably estimated, but the degree of anticipated injury can be substantially modified depending on whether the helicopter was equipped with an energy absorbing seat, the seat stayed rigidly attached, the crewmember was in an upright position at time of impact, the harness functioned properly in restraining the crewmember and the aircraft structure was sufficiently deformed to interact with the crewmember. Obviously all these factors complicate an injury potential assessment and in a given crash event any one of them may be the causative factor in an injury or fatality.

While no current method exists that can factor in all such eventualities and provide a meaningful absolute injury probability prediction, analytical models are being developed that can assess the relative effects of system designs, procedures and crash conditions.

One such model has been developed by the USAF to predict stresses developed in the spine due to abrupt accelerations applied to the torso (Ref. 10). This is a three dimensional, discrete model of the human spine, torso and head developed for the purpose of evaluating mechanical response in pilot ejection. It was developed in sufficient generality to be applicable to other body impact problems, such as occupant response in aircraft crashes and arbitrary loads on the head-spine structure.

A graphical representation of the Head-Spine model structure is shown in Figure 5. The anatomy is modeled by a collection of rigid bodies, which represent skeletal segments such as the vertebrae, pelvis, head and ribs, interconnected by deformable elements, which represent ligaments, cartilageneous joints, viscera, and connective tissues. Techniques for representing other aspects of the environment, such as harnesses and the seat geometry, are also included. The model is valid for large displacements of the spine and treats material nonlinearities.

The basic model is modular in format, so that various components may be omitted or replaced by simplified representations. Thus, while the complete model is rather complex and involves substantial computational effort, various simplified models are available that are quite effective in duplicating the response of the complete model within a range of conditions.

Various conditions can be studied using the model, including different acceleration pulses, harness configurations and elasticities, seat geometries and initial spinal postures. Predictions include spinal deformation, local stresses and a thoracic/lumbar compression fracture probability. The latter prediction is based on the predictive calculation of vertebral body stresses during a dynamic exposure event and the comparison of these stresses to measured strength properties of human vertebral bodies (Ref. 11).

The injury prediction is given in terms of an Injury Potential Function which is obtained by dividing the maximum predicted stress at each vertebral level by the corresponding vertebral level mean failure stress. An example of the Injury Potential Function for a fully upright and tightly restrained individual is shown in Figure 6. Four levels of vertically applied acceleration ranging from 14 to 20 g's are considered. A bi-model response is observed with peaks at T8 and L3. The steep increase of the curves at T4 is probably not realistic because of the relatively unstable response of the upper thoracic and cervical spine structure. The higher probability of injury predicted in the middle thoracic region than the lumbar region, which is the more common region for spinal compression fractures, illustrates the model's capability to examine the effects of varying conditions. In this case it showed that a highly upright seated position and restricted column bending due to very tight torso restraint may move the most likely site of spinal injury up the spine.

Another model adapted by the USAF to study gross human body dynamics is the Articulated Total Body (ATB) model (Ref. 12). This model is a derivative of the Crash Victim Simulator (Ref. 13) originally developed to study road vehicle occupant motion during crashes. The model is totally three-dimensional and its analysis method is based on coupled rigid body dynamics. Its standard configuration, consisting of 15 segments and 14 joints, as shown in Figure 7.

A graphical depiction of the model is shown in Figure 8 where the segments are depicted by ellipsoidal surfaces. The graphical display can be presented from any direction and distance and its image projected through a point. Such a graphical display allows direct comparison to video images as well as being a convenient means for examining body position and motion relative to support and potentially interacting structures.

The modeled body structure is assumed to be passive in that muscle forces do not act. The dynamic response of the body can be induced by interactions with the seat and harness system or windblast, gravitational or local contact forces acting on the segments. In addition to the graphical depiction of body motion, the linear and angular displacements, velocities and accelerations of all segments, the forces and moments in all joints, the points on segments and forces of contact and the loads in the harness system are predicted.

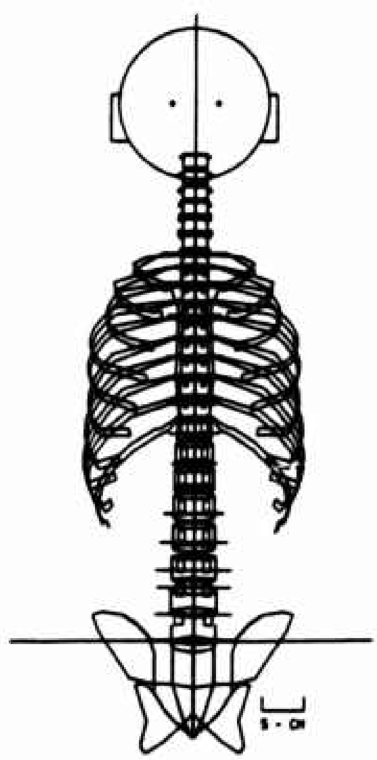

Figure 5 Graphical Representation of the Head-Spine Model

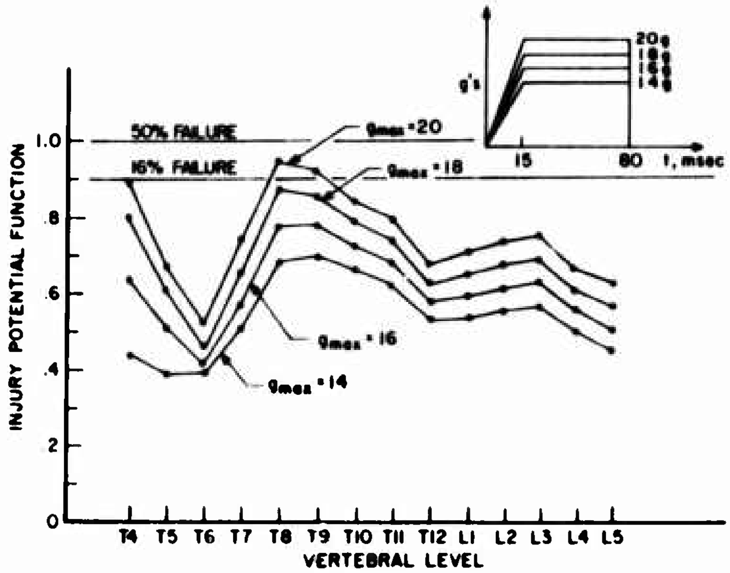

Figure 6 Injury Potential Function from Head-Spine Model in Upright and Highly Restrained Posture for Gz Impacts

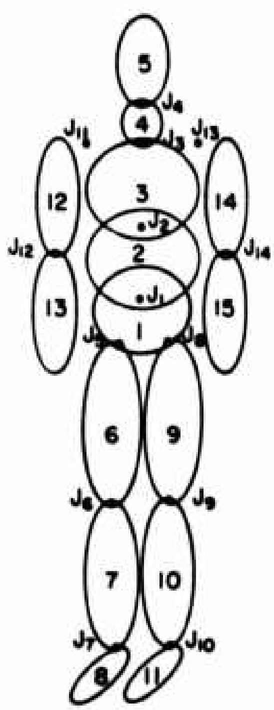

Figure 7 ATB Model Segment and Joint Configuration

Figure 8 Graphical Representation of the Articulated Total Body Model

Various data bases for different size males, females and children (Ref. 14), as well as the Part 572 dummy (Ref. 15) and the Hybrid III dummy (Ref. 16) have been developed. These allow a broad choice of occupant sizes and can be used to investigate effects on dynamic response of size variation.

Human response validation (Ref. 17) and dummy response validations (Ref. 18) have been made with the model. Additionally a number of simulations have been performed with excellent agreement between predicted and observed responses (Ref. 19 and 20).

The ATB model is an excellent tool for clearance or body trajectory prediction. The model was used to investigate child motion in an automobile during panic braking and subsequent impact (Ref. 21). Three child sizes and seven different initial positions were chosen. In Figure 9 a child initially facing forward experiences a .5G vehicle deceleration while his feet have a .25 seat friction coefficient. In Figure 10 a child in the same initial position experiences a .7G vehicle deceleration while his feet have a

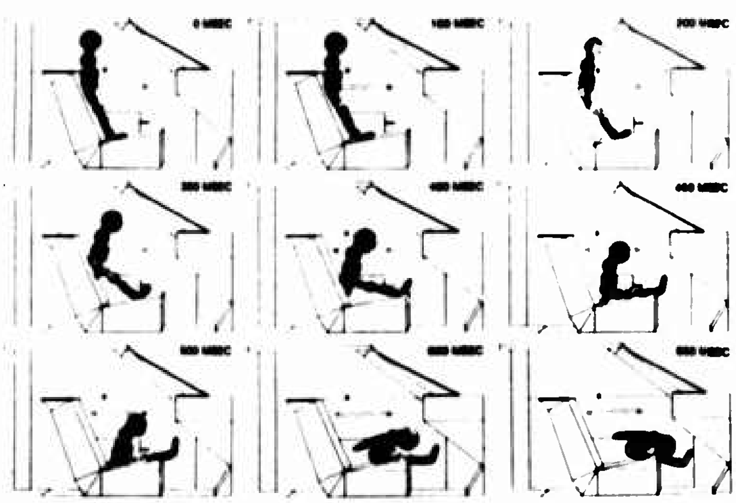

Figure 9 Two-and-One-Half-Year-Old Child Motion During .50 G Panic Braking Deceleration with .25 Seat Friction Coefficient

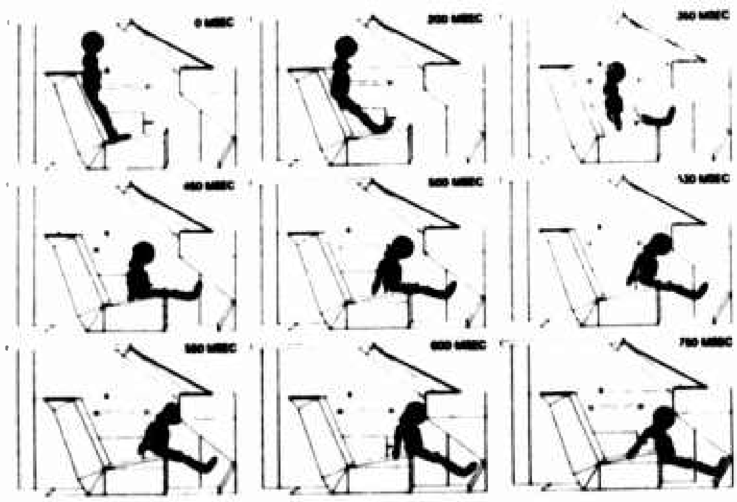

Figure 10 Two-and-One-Half-Year-Old Child Motion During .70 G Panic Braking Deceleration with .20 Seat Friction Coefficient

.20 seat friction coefficient. In the first case the motion is relatively benign with the child not contacting the dash and coming to rest on the front part of the seat. A 20 percent reduction in the seat friction coefficient, combined with an increase in vehicle deceleration from .5G to .7G, substantially modified the resultant body motion leading to a significant head impact with the dash and final body location on the vehicle's floor.

The above simulations are only two of over 160 that were performed in which various conditions and parameters were varied, but they illustrate the ease of examining the relative effects of such changes. The model has also been used in a number of internal USAF studies to look at body motion and clearances of limbs during escape from aircraft.

IV. ADVANCED MANIKIN DEVELOPMENT

The use of mechanical human surrogates or dummies is becoming a more common and relevant approach for assessing the safety of crash protection systems and procedures. Early dummies were developed to provide inertial loading similar to that of the human body and were primarily used to test the proper operation of harnesses, seat structures and ejection seats. In these tests the concern was with the response of the equipment as affected by the inertial effects of the human body. Typical dummies used for such applications were developed by Sierra and Alderson in the 1950s primarily to provide human-like ballast for ejection seats. While their overall mass distribution properties were quite good, their joint mobility and body flexibility were highly limited. This resulted in a highly rigid responses to external forces and internal dynamic measurements that did not compare well to human responses for similar exposures (Ref. 22).

A new generation of dummies was developed in the 1960s and 1970s, primarily driven by increased emphasis on road motor vehicle safety. The most common of these is the Hybrid II dummy originally developed by General Motors and adopted by the National Highway Traffic Safety Administration as the standard automotive safety compliance testing dummy. This dummy, most commonly known as the Part 572 dummy from its documentation designation, had considerably improved human-like fidelity and was designed to provide internal response measures that could be correlated to equivalent human responses and possibly, likelihood of injury. Several other dummies were developed in the United States, Great Britain and Sweden in this same time period that attempted to improve response characteristics, but none achieved the degree of standard acceptance as had the Part 572 dummy. In the late 1970s General Motors developed the Hybrid III, which had improved bio-fidelity and instrumentation capability over the Hybrid II.

This evolutionary process did improve the state-of-the-art in dummy design sophistication, biofidelity and response measurement capability. Most of it, however, was directed at road vehicle safety design considerations with considerable emphasis on chest and head impact responses, horizontal plane impact events and testing under highly controlled conditions.

Attempts to use these types of dummies in aerospace environments led to the identification of a number of shortcomings. These included the lack of a proper dynamic longitudinal spinal axis response, which is the predominant loading direction for aircraft related force exposures. Standard data retrieval by means of an umbilical cord limiting freedom of dummy motion and requiring a separate data acquisition system. Durability sufficient only to withstand relatively low forces compared to those encountered in aircraft crashes or escape from aircraft.

To address these shortfalls, the US Air Force has pursued the development of an Advanced Dynamic Anthropomorphic Manikin (ADAM) to be used in the testing of escape systems and various protection systems and procedures (Ref. 23). This effort, initiated in 1981 has resulted in the production of a small and a large male prototype manikin. These manikin sizes are based on an Air Force male aviator anthropometric survey conducted in 1967 (Ref. 24) with the specific dimensions and inertial properties taken from US tri-service recommendations (Ref. 25). The small ADAM whith full skin covering is shown in Figure 11. The small ADAM with upper torso, right arm and right leg skins removed to show the mechanical structure, battery storage compartment in upper leg and the instrumentation package located in the thorax, is shown in Figure 12. Also shown in Figure 12 is a head mounted antenna used for data transmission.

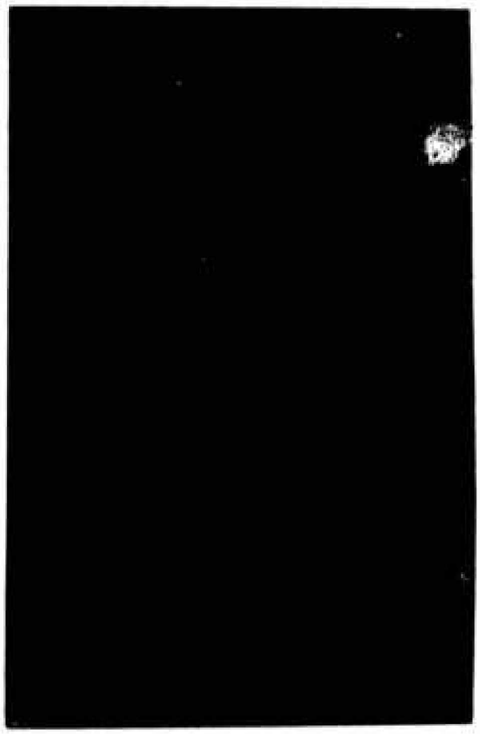

Figure 11 Small ADAM with Full Skin Covering

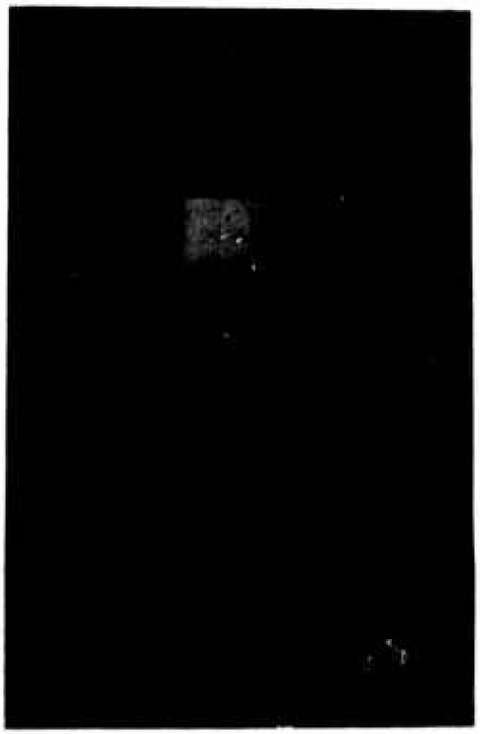

Figure 12 Small ADAM with Skins Partially Removed to Show Mechanical Structure and Instrument Package in Thorax

The design for this manikin stresses faithful human joint articulation and torso axial deformation to properly reflect the mass shifts and limb motion, as well as dynamic spinal compression, that an actual crewmember would experience during a whole body impact. All the joints with the exception of the neck and spine, are single or compound revolute joints with precisely defined axes orientations, joint stops with soft snubbers, adjustable friction pads and position sensing potentiometers. These features can be seen in Figures 13 and 14 which are the knee and shoulder joints respectively. The axial spine element is a combined spring and hydraulic damping element which is tuned to provide longitudinal impact response with a natural resonance in the 10 to 12 Hz range. Re-tuning may be accomplished by spring replacement and use of different viscosity hydraulic fluid. Below the axially deforming spinal element is a universal joint that allows for yaw motion and flexural and lateral bending. This compound articulation is approximately in the lumbar anatomical region and provides the only bending articulation in the torso. The total spine structure is shown in Figure 15.

Figure 13 ADAM Knee Joint Showing Flexure and Torsion Articulations, Friction Pads and Wire Connections to Position Measuring Potentiometers

Figure 14 ADAM Shoulder Joint Showing Multiple Revolute Articulations

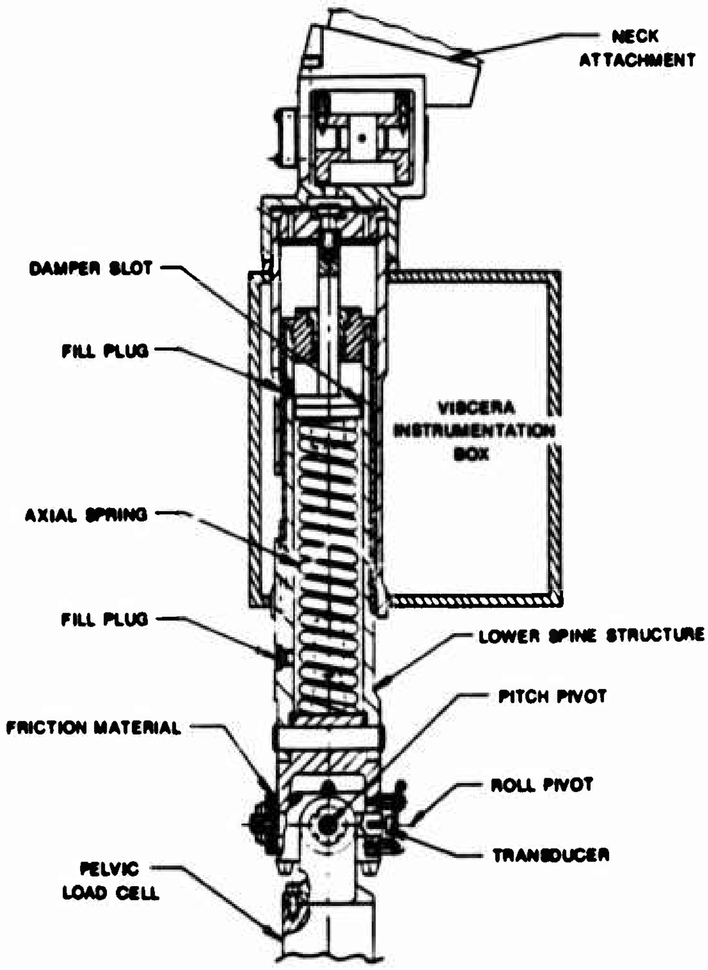

Figure 15 ADAM Spine

The durability of the manikins was specified in terms of a violent exposure environment in which they must be able to operate without any functional degradation. This environment is the ejection into a 700 KEAS wind stream, in an ejection seat with unrestrained limbs for the large manikin and restrained arms and legs for the small manikin. They must also be able to sustain 45G impact loads in the Gx, Gy and Gz directions without functional degradation or permanent structural damage.

The standard Hybrid III dummy head and neck are used, but, as opposed to the Hybrid III design, the head skin covering extends over the neck as can be seen in Figure 11.

The total instrumentation and data acquisition system for ADAM is a substantial advancement over any other current manikin. The system is located in the thorax, is computer controlled and, in its standard configuration, can collect 128 channels of data at 1000 samples/channel-sec and store up to 4 seconds of data as well as telemeter this data in real time to a ground station. The system configuration can be modified by an operator, through computer input, to change the number of channels, the sampling rate and the filter bandwidths. The circuit board configuration, from a rear view, is shown in Figure 16.

Figure 16 ADAM Instrumentation Package with Rear Access Panel Removed

The availability of 128 channels allows extensive monitoring of the manikin's responses as well as collection of external data. ADAM has been designed for measurement of three orthogonal acceleration components in the head, thorax and pelvis; six force and moment components both between the head and the top of the neck and between the lumbar spine and the pelvis; and the position of all revolute joints. Additionally, load cells are located at the joints in the lower legs to measure torsional moments. The instrumentation system provides for signal conditioning, analog to digital data conversion and pre and post data collection calibration for all of these transducer channels. A listing of these transducer channels, including ones for internal temperature and parachute riser loads, are presented in Table 1. The other channels may be used for additional ADAM sensors or to collect external information.

TABLE 1
ADAM TRANSDUCER CHANNELS

1	Left Hip Abduction/Adduction Position
2	Right Hip Abduction/Adduction Position
3	Left Hip Flexion Position
4	Right Hip Flexion Position
5	Left Hip Medial/Lateral Position
6	Right Hip Medial/Lateral Position
7	Left Knee Flexion Position
8	Right Knee Flexion Position
9	Left Knee Medial/Lateral Position
10	Right Knee Medial/Lateral Position
11	Left Shoulder Arm-Joint Abduction/Adduction Position
12	Right Shoulder Arm-Joint Abduction/Adduction Position
13	Left Shoulder Thoracic-Joint Abduction/Adduction Position
14	Right Shoulder Thoracic-Joint Abduction/Adduction Position
15	Left Shoulder Flexion/Extension Position
16	Right Shoulder Flexion/Extension Position
17	Left Shoulder Medial Lateral Position
18	Right Shoulder Medial/Lateral Position
19	Left Arm Raising/Lowering Position
20	Right Arm Raising/Lowring Position
21	Left Elbow Flexion Position
22	Right Elbow Flexion Position
23	Left Forearm Supination/Pronation Position
24	Right Forearm Supination/Pronation Position
25	Left Lower Leg Torque
26	Right Lower Leg Torque
27-32	Neck Forces and Moments (6 axis)
33-38	Lumbar Forces and Moments (6 axis)
39-41	Head Accelertion (triaxial)
42-44	Head Rotation Rate (3 rate sensors)
45-47	Chest Acceleration (triaxial)
48-50	Pelvis Acceleration (triaxial)
51-52	Parachute Loads, Right and Left Risers
53	Temperature Measurement
54-58	Lumbar Position
59	Viscera Position
60	Viscera Acceleration
61	Sternoclavicular Elevation/Depression Position
62	Sternoclavicular Pronation/Retraction Position

V. CONCLUSIONS

It is suggested that a comprehensive assessment of injury potential in a crash or other exposure to violent mechanical forces requires a broad based methodology including direct injury prediction based on environmental conditions; analytical or modeling approaches that an provide interpolative, extrapolative and sensitivity information; and the use of mechanical surrogates that can provide direct measures of what the human body would experience in a given environment.

The direct tolerance criteria, while usually the easiest to use, often have limited utility because they apply to very specific modes and mechanisms of injury. For example, the DRI as specified for evaluating allowable ejection seat accelerations (Ref. 5) is only applicable to Z axis accelerations and is strictly based on observed spinal compression fractures as the injury mechanism.

Analytical methods and models can provide a means of relating a general body exposure to a specific susceptible body structure and, given the appropriate strength properties of the local structure, allow prediction of failure/injury of that particular structure. From an engineering point this is a straight forward process; however, in application it can be difficult due to the large variability in biological material properties, the structural complexity of the human body and the generally unknown extent of active muscle participation. Where the modeling methodology is particularly useful is in relative assessments of system design changes, procedure modifications and operation condition variations. The assessments are made on the basis of predicted changes in local stresses, deformations and accelerations; external contact and harness forces; maximum ranges of limb motion; and amount of clearance between body segments and structural elements.

To define the exposure environment, tests must be conducted that, as closely as possible, replicate the anticipated operational conditions. This not only requires that the vehicle or occupants external environment is properly controlled, but that the occupant be an appropriate surrogate for a crewmember. The ADAM has been designed not only to provide correct reactive loads into the harness, seat and any other interactive structures, but to also be sufficiently internally biofidelic so that its internal response measures may be related to equivalent human responses under the same exposure conditions. While substantial validation still needs to be performed, the biofidelity and extensive response measurement capability of ADAM should make it a powerful tool for the safety assessment of aircraft, subsystems and procedures.

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