BIOMECHANICALLY BASED PERFORMANCE CRITERIA FOR CRASHWORTHY AIRCRAFT SEATS

by

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Introduction

One of the most critical problems in the protection of aircraft occupants in a crash environment is posed by vertical acceleration levels which may exceed those that the human body can withstand without injury in a direction parallel to the vertebral column. Because sufficient crush space does not exist within the floor structure of helicopters and light aircraft for adequate energy absorption, the seat must play a significant role in attenuating these potentially injurious forces to tolerable levels. Several recently designed aircraft have been equipped with seats that incorporate systems for the absorption of energy in the vertical direction. The seats are designed to comply with criteria that were developed and documented in 1971, and although these seats are far superior to any prior systems, there are several areas of uncertainty in the design criteria that require additional research to enable further progress to be made in development of future systems.

A primary problem is the limited extent of knowledge concerning human tolerance to $+G_z$ acceleration. In fact, little new information concerning human tolerance to acceleration in this direction has been developed in many years. Extensive effort has been expended on the critical areas of human head and neck response and the effects of restraint system variables on acceleration loads in the forward, $-G_x$, and lateral, G_y, directions. However, essentially no emphasis has been placed on the vertical direction, which is critical to survival in many aircraft accidents. Although aircraft occupants can withstand the full 95th-percentile survivable crash acceleration conditions in the lateral and longitudinal directions with no energy absorption, but only proper restraint, the human body cannot tolerate the vertical impact forces typical of a severe accident without energy absorption.

Design principles and criteria for aircraft seats and restraint systems are presented in the Aircraft Crash Survival Design Guide (1)*, which was originally published in 1967 and has been updated several times. The most recent revision includes results of more than a decade's research and design experience. It describes desirable strength and deformation characteristics, material selection, attachment to the airframe structure, cushion properties, energy-absorbing mechanisms, restraint system configurations and characteristics, and means for system evaluation by analysis and testing.

Criteria for crashworthiness are also concisely stated in a U.S. military specification, MIL-S-58095 (2). Key requirements pertaining to crashworthiness are minimum static strengths and limits on the seat's load-deformation characteristics in the longitudinal, lateral, and vertical directions. Also, it is

*Numbers in parentheses indicate references listed at end of paper.

specified that provision must be made for energy absorption in the vertical direction, and the energy-absorbing mechanism must protect occupants ranging in size between the 5th and 95th percentiles from experiencing accelerations exceeding human tolerance.

An obvious problem in designing for the two extremes of occupant weight is to provide sufficient stroking distance for the 95th percentile while ensuring that the 5th percentile, utilizing a shorter stroke distance, will not suffer excessive acceleration. The specified compromise sets the energy absorber limit load for a static load factor of 14.5 G, based on the combined weight of the 50th-percentile occupant and the movable part of the seat.

The performance of a seat designed in accordance with MIL-S-58095 is evaluated by means of six static tests and two dynamic tests. Both dynamic tests utilize a 95th-percentile anthropomorphic dummy and are conducted according to the conditions illustrated in Fig. 1. Both demonstrate the structural integrity of the seat under simulated crash conditions. Furthermore, in the first test the energy-absorbing mechanism is required to maintain the acceleration measured on the seat below a specific level, illustrated in Fig. 2, which is based on data contained in Ref. 3. For example, the vertical acceleration component should be less than 23 G for durations in excess of 0.006 sec.

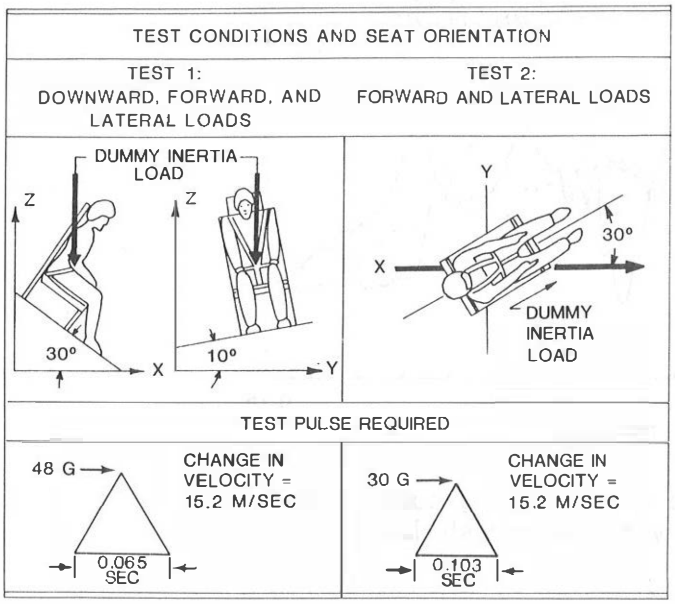

Fig. 1. Dynamic test requirements (2).

In tests of seats designed since development of these criteria, a characteristic seat pan z-axis acceleration response has been observed. In this characteristic curve, illustrated in Fig. 3, the seat pan acceleration rises sharply during the onset of the input pulse, then drops rapidly, sometimes passing through zero. It then rises sharply in a secondary spike before damping out around the load factor used in the design of the energy-absorbing system. In most of the tests conducted during the time period between 1971 (when the criteria were established) and the present, the secondary spikes have exceeded the criteria limit of 23 G and have been a source of concern. One question of concern is whether the secondary spike is a natural response of the seat and occupant dynamic system, or if it is caused by some external source. Also, it is not known whether the acceleration spike is hazardous to the seat occupant.

Simple analyses of the test data have indicated that the secondary spike is a natural response of the seating system and, in itself, does not increase the hazard to the occupant. However, the analyses have also suggested that the crash pulse might be hazardous to the occupant at times other than during the secondary spike. It was concluded that the acceleration-limiting criterion as

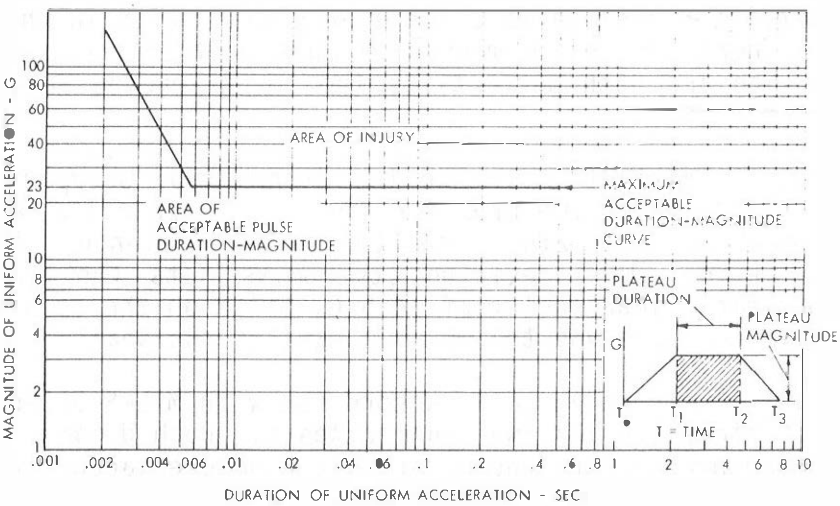

Fig. 2. Maximum acceptable vertical seat acceleration (3).

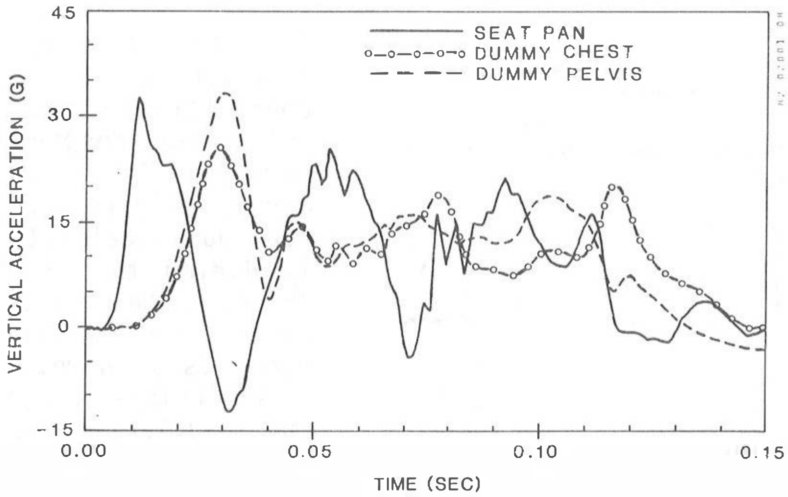

Fig. 3. Typical response of seat pan, dummy chest, and dummy pelvis to vertical crash loading.

it now exists is not sufficiently comprehensive, and that additional research should be initiated to enable development of improved criteria.

A research program was initiated to meet the following objectives:

- Establish the sensitivity of seat and occupant response to system variables.

- Determine the effect on system performance of the type of dummy being used for testing and establish an appropriate standardized dummy for aircraft seat system evaluation.

- Investigate the performance of the seat with occupants more nearly representative of the human occupant than are anthropomorphic dummies, i.e., human cadavers.

- Establish, through dynamic testing, additional information concerning human tolerance to acceleration loads in the $+G_z$ direction.

The purpose of this paper is to summarize the status of this overall effort and to present some preliminary information available from these current programs.

Typical System Response

An energy-absorbing aircraft seat generally consists of a relatively rigid frame attached to the aircraft structure and a movable bucket in which the occupant is seated and restrained. Because of space limitation, the motion of the bucket relative to the frame is usually restricted to the vertical direction, where forces in excess of human tolerance are likely to be present. Relative motion in the vertical direction is controlled by an energy-absorbing device which is designed to stroke under a predetermined constant force. In order to achieve low weight in the overall system, practical seat energy absorbers utilize plastic deformation of metal as their primary operating mechanism.

An impact applied at the base of the seat initially decelerates only the movable seat mass, as the uncompressed cushion and body, illustrated in Fig. 4(a), cannot immediately support loads. The resistive force designed into the energy-absorbing system is based on the total mass of seat and occupant. Therefore, when it is initially applied to the seat mass alone, the seat acceleration significantly exceeds the limit load factor for which the system was designed, as shown in Fig. 3. Meanwhile, the downward velocity of the body segments relative to the seat increases until the springs associated with the cushion, buttocks, and torso are compressed sufficiently to decelerate the segments above them. At this point, illustrated in Fig. 4(b), the accelerations of the body segments reach their maximum values. The net upward force on the seat is reduced at this time, perhaps even below zero, so that the seat acceleration is at a minimum value. Unloading of the occupant then releases the downward force on the seat, and the seat acceleration climbs to a second peak. Eventually, the velocities of the masses in this dynamic system, the body segments and the seat, approach each other, as shown in Fig. 4(c). A steady-state level of acceleration, near the design load factor of the system, is attained.

It is important to note that the peak accelerations of the seat pan do not necessarily coincide with peak decelerations of the occupant pelvis or chest, and thus are not necessarily hazardous to his safety. However, the criterion based on the Eiband tolerance data does not consider the seat pan acceleration excursions from the average. Therefore, more comprehensive criteria are needed.

Summary of Current Research Programs

As part of this effort to develop more rigorous criteria, a number of dynamic tests have been conducted using both anthropomorphic dummies and human cadavers, in both rigid and energy-absorbing seats. The 75 dummy tests have been conducted at four different facilities, two with horizontal sleds and two with vertical drop towers. Variables that are being investigated in the dummy tests

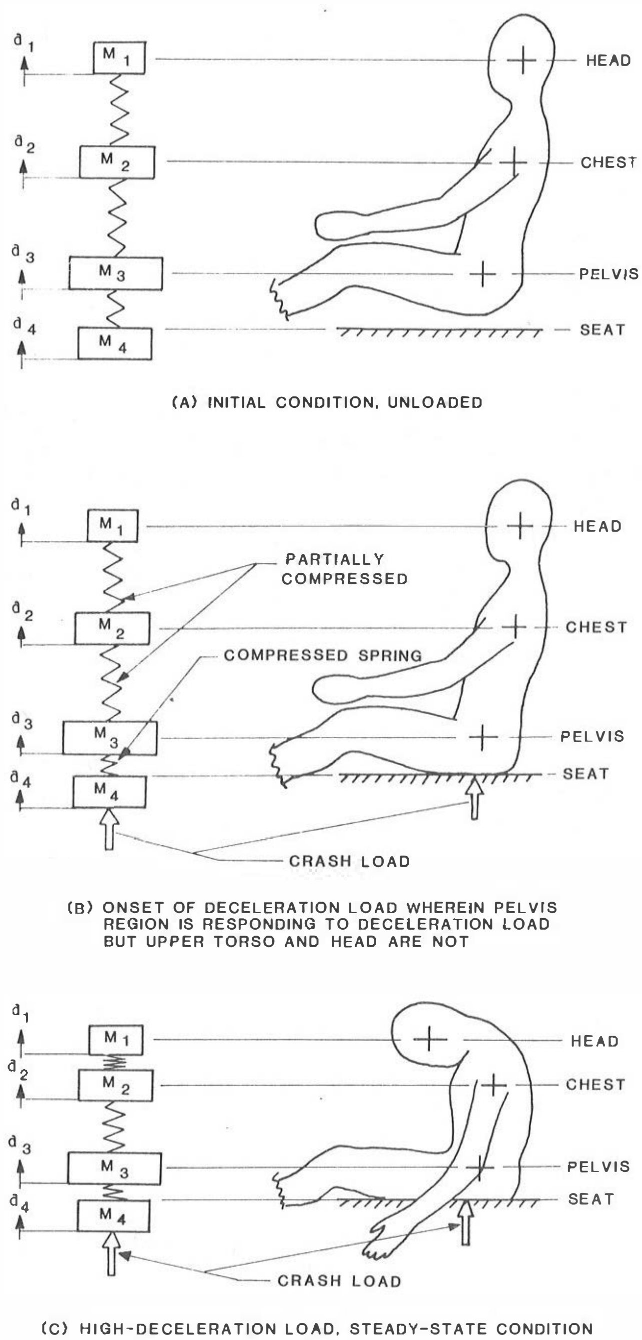

Fig. 4. Spring-mass representation of seat-occupant system.

include the shape and magnitude of the input deceleration pulse, the velocity change, the type and size of the dummy, the energy absorber limit load, the seat weight, the cushion characteristics, the seat orientation, and the structural spring rate of the seat. The baseline configuration for the dummy tests that use the energy-absorbing seat is illustrated in Fig. 5.

The cadaver tests are being conducted using the same seats, with two seat orientations and three energy absorber limit loads. An initial series of tests was conducted using a rigid seat, in which the vertebral column of the cadaver was aligned with the impact vector. Each of the three cadavers was tested repeatedly while the peak input acceleration was increased in increments of 2 G, until a vertebral fracture was observed by posttest x-ray. Fractures occurred at levels of 8, 13, and 30 G. Then a second phase of the program was initiated, using a production energy-absorbing seat, the crewseat for the UH-60A Black Hawk helicopter. This phase of the program consisted of nine tests divided into two series. The first series consisted of a maximum of two dynamic tests with each cadaver. The first test was conducted in a configuration similar to that illustrated in Fig. 5, with the seat back oriented within 4 degrees of the impact vector. Applying a $+G_z$ acceleration to the occupant, this configuration is referred to as "vertical." In the second test the seat was pitched an additional 17 degrees forward. Applying both $-G_x$ and $+G_z$ acceleration to the occupant, this second configuration is referred to as the "combined" orientation. The standard 14.5-G energy absorber limit load was used in all these tests.

Two cadavers successfully passed the first test orientation, but received vertebral fractures in the combined orientation; two other cadavers received

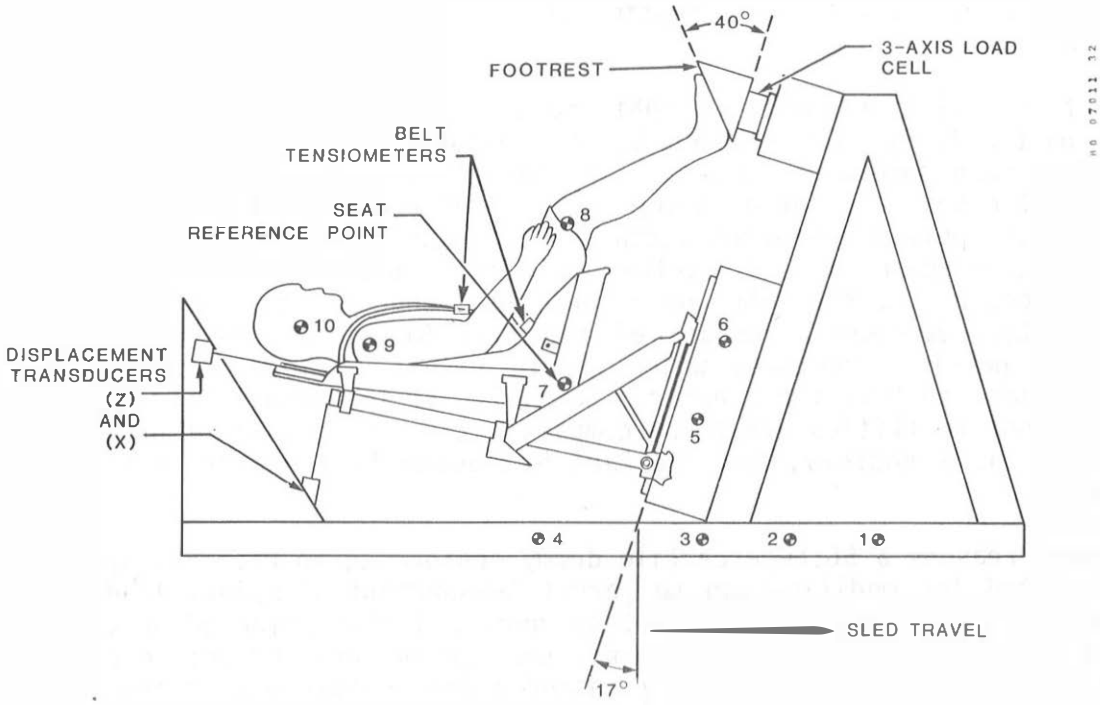

Fig. 5. Test configuration for horizontal sled tests with energy-absorbing seats.

fractures in the first test. For the remaining three tests in that phase of the program, 11.5-G energy absorbers were used, with the seat in the more severe combined orientation. All three of these subjects also received vertebral fractures.

The second phase of the program, which is being conducted in 1982, includes six tests in the combined orientation and 8.5-G energy absorbers. Three tests have been completed in this phase with one vertebral fracture occurring. Details of the tests in this part of the overall program can be found in Ref. 4. It should be emphasized that bone strength for living humans, especially the aviator population, is significantly higher than for the cadaver population tested. Determination of the bone strength relationship between the cadaver and aviator populations, supported by recent operational experience with energy-absorbing seats, will justify use of a higher limit load than the threshold determined in this program.

Modification of Dummies for Spinal Load Measurement

Analyses of the data from the testing program described above demonstrated that the current criterion for seat evaluation based on seat pan acceleration is not sufficiently sensitive or repeatable to provide a measure of hazard to the occupant. Spinal forces and moments, which are the vertebral-injury causing mechanisms, need to be measured directly during the test.* Therefore, a dummy

*Fatalities in aircraft accidents can have many causes. However, nonfatal irreversible trauma generally involve the vertebral column. Therefore, spinal injury tolerance is a critical factor in evaluation of a protective system's performance.

to be used in aircraft seat testing should include capability for measurement of such loads.

None of the existing anthropomorphic dummies has been designed for vertical impact, particularly with respect to the spinal column, which is a critical region for human tolerance to impact in that direction. At present, the Part 572 dummy is the best available choice. The reinforced rubber cylinder used as its lumbar spine permits more consistent positioning than the steel ball-and-socket configuration used in some earlier dummies. Instability in the earlier type could affect response of the upper torso with concomitant penalties on test repeatability. Another advantage of the Part 572 dummy for seat testing is a humanlike pelvic structure, which should result in load distribution on the cushion close to that for a human. Secondly, if the results of tests conducted at different facilities are to be compared, standardization of dummies and test procedures is mandatory, and the Part 572 dummy is the only existing standard device.

For these reasons a 50th-percentile dummy conforming to Part 572 specifications was selected for modification to permit measurement of spinal loads. An adapter was designed and manufactured to permit installation of a six-axis load cell at the base of the lumbar spine, protruding into the pelvic accelerometer cavity. The transducer, actually intended for measurement of femur loads, has axial, shear, and bending capabilities of 30,000 N, 15,000 N, and 350 N-m, respectively.

Because a 95th-percentile dummy is required by present seat testing specifications (2), a second dummy was also modified. The particular dummy selected was that most nearly resembling the Part 572 design, an Alderson VIP-95, with a pelvic structure and lumbar region of Part 572 configuration, and with an elastomeric spine.

Although loads and moments in the lumbar spine are of primary interest in the program being described here, measurement of forces and moments in the cervical spine are also of interest. First, based on cadaver testing, tolerable limits for both shear and bending of the neck have been developed (5). Installation of one of the six-axis load cells in the dummy neck segment was planned in order to provide forces and moments for direct comparison with these values. Such data should be particularly useful in evaluation of various helmets, whose weight may increase the potential for neck injury. A second reason for measuring neck loads would be to obtain a greater understanding of overall occupant response to loads transmitted by the seat. Such information would assist in validation of mathematical models of seat/occupant response.

Unfortunately, the Part 572 dummy (50th-percentile) does not have sufficient space available at either end of the neck segment to install a 6.35-cm-long load cell without significantly altering the head-neck length. The VIP-95 dummy, on the other hand, has a tubular steel section in the lower end of the neck. Adapters were designed to allow replacement of this element by a transducer.

Details of the modifications to the dummies and their testing are presented in Ref. 6.

Summary of Results

Although a complete presentation of the results of more than 100 tests cannot be offered here, some of the more interesting observations pertaining to the relationships between cadaver and dummy measurements, and their implications in development of new seat criteria, will be discussed.

Comparison of Test Facilities

The four facilities conducting the tests were given specifications for their baseline conditions similar to those imposed by present seat test criteria. The baseline deceleration pulse for the Federal Aviation Administration Civil Aeromedical Institute (CAMI), where most of the dummy tests were conducted, is compared in Fig. 6 with the deceleration pulse for a comparable test at Wayne State University (WSU), where the cadaver tests were run. The average impact velocity was 12.8 m/sec at CAMI and 12.6 m/sec at WSU. Although the peak sled deceleration at CAMI was higher, 40.7 G versus 36.1 G at WSU, the rate of increase of deceleration was lower, 980 G/sec versus 1700 G/sec.* Tables 1 and 2 present results from the two facilities, using the same dummy, in the configuration illustrated in Fig. 5 and under the conditions described above. Although significantly greater seat stroke was achieved at CAMI, the z-component of seat acceleration, which is used in the present test criterion, appears quite similar at the two facilities. Within the dummies, however, differences in response can be noted. CAMI tests resulted in higher lumbar loads and higher accelerations in the x-direction, along with lower z-accelerations. The seat pan Dynamic Response Index (DRI), which is computed from the response of a single-degree-of-freedom, damped, spring-mass model of the human torso that has been correlated to ejection seat injuries (7, 8), is slightly higher for the CAMI data.

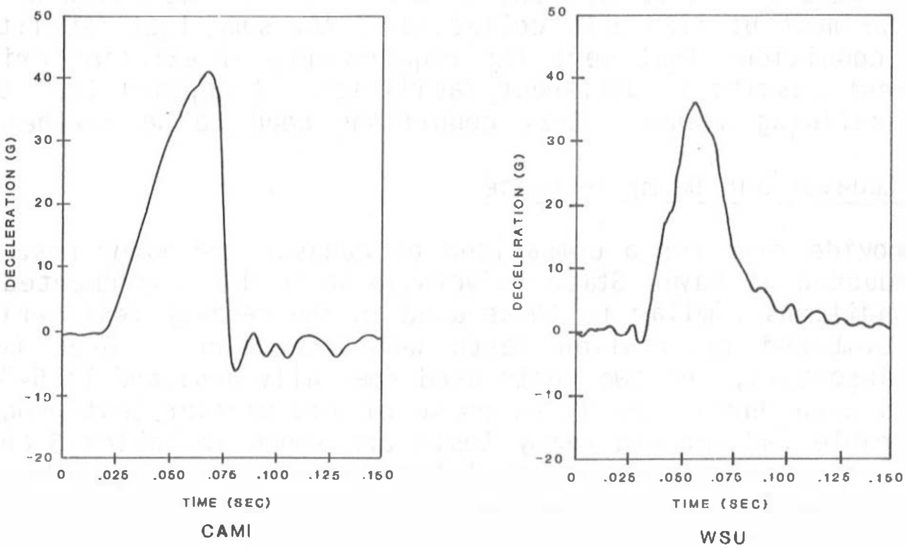

Fig. 6. Typical baseline deceleration pulses for two test facilities.

*The rise time for the deceleration was computed using a straight line connecting points on the plotted waveform at 10 and 90 percent of the peak deceleration, as described in Ref. 1, pp. 194-95.

TABLE 1. COMPARISON OF SEAT AND OCCUPANT RESPONSE FOR BASELINE TESTS AT TWO TEST FACILITIES

Test Facility	Seat Stroke (in.)	Peak Acceleration (G) Seat Pan x	Seat Pan z	Pelvis x	Pelvis z	Chest x	Chest z	Head x	Head z	Duration of Seat Pan z Acceleration at 23 G (sec)	DRI
CAMI	10.6	25.1	26.4	30.5	30.5	18.1	30.4	31.5	36.5	0.007	21.0
	10.5	26.8	29.4	25.1	30.8	24.1	25.8	38.5	36.5	0.016	19.7
WSU	6.3	15.3	28.9	13.5	39.7	13.3	41.3	21.8	39.6	0.011	18.4

TABLE 2. COMPARISON OF LUMBAR FORCES AND MOMENTS AT TWO TEST FACILITIES

Test Facility	Axial Force (N)	Shear Force (N)	Moment (N-m)
CAMI	5,410	1,600	143.0
WSU	3,850	1,280	68.5

Examining these results, it is apparent that any direct comparison of dummy and cadaver response must utilize data collected at the same test facility. Also, because input conditions that meet the requirements of existing criteria can produce different results at different facilities, it appears that tolerances on parameters defining standard test conditions need to be further reduced.

Comparison of Cadaver and Dummy Response

In order to provide data for a comparison of cadaver and dummy response, five tests were conducted at Wayne State University with the instrumented Part 572 dummy under conditions similar to those used in the cadaver test series. Both vertical and combined orientation tests were conducted. Three tests used 14.5-G energy absorbers, and two tests used specially designed 11.5-G devices, which were also used during the first phase of the cadaver test program. Results of comparable cadaver and dummy tests are shown in Tables 3 and 4. Resultant body accelerations are presented for comparison because the accelerometer orientation in the cadaver does not necessarily correspond to the standard dummy coordinate system.

Seat stroke values presented in Tables 3 and 4 indicate that, in general, the Part 572 dummy requires slightly less stroke distance than does the cadaver. The seat pan vertical accelerations, presented in Figs. 7 and 8 for the vertical and combined tests, respectively, show that the interaction between the

TABLE 3. COMPARISON OF SEAT AND OCCUPANT RESPONSE FOR DUMMIES AND CADAVERS

	Seat Stroke (in.)	Peak Acceleration (G) Seat Pan x	Seat Pan z	Pelvis Resultant	Chest Resultant	Head Resultant	Duration of Seat Pan z Acceleration at 23 G (sec)	DRI
Dummy, Vertical	6.3	15.3	28.9	40.3	43.4	41.7	0.011	18.4
Cadaver, Vertical	7.6	17.5	26.5	33.9	N/A	49.7	0.004	21.8
Dummy, Combined	4.5	27.8	23.9	35.2	31.6	46.9	0.004	17.8
Cadaver, Combined	4.5	37.3	25.4	22.8	N/A	59.6	0.004	22.2
	5.5	40.5	21.0	44.4*	N/A	97.3*	0.	19.3

*Impact between mouth-mount accelerometer and thigh.

TABLE 4. COMPARISON OF SEAT STROKE FOR CADAVERS AND DUMMIES IN THE UH-60A CREWSEAT

Test Description	Cadaver Tests Occupant Weight (lb)	Cadaver Tests Seat Stroke (in.)	Dummy Tests 50th% Part 572 Occupant Weight (lb)	Dummy Tests 50th% Part 572 Seat Stroke (in.)
14.5-G E/A, Vertical Orientation, 42-45 G Peak Input Acceleration	166 160 140 148	7.6 7.4 7.1 7.1	164 164	6.3 7.0
14.5-G E/A, Combined Orientation, 42-45 G Peak Input Acceleration	166 140	5.5 4.5	164	4.5
11.5-G E/A, Combined Orientation, 42-45 G Peak Input Acceleration	218 141 160	9.4 7.0 9.0	164	6.5
8.5-G E/A, Combined Orientation, 42-45 G Peak Input Acceleration	200 143	13.1 9.7		

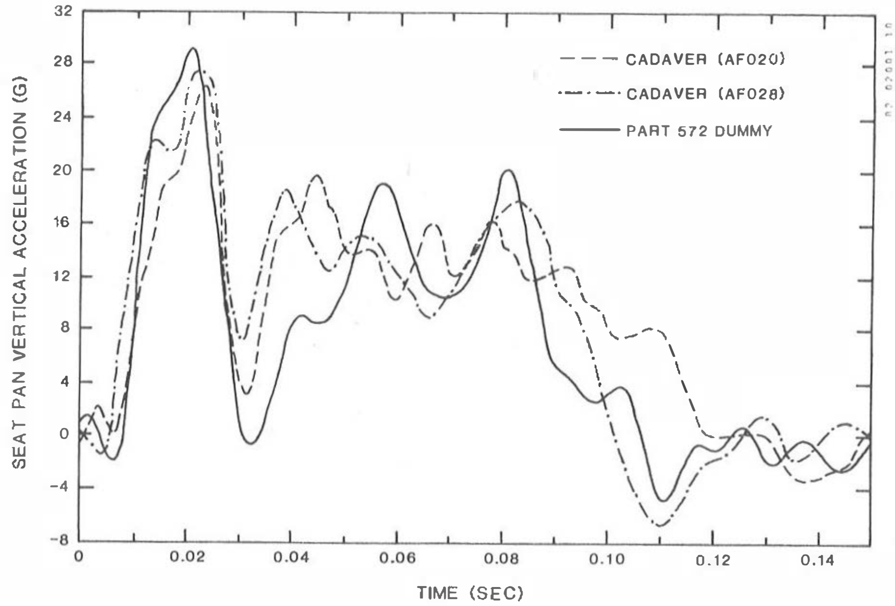

Fig. 7. Seat pan vertical acceleration for a Part 572 dummy and two cadavers measured in vertical mode tests with 14.5-G energy absorbers.

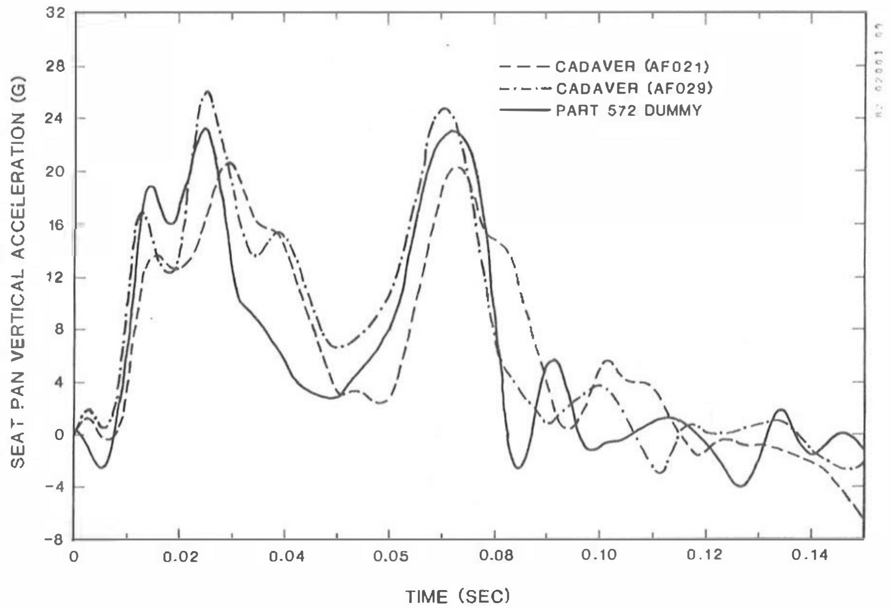

Fig. 8. Seat pan vertical acceleration for a Part 572 dummy and two cadavers measured in combined mode tests with 14.5-G energy absorbers.

Part 572 dummy and seat pan is very similar to the response measured with human cadavers. The comparison between body accelerations for the dummy and cadaver, however, does not show a good correlation. Results of this limited comparison seem to indicate that seat performance criteria based on seat pan acceleration may not be as sensitive to occupant type as a criterion based on body segment acceleration. However, it may also indicate that injury mechanisms within the body, e.g., spinal deformation, cannot be reliably predicted from seat pan acceleration, as internal body response can vary significantly for various occupant types with similar inputs from the seats.

The moment-rotation characteristics for the VIP-95 neck with the load cell installed were measured statically at CAMI, and the results are compared in Fig. 9 with a flexion response envelope which was proposed in Ref. 5 as a result of human volunteer and cadaver tests. It appears that the dummy neck response characteristics provide an adequate human simulation.

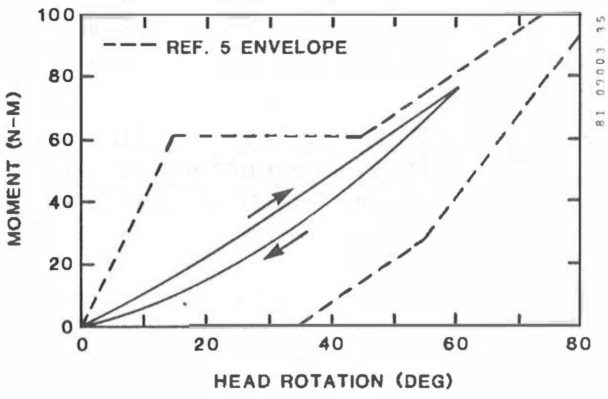

Fig. 9. Response of VIP-95 neck compared with proposed flexion response envelope of Ref. 5.

In the 40-G vertical test with the VIP-95 dummy, the load cell installed in the neck measured a maximum shear force in the x-direction of approximately 3,560 N. Ref. 5 reported measuring shear forces of 1,590 and 1,940 N without damage in dynamic tests of two cadavers. Also, in the same 40-G dummy test, the neck load cell measured maximum bending moments (y-axis) of approximately 102 N-m in extension and in excess of 226 N-m in flexion. Ref. 5 suggests tolerable levels of 757 N-m in extension and 190 N-m in flexion, measured with respect to the occipital condyles. For direct comparison with those values, the moments measured in the dummy near the base of the neck would have to be reduced to account for the difference in moment arm.

Conclusions

The results of this program indicate that forces and moments in the spine of an anthropomorphic dummy of Part 572 or similar design can be measured with a rather simple modification. However, if these measurements are to be used as predictors of vertebral injury, they must be correlated to fractures in the human spine. This is presently being accomplished through simulation of dummy and cadaver tests using the mathematical model described in Ref. 9. However, the vertebral strengths for the cadavers must be related to that of the flying population in order to develop a meaningful tolerance criterion for seat evaluation.

Tolerable levels of shear force and bending moment for the neck have been proposed in Ref. 5, but a detailed comparison of dummy and cadaver head-neck anthropometry is required to relate the load cell measurements to these levels.

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