DYNAMIC BIOFIDELITY OF THE PART 572 AND HYBRID III ANTHROPOMORPHIC TEST DUMMY HEADS

by

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ABSTRACT

Two anthropomorphic test dummy (ATD) heads are evaluated for biodynamic fidelity in a combined experimental and analytical study. In the experimental investigation, a Part 572 head, a Hybrid III head, and a series of cadaver heads were subjected to the same impacts. Blows were delivered by a padded, pneumatically powered impactor to three locations on the head. The area over the frontal and mandible bones were impacted in the midsagittal plane. The side of the head (parietal bone region) was impacted in the lateral direction. Head response measures are compared for impact velocities between 16 and 27.4 km/h. Limitations on the use of ATD head accelerations as human injury predictors are examined. Intracranial pressures are calculated from the head accelerations using a finite element brain model.

INTRODUCTION

In vehicle occupant protection research more reliance is being placed on anthropomorphic test dummies (ATDs). The difficulty is that at best ATDs can only approximate human kinematic and dynamic response; they cannot duplicate the human structure. Other restrictions on the ATDs are that they must produce repeatable results, and have secure instrumentation mounts. Also dummies must withstand injury producing loads without breaking. Thus, design compromises and tradeoffs affecting dynamic biofidelity cannot be avoided.

In this paper the dynamic response of two dummy heads is analyzed: the Part 572 (the ATD specified in the United States federal regulations) and the Hybrid III, the ADT being proposed for adoption by the General Motors Corporation. To keep the tests as simple as possible and to minimize the number of variables, frontal, mandible, and lateral impacts to the dummy heads were repeated on human cadaver heads, and the resultant head accelerations compared. The effect of the variations on injury assessment and brain response is discussed.

BACKGROUND

Comparisons of human and dummy head response have been reported in the literature (1-5). In these studies dynamic tests were conducted on ATD heads and repeated on humans and human cadaver subjects. The results of these tests vary; in some the dummy response is similar to the human, while in others it is not. In the non-impact event Muzzy (1) found that the dummy head motion varied from that of the living human, but not appreciably. In sled tests, the dummy head response compared favorably with the human subjects up to the time of peak angular head velocity. After the peak, differences in angular velocity, displacement, and linear acceleration between the dummy and human developed. Muzzy describes his dummy as having a Part 572 head with a Hybrid III design neck. Walsh (2) also found similar kinematics between a Part 572 dummy head and cadaver subjects in air bag tests. However, in the belt restrained tests the correlation degraded, and the dummy had a higher resultant head acceleration. Pritz (3) simulated pedestrian impacts with a Part 572 dummy and cadaver subjects. In his tests the cadaver sustained the highest head accelerations. This difference was attributed to the design of the dummy neck and shoulder. Hu (4) also found higher head accelerations in the cadaver subject when he simulated rear-end impacts. The dummy and cadaver head acceleration traces had different shapes. He believed that this difference was due to the design of the Part 572 neck. In reconstructions of actual accidents, Cesari (5) found many differences in head response between the cadavers and dummies. The Sierra dummy had higher head accelerations, and there was no correlation between the injuries in the accident and the HIC values calculated in the experiments.

In all of these previous studies correlation was best when the head did not strike anything, or struck an air bag. The greatest differences were obtained when the head hit structures such as the hood of the vehicle. In summary, these findings show that for a direct impact, the response of the dummy head is different from that of the human.

TEST PLAN

Six test subjects were used. Impacts were performed on the two ATD heads (the Part 572 and Hybrid III), and then repeated on one embalmed and three unembalmed human heads. Initially three types of padding were used (refer to Table 1). However, after 36 tests it was apparent that differences between padding A and B impacts were insignificant, and the use of padding A was discontinued.

Heads were impacted at three locations:

1. On the frontal bone or forehead, in the midsagittal plane (Fig. 1)
2. On the mandible bone or chin in the midsagittal plane (Fig. 2)
3. On the parietal bone or side of the head in the lateral direction (Fig. 3)

In every impact the force vector was directed towards the head center of gravity (C.G.), to minimize head rotation. Three different impactor velocities were employed for each padding type. These velocities ranged between 16 and 27.4 km/h. A total of 115 impact tests were conducted.

TEST METHODOLOGY

The loading impulse was provided by a pneumatically powered piston weighing 121 N. The piston mass was accelerated to the desired velocity in a stroke distance of 0.28 m. Its velocity was determined over the last 2.5 cm of travel prior to impact with a magnetic probe. The impactor surface was a 12.7 cm diameter flat aluminum disc. Padding materials were interposed to alter the impulse magnitude and time duration. The heads were inverted and suspended at three points by lengths of beaded chain. The opposite ends of the chain were attached to an overhead plate which was a distance of 1.78 m from the C.G. of the head. This pendulum arrangement allowed the head to

Figure 1. Frontal impact.

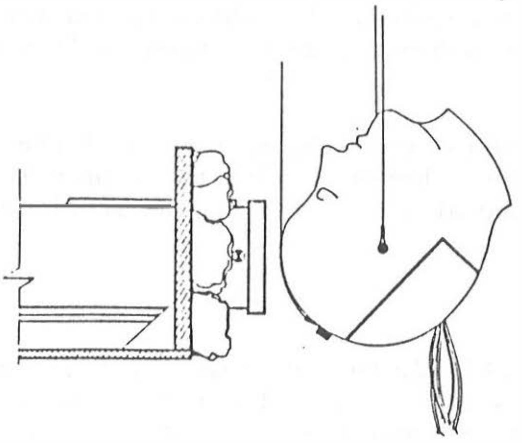

Figure 2. Mandible impact.

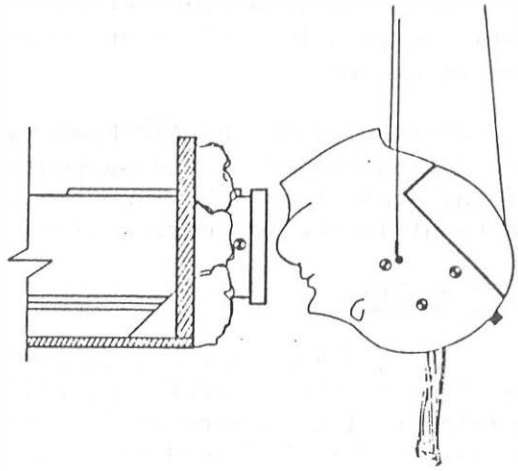

Figure 3. Lateral impact.

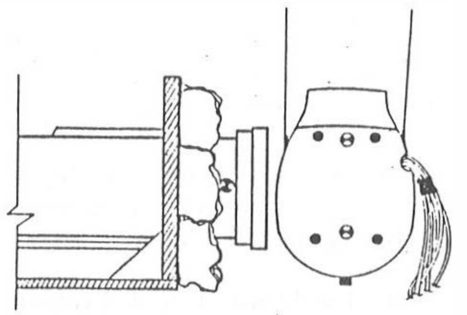

Table 1 Padding

Symbol	Material	Thickness cm (in)
A	ensolite	2.54 (1.0)
B	polystyrene	2.29 (0.9)
C	polystyrene	4.57 (1.8)

Table 2 Head Mass

Test Subject	Head Mass (slugs)	Scale Factor
Part 572	.328	1.0
Hybrid III	.296	.904
Embalmed No 1	.321	.979
Unembalmed No 1	.393	1.199
Unembalmed No 2	.321	.979
Unembalmed No 3	.291	.887

swing free during and after the impulse.

The cadaver skulls were removed from the donor subjects at the C1 level. An 11 mm diameter, 17.8 cm in length aluminum rod was inserted in the lateral direction through the soft tissue mass, at the base of the skull, at approximately the level of the foramen magnum. Each end of the rod was drilled and tapped to allow attachment of the chain suspension. The third suspension line terminated in a loop of suture sewn to the subject's nose. Head positions are shown in Figures 1-3.

Positioning of the ATD heads were similar with the exception of the lateral attachments of the suspension. Here, holes drilled and tapped in the parietal area of the skull casting, and one at the vertex of the skull, were used to affix the suspension lines.

IMPACT RESULTS

The head masses vary, as shown in Table 2, For the same impact, the acceleration of the heavier head will be less. To remove the mass effect, the acceleration traces were scaled or normalized, using a mass ratio factor. Mass of the Part 572 head was considered the standard, making the scaling factor the mass ratio of the impacted head to that of the Part 572 head. Thus, if the head is light compared to the Part 572 head, the acceleration is reduced; if it is heavy it is increased. For each test, the maximum value of the scaled resultant head acceleration is listed in Table 3.

Pulse duration is computed from the resultant acceleration trace. It is the time period during which the resultant acceleration exceeds 10% of its maximum, or peak value. Pulse durations for each test are listed in Table 3.

To illustrate the differences between the impacts, peak head accelerations versus impactor velocity, and peak head acceleration versus pulse duration were plotted. Refer to Figures 4-9.

<u>ATD Impact Results</u>. In the frontal and lateral impacts, the differences between the Hybrid III and Part 572 heads are substantial. For the same energy frontal impacts, with 2.29 to 2.54 cm of padding, the Part 572 head accelerations are more than twice that of the Hybrid III (Figure 4). The same is true of the lateral impacts at impactor velocities above 21 km/h (Figure 6). For paddings A and B, the differences increase with impactor velocity. When the padding is thicker (padding C), the differences are less. In all the frontal and lateral impacts, the Hybrid III tends to have a longer pulse duration (Figures 5 and 7). In the mandible impacts, the Part 572 and Hybrid III head accelerations are similar in magnitude and duration.

<u>Human Subject Impact</u>. In the frontal and lateral impacts the response parameters are similar. The points fall within definable regions, exhibiting the same trend. Refer to Figure 4. As would be expected, head accelerations are somewhat higher in the embalmed subject due to the increase in tissue stiffness. A consistent trend of decreasing pulse duration with increasing head accelerations is exhibited in Figure 5. In the mandible impacts, the accelerations are lower and relatively constant (Figure 8). Even in the high energy impacts, the accelerations are below 200 G's. Apparently, energy is dissipated by motion of the mandible relative to the skull. The pulse durations are grouped according to padding types, but the trends are not obvious.

TABLE 3 Test Results
Resultant Scaled Pk.G's and
Computed Maximum Intracranial Pressures

Test Subject	Case No.	Pad	Impact Site	Impactor Vel km/h (mph)	Scaled Pk.G's	Duration (ms)	Presure N/cm (psi)
Hybrid III	96	A	F	13.7 (8.5)	61.	14.0	12.2 (17.7)
	97	A	F	21.1 (13.1)	186.	8.0	16.9 (24.6)
	98	A	F	17.7 (11.0)	112.	10.1	10.3 (15.0)
	99	A	F	27.5 (17.1)	344.	4.6	40.8 (59.2)
	100	B	F	18.3 (11.4)	101.	8.8	9.4 (13.6)
	101	B	F	21.9 (13.6)	191.	7.0	17.8 (25.8)
	102	B	F	27.4 (17.0)	329.	3.6	39.1 (56.7)
	103	C	F	27.4 (17.0)	134.	13.5	11.2 (16.2)
	104	C	F	21.9 (13.6)	84.	15.2	6.8 (9.8)
	105	C	F	17.7 (11.0)	55.	19.5	4.3 (6.2)
Part 572	106	C	F	17.5 (10.9)	68.	16.7	5.4 (7.9)
	107	B	F	17.2 (10.7)	313.	3.8	34.8 (50.5)
	108	A	F	17.4 (10.8)	246.	6.8	21.8 (31.6)
	109	A	F	21.1 (13.1)	450.	2.3	49.8 (72.3)
	110	B	F	22.5 (14.0)	497.	2.2	56.8 (82.4)
	111	C	F	22.5 (14.0)	141.	11.2	12.5 (18.1)
	112	C	F	27.5 (17.1)	262.	8.3	23.4 (34.0)
	113	B	F	27.4 (17.0)	749.	1.4	84.5 (122.6)
	114	A	F	27.4 (17.0)	715.	1.8	80.2 (116.4)
	115	C	M	16.7 (10.4)	67.	16.3	3.2 (4.6)
	116	B	M	18.0 (11.2)	161.	7.0	9.1 (13.2)
	117	B	M	23.7 (14.7)	256.	5.0	17.9 (26.0)
	118	C	M	23.2 (14.4)	123.	11.8	6.5 (9.5)
	119	C	M	25.4 (15.8)	196.	9.0	11.4 (16.6)
	120	B	M	26.5 (16.5)	378.	3.7	28.0 (40.6)
	121	C	L	26.9 (16.7)	211.	8.9	10.4 (15.1)
	122	B	L	25.4 (15.8)	741.	2.3	44.7 (64.9)
	123	B	L	20.4 (12.7)	386.	3.6	22.4 (32.5)
	124	C	L	21.7 (13.5)	107.	12.4	5.4 (7.8)
	125	C	L	17.2 (10.7)	64.	17.0	2.8 (4.1)
	126	B	L	19.3 (12.0)	196.	6.0	10.7 (15.5)
Hybrid III	127	B	L	17.2 (10.7)	109.	9.3	6.2 (9.0)
	128	C	L	17.9 (11.1)	60.	19.0	3.0 (4.3)
	129	C	L	20.4 (12.7)	79.	15.0	
	130	B	L	20.4 (12.7)	194.	6.6	12.1 (17.6)
	131	B	L	24.8 (15.4)	308.	4.3	19.5 (28.3)
	132	C	L	27.7 (17.2)	165.	11.8	8.8 (12.7)
	133	C	M	27.7 (17.2)	289.	4.8	22.5 (32.6)
	134	B	M	26.1 (16.2)	380.	2.9	31.8 (46.1)
	135	B	M	21.2 (13.2)	243.	4.8	16.5 (23.9)
	136	C	M	22.8 (14.2)	146.	8.4	9.1 (13.2)
	137	C	M	17.1 (10.6)	68.	17.4	4.0 (5.8)
	138	B	M	19.3 (12.0)	151.	4.8	10.3 (14.9)

TABLE 3 Test Results
Resultant Scaled Pk.G's and
Computed Maximum Intracranial Pressures (cont.)

Test Subject	Case No.	Pad	Impact Site	Impactor Vel km/h (mph)	Scaled Pk.G's	Duration (ms)	Presure N/cm (psi)
Emb No 1	140	B	F	16.9 (10.5)	257.	6.2	
	142	B	F	20.1 (12.5)	447.	5.7	
	143	C	F	19.6 (12.2)	133.	11.7	
	144	C	F	25.1 (15.6)	258.	10.0	
	145	B	F	25.1 (15.6)	389.	6.2	
	146	C	F	16.1 (10.0)	70.	16.1	
	147	C	M	16.1 (10.0)	76.	17.0	
	148	B	M	16.7 (10.4)	85.	13.0	
	149	B	M	20.4 (12.7)	96.	11.5	
	150	C	M	20.0 (12.4)	86.	18.6	
	151	C	M	25.1 (15.6)	100.	14.3	
	152	B	M	25.1 (15.6)	126.	13.2	
	153	C	L	25.1 (15.6)	188.	12.9	
	154	B	L	26.1 (16.2)	321.	8.3	
	155	B	L	20.1 (12.5)	296.	9.0	
	156	C	L	21.1 (13.1)	102.	16.3	
	157	C	L	17.4 (10.8)	67.	18.0	
	158	B	L	17.2 (10.7)	99.	10.7	
Unemb No 1	163	B	F	17.2 (10.7)	169.	13.0	10.1 (14.7)
	164	C	F	17.2 (10.7)	70.	19.2	3.9 (5.7)
	165	C	F	19.6 (12.2)	105.	16.9	6.2 (9.0)
	166	B	F	19.6 (12.2)	260.	4.7	24.8 (36.0)
	167	B	F	24.9 (15.5)	390.	4.5	32.9 (47.7)
	168	C	F	24.9 (15.5)	316.	12.0	9.5 (13.8)
	169	C	L	24.9 (15.5)	191.	12.9	8.0 (11.6)
	170	B	L	24.9 (15.5)	280.	8.0	14.1 (20.5)
	171	B	L	19.6 (12.2)	200.	8.4	8.4 (12.2)
	172	C	L	21.1 (13.1)	124.	14.9	4.6 (6.7)
	173	C	L	17.2 (10.7)	83.	18.1	3.9 (5.7)
	174	B	L	17.4 (10.8)	143.	10.8	6.2 (9.0)
	175	B	M	17.2 (10.7)	155.	6.2	2.5 (3.7)
	176	C	M	16.9 (10.5)	47.	20.7	1.5 (2.2)
	177	C	M	22.0 (13.7)	141.	15.5	2.4 (3.5)
	178	B	M	20.9 (13.0)	102.	14.1	3.4 (5.0)
	179	B	M	25.1 (15.6)	106.	15.3	4.2 (6.1)
	180	C	M	23.7 (14.7)	90.	18.9	2.4 (3.5)
Unemb No 2	181*	B	F	16.7 (10.4)	232.	4.1	17.6 (25.6)
	182	C	F	17.1 (10.6)	59.	19.1	2.5 (3.7)
	183	C	F	22.8 (14.2)	112.	15.3	5.5 (8.0)
	184*	B	F	22.8 (14.2)	420.	2.4	28.5 (41.3)
	185*	B	F	24.9 (15.5)	641.	2.4	44.6 (64.8)

*unreliable

TABLE 3 Test Results
Resultant Scaled Pk.G's and
Computed Maximum Intracranial Pressures (cont.)

Test Subject	Case No.	Pad	Impact Site	Impactor Vel km/h (mph)	Scaled Pk.G's	Duration (ms)	Presure N/cm (psi)
Unemb No 2	186	C	F	24.9 (15.5)	152.	12.7	8.5 (12.3)
	187	C	L	16.7 (10.4)	170.	9.1	9.9 (14.4)
	188*	B	L	17.1 (10.6)	357.	5.0	19.2 (27.9)
	189*	B	L	22.8 (14.2)	283.	5.0	28.3 (41.1)
	190	C	L	22.8 (14.2)	200.	12.7	10.1 (14.6)
	191*	C	L	24.9 (15.5)	161.	4.7	10.0 (14.5)
	192*	B	L	24.9 (15.5)	206.	6.1	13.1 (19.0)
	193	C	M	17.2 (10.7)	47.	16.5	4.3 (6.2)
	194	B	M	17.2 (10.7)	61.	12.4	4.8 (7.0)
	195	B	M	22.0 (13.7)	73.	16.9	5.9 (8.6)
	196	C	M	22.8 (14.2)	55.	13.9	4.9 (7.1)
	197	C	M	24.9 (15.5)	82.	11.9	6.9 (10.0)
	198	B	M	24.9 (15.5)	93.	10.9	7.4 (10.8)
Unemb No 3	199	B	F	17.2 (10.7)	247.	3.1	8.7 (12.6)
	200	C	F	17.2 (10.7)	81.	13.8	3.8 (5.5)
	201	C	F	20.5 (12.7)	115.	14.5	9.4 (13.6)
	202	B	F	22.8 (14.2)	340.	3.6	41.9 (60.8)
	203	B	F	24.9 (15.5)	363.	6.7	30.5 (44.2)
	204	C	F	24.9 (15.5)	142.	13.5	10.3 (15.0)
	205	C	L	27.4 (17.0)	181.	12.0	12.2 (17.7)
	206	B	L	24.9 (15.5)	347.	5.4	26.8 (38.9)
	207	B	L	21.1 (13.1)	202.	8.8	13.9 (20.2)
	208	C	L	20.4 (12.7)	94.	15.1	6.4 (9.3)
	209	C	L	17.2 (10.7)	66.	17.5	4.5 (6.5)
	210	B	L	17.9 (11.1)	147.	10.7	9.8 (14.2)
	211	B	M	17.2 (10.7)	151.	8.0	5.6 (8.1)
	212	C	M	17.4 (10.8)	52.	15.8	2.6 (3.8)
	213	C	M	20.1 (12.5)	101.	14.5	3.7 (5.4)
	214	B	M	20.4 (12.7)	194.	7.8	12.8 (18.6)
	215	C	M	25.1 (15.6)	112.	13.8	9.0 (13.0)
	216	B	M	24.9 (15.5)	105.	14.6	10.2 (14.8)

*unreliable

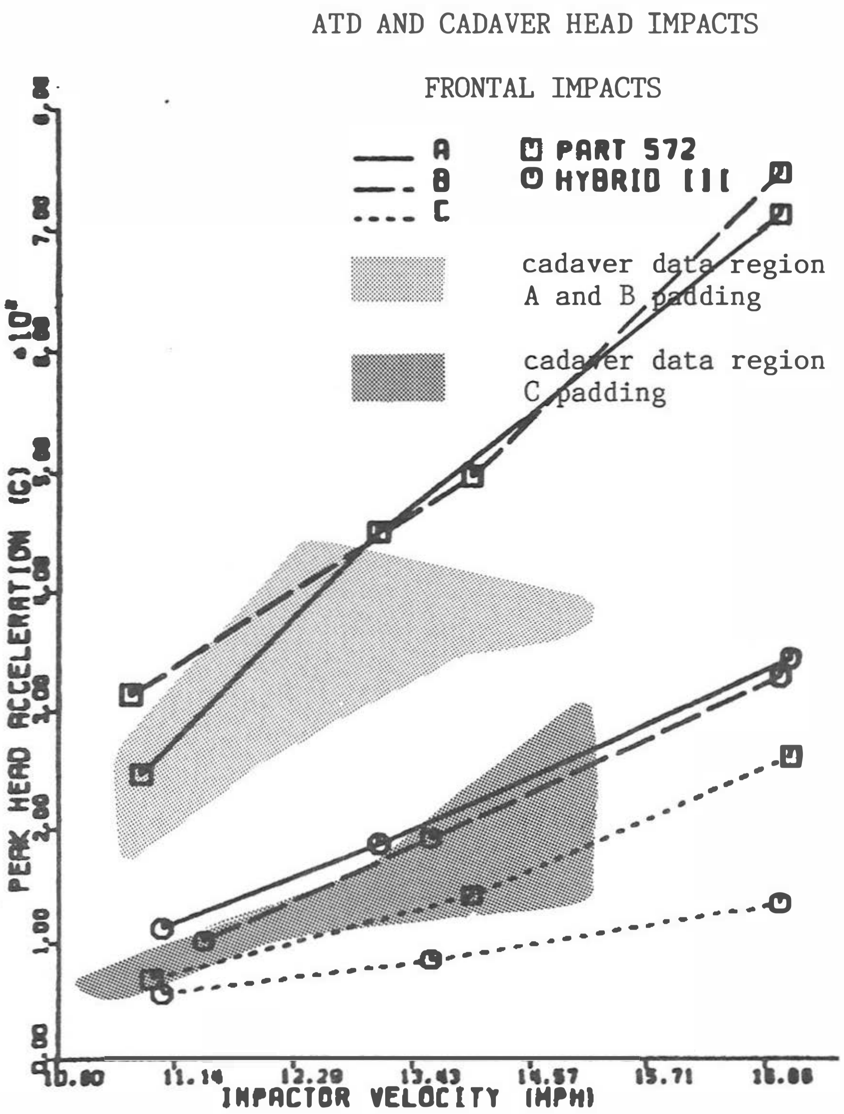

Figure 4. Peak head acceleration versus impactor velocity.

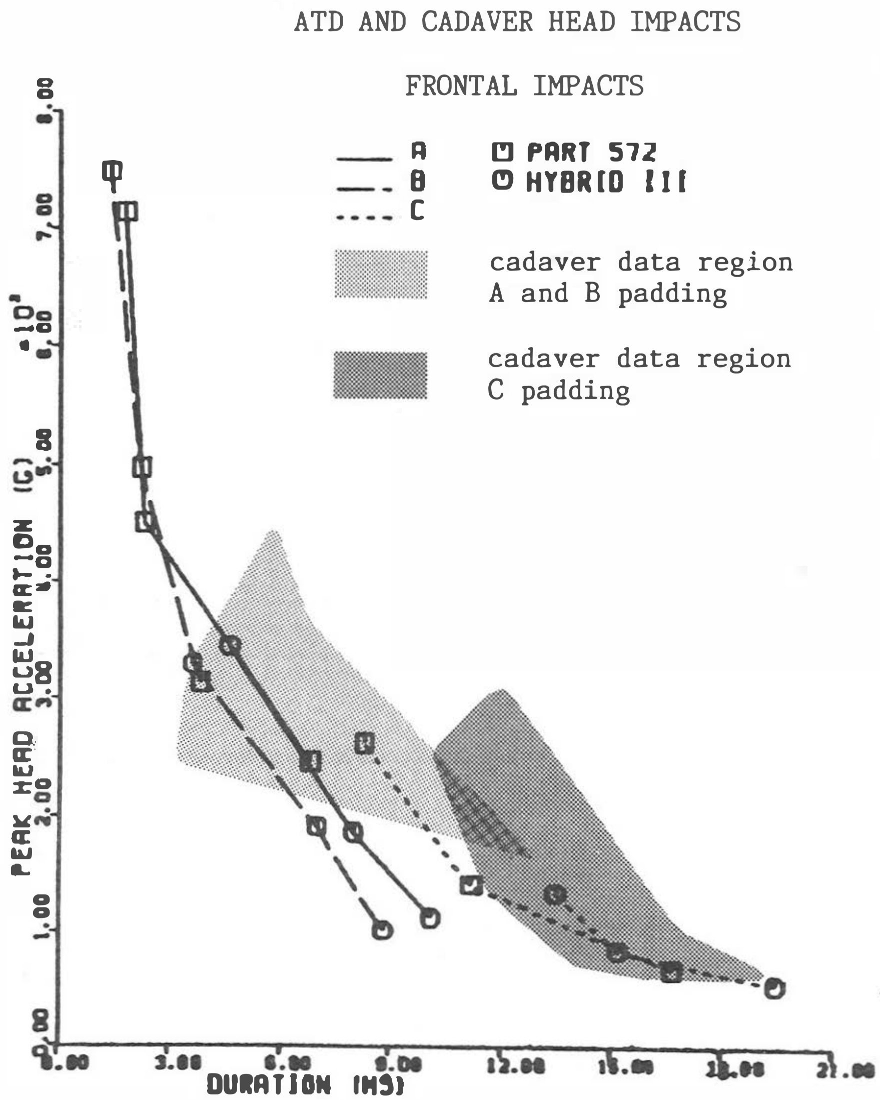

Figure 5. Peak head acceleration versus duration.

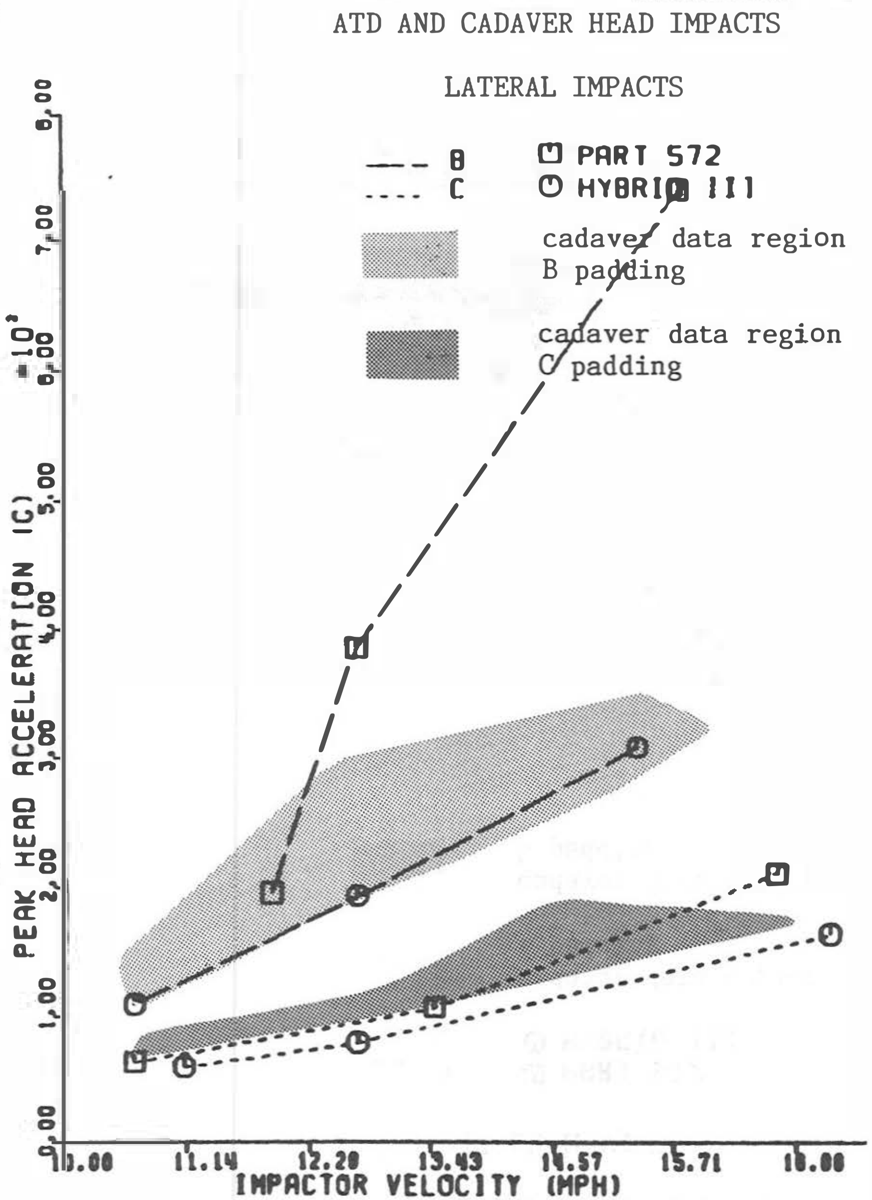

Figure 6. Peak head acceleration versus impactor velocity.

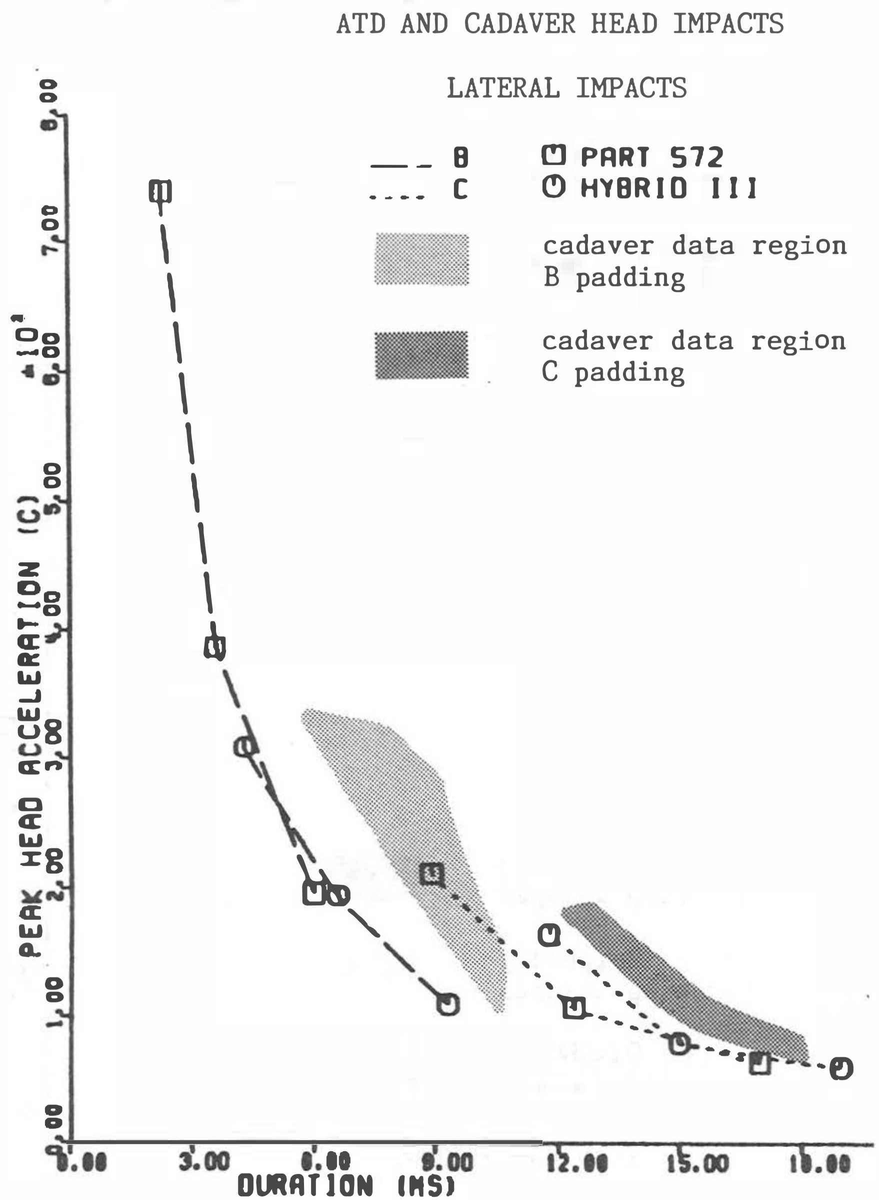

Figure 7. Peak head acceleration versus duration.

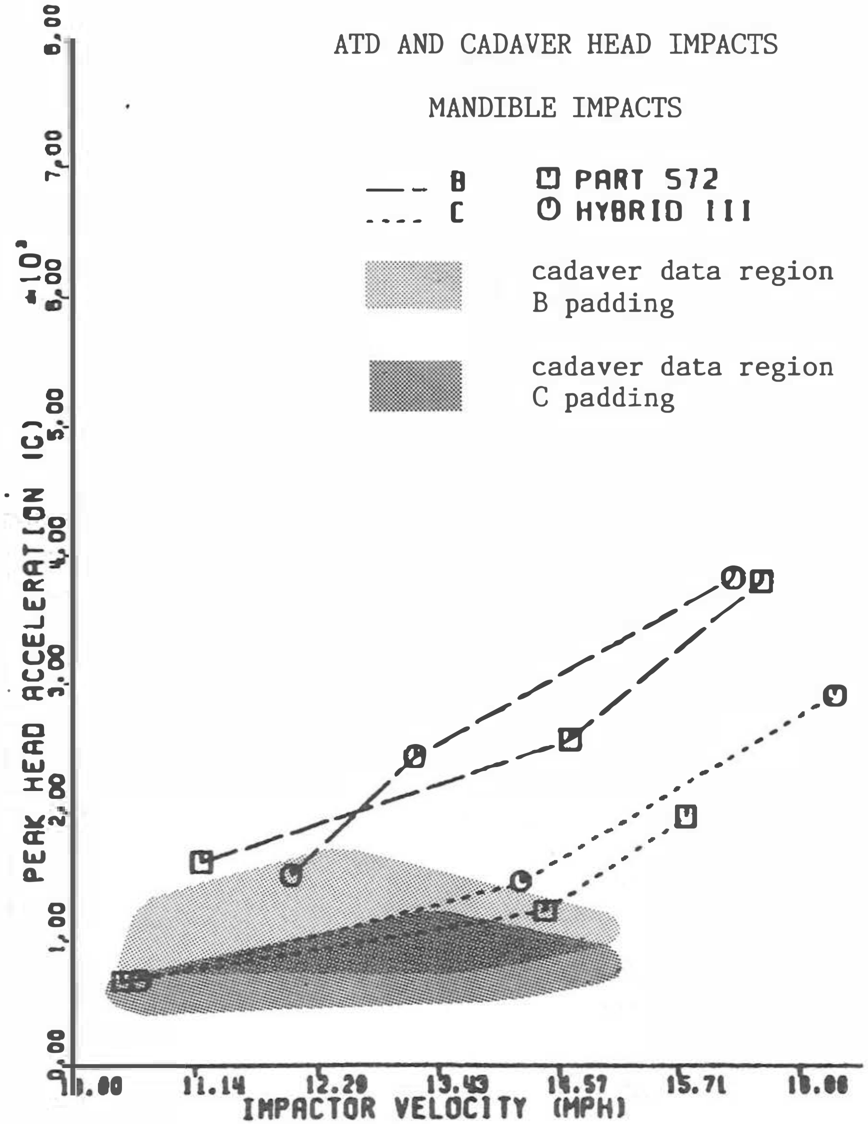

Figure 8. Peak head acceleration versus impactor velocity.

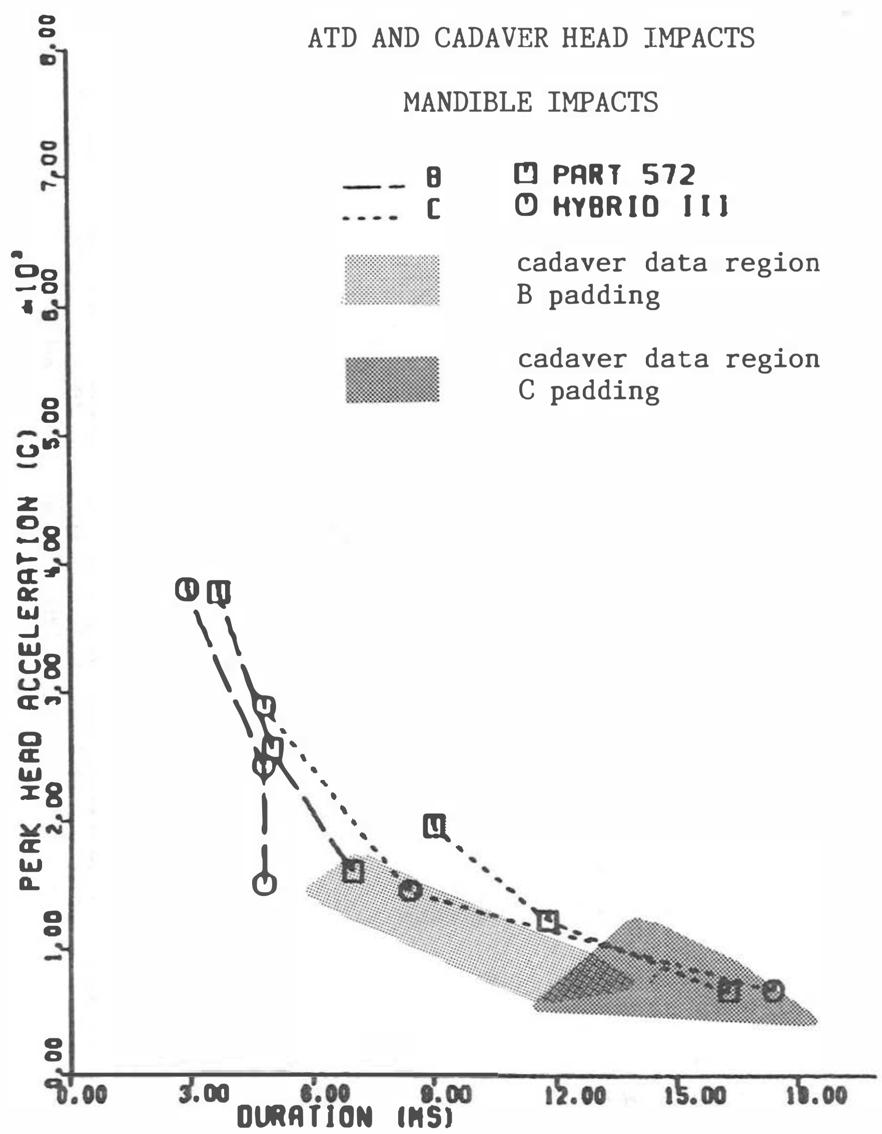

Figure 9. Peak head acceleration versus duration.

Comparison of ATD and Human Subject Impact Results. When thin padding is used in the frontal and lateral impacts, the Part 572 accelerations are higher than the unembalmed human accelerations (Figure 4 and 6). In these impacts, the Part 572 pulse duration is shorter than those of the human subjects (Figures 5 and 7). The reverse is true for the Hybrid III. In frontal impacts, the Hybrid III head accelerations are lower than the human data (Figure 4). In lateral impacts, the Hybrid III better approximates the human subject acceleration magnitudes (Figure 6), but tends to have too short a pulse duration (Figure 7). In mandible impacts, both ATDs have higher accelerations and shorter pulse durations than the human test subjects (Figures 8 and 9).

MATHEMATICAL SIMULATION USING THE FINITE ELEMENT BRAIN MODEL

Using the measured head accelerations as if they were actual human head accelerations, the stresses which would result at six brain locations were calculated. A finite element model of the brain is used in combination with a convolution solution procedure.

Finite Element Model. In this mathematical idealization, the brain tissue and fluids are represented with six-sided block elements, the assembled elements approximating the irregular shape of the brain. Refer to Figure 10. Internal to the model, four-node membrane elements simulate the partitioning internal folds of dura (the falx and the tentorium). In all elements, the mass is considered concentrated at the corners or nodes. The external shape of the brain is maintained to simulate the inner skull surface, forming a container for the brain. An opening representing the foramen magnum is modeled, which allows movement of the cervical cord into and out of the cranial cavity.

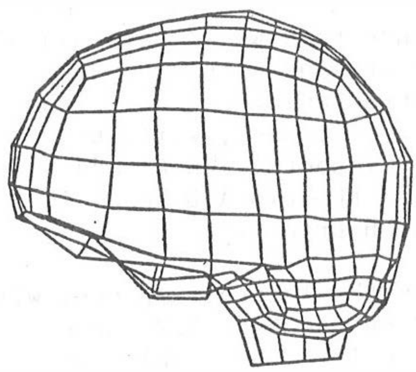

Figure 10. Finite element brain model.

Material Properties. Because the brain material is strain rate dependent and the appropriate material constants have not been defined, values from an earlier parametric study were used. In that study, measured and computed intracranial pressures were compared for a range of material constants (6). Properties which provided good correlation were selected. A Young's Modulus of 650 k Pa for the composite brain, vasculature and contained fluids was used. The effective compressibility of the composite material has been shown to be strain, or loading rate, dependent. At higher rates of onset, the material becomes less compressible. This is thought to be a function of flow into and out of the cranial cavity; at a slow rate of onset, the pressure-relieving flow has a greater influence on response. In the brain material elements, the compressibility, as defined by Poisson's ratio, is varied between the values 0.49 and 0.499. In these simulations the value selection is based on the average rate of change of acceleration (jerk) between 10 percent of peak and the peak value. For jerk values above 75,000 g's/sec, a Poisson's ratio of 0.499 is used. For jerk values below 75,000 g's/sec, a Poisson's ratio of 0.49 is employed.

Solution Procedure. In the finite element calculations the equations are generated in terms of a skull fixed axis. Head motion is imposed by mathematically translating the axis frame. Using this procedure, any computational inaccuracies caused by large displacements of the head are eliminated.

SIMULATION RESULTS

The brain has a characteristic response: it tends to lag the motion of the skull due to its inertia. Brain tissue compresses against the skull near the impact site and is in tension opposite the impact. The result is a pressure, or stress gradient, through the brain. Motion of the cervical cord through the foramen magnum prevents high magnitude stresses from developing in the posterior fossa. Shear strains develop along the brain skull interface and the boundaries of the falx and tentorium, as the brain rotates inside the skull.

In the frontal impacts, stresses or pressures are highest in the frontal lobe. Traces of these stresses resemble the shape of the resultant head acceleration. In the lateral impacts, high tension stresses develop on the side of the head opposite the impact. The same is true in the mandible impacts. The maximum pressures (hydrostatic stresses) which would develop in a head are tabulated for each impact in Table 3. Peak intracranial pressure is a measure of the brain response magnitude and has been correlated with the occurrence and severity of brain injury (7). In the frontal impacts using the thinner padding, accelerations from the Part 572 predict much higher intracranial pressure than would be produced in the human head. The Hybrid III predicts lower pressures than those in a human subject. Using an injury criteria based on intracranial pressure, the Part 572 dummy head would predict brain injury for every impact, while the Hybrid III would predict injury only at the higher energy impacts. When padding C was used, the resulting pressure variation between the two ATD heads is less, but at 25.7 km/h the Part 572 is twice that of the Hybrid III.

In the lateral impacts, with 2.54 cm of padding, a pressure prediction based on the Part 572 impact would be high, while the Hybrid III results would be more like those which occur in a human subject. The Part 572 is adequate when the thick padding is used, but the Hybrid III pressure values tend to be low.

In the mandible impacts, the ATD predicted pressures are higher than would occur in the human subject. This is true for all types of padding.

DISCUSSION

The impact tests have pointed out major response differences between the two ATD heads. These variations were most pronounced when the thinner padding was used. The greatest variations occur in the frontal impacts where the Part 572 accelerations are twice that of the Hybrid III. When these higher accelerations are input to the finite element model, a higher brain response is predicted. The acceleration and simulated pressure pulse for the Part 572 are also shorter than that for the Hybrid III. In the mandible impacts the reverse is true; the Hybrid III has the highest acceleration indicating that the chin of the Hybrid III is more rigid.

Padded frontal impacts were recently conducted by H. Mertz at General Motors. He records higher accelerations for the Hybrid III head with less variation between the two ATD head types. Mertz allocates part of the difference between his results and those reported in this paper to the pad characteristics. In an evaluation of pad impact history, he shows that the first impacts on a pad have lower accelerations with a greater standard deviation than later impacts. Because our Hybrid III tests were conducted first in our series, the head accelerations would tend to be lower in comparison with later impacts. The magnitude of this effect can not be determined at this time. But if the Hybrid III accelerations were raised to compensate for the initially greated pad attenuation, the comparison with the cadaver data would be improved. G.M. avoided this effect by using new pads in each of their tests and conducting a larger number of tests. Other factors which would influence the results are differences in test protocal and equipment. G.M. used a pendulum instead of a pneumatic impactor, and two support attachments on the head instead of three. The head weights are different, indicating that the neck transducer mount was attached the the G.M. head. This would influence the mass distribution. Rotation was not recorded in these tests and, although small, could be different in the two test programs. Although both Hybrid III heads passed the drop test calibration, it is conceivable that there were small variations due to manufacturing and simulated skin condition.

Data obtained from the cadaver subjects is consistent; that is, the parameters fall within definable regions. In these impacts, the embalmed subject usually has the highest accelerations which is due to its increased tissue stiffness and dehydration. The difference between the embalmed and unembalmed heads is less in the higher energy impacts where the properties of the bone become important.

Both ATDs predict higher head accelerations for the mandible impact than would occur in the living human. Although overall the Hybrid III head better approximates the human head, the use of the Hybrid III could have serious consequences. In the frontal impacts, this ATD head predicts head accelerations lower than that which would occur in the human. Thus if a known head acceleration produces injury in the human head, injury would not be accurately predicted by the Hybrid III.

The differences in the head acceleration time history profiles between the ATD and human subjects would result in different values for severity indices based on these profiles. This could compromise the use of these indices. However these tests were for specific laboratory conditions. The authenticity of the ATD's simulation of human head response should be reexamined for impact conditions likely to be encountered in vehicle passenger compartments.

CONCLUSIONS

1. In similar impacts the Part 572 head had higher accelerations than the Hybrid III; the only exception being impacts to the chin.

2. Differences between the two ATD heads were greatest with the 2.54 cm

padding, the thinnest padding.

3. In comparison with human subject impacts, the Part 572 head produces higher accelerations. In the frontal and lateral impacts with thin padding, the Part 572 accelerations far exceed those measured on the human unembalmed subjects.

4. In comparison with human subject impacts the Hybrid III head accelerations are low in the frontal impacts, approximately the same as the human head in the lateral blows, and higher than the human head in the mandible impacts.

5. Intracranial pressures, i.e. brain response, predicted from ATD head accelerations would not be the same as the pressures in a human head in most cases.

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