INTRACRANIAL PRESSURE AND BRAIN INJURY IN FRONTAL IMPACTS

by

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INTRODUCTION

In the past, the dynamic response of the human brain to head impact could only be described in general, nonspecific terms. It was known that a pressure gradient develops in the brain; but, with few exceptions,* pressure magnitudes, their time durations, and effects on injury were unknown. The mechanics of closed skull brain injury had been theorized but never proven, and the validity of head injury indices computed from head accelerations have been questioned. Quantitative cause and effect relationships do not exist although head trauma is a serious and common injury. The brain and its surrounding tissue and contained fluids form a complex dynamic system. The dynamic response characteristics of the integral system are just beginning to be understood. In this report a technique for predicting intracranial pressure is presented and a relationship between injury severity and frontal pressure is shown.

DISCUSSION

Intracranial pressures can be predicted for frontal head impacts using recently improved models of the human brain. These linear finite element models employ elastic brick elements to represent the soft tissue and membrane elements to represent the partitioning internal folds of dura, the falx and tentorium. The exact external shape of the intracranial contents is maintained, but the skull itself is not required.** Head impacts are simulated by imposing the skull translational and rotational accelerations on the model. The solution technique is described in Ref. 3.

*Some pressure gradients were measured in cadavers (Ref. 1) and fluid filled skulls (Ref. 2).

**Comparison of intracranial pressures from a brain model with and without an attached skull showed the effect of the skull deformation on intracranial pressures is minimal.

In the past two years the models have been changed and upgraded to improve correlation between measured and model predicted responses. Although live animal brain responses have been used, the best data for substantiating the models was obtained from unembalmed pressurized cadaver tests, Ref. 4 and 5. In these experiments the cadaver was seated in front of a sliding impact device. Suture attachments to an overhead frame positioned the head but did not interfere with movement. To minimize head rotation the anatomical axis of the skull, the Frankfort Plane, was inclined 45 degrees relative to the horizontal. Normal _in vivo_ pressures were obtained in the vascular and cerebrospinal fluid systems prior to impact. The vascular fluid contained india ink to mark contused areas by turning them black just as bleeding would turn them red in a living brain. In one series of tests, pressures were measured subdurally at five or six locations on the brain surface. These tests were simulated using the brain models, and the measured and computed pressures were compared. The responses of the earlier (1976) models were slow and lagged the measured pressures as shown in Figure 1. Although this delay was corrected by changing the material properties, computational inaccuracies due to the brain's near incompressibility had to be eliminated. The regular element in the 1977 model (Figure 1) was replaced with a new split energy element. This element, described in Ref. 6, is accurate for all values of compressibility. Using a wide variety of check problems the new element was studied and validated.

The in situ material properties for the brain and contained fluids have not been adequately defined. Experimental research on specimens of brain indicate that the response is viscoelastic, loading-rate sensitive, and nonlinear, Ref. 7. However, because the event is of short duration, the nonlinear properties are approximated with effective linear material constants. These constants were determined in a parametric study. Thirteen impact tests were simulated using a range of material constants, and the computed and measured pressures compared. This study revealed that for the composite material (brain, vascular system and fluid), a Young's Modulus (E) of 6,6000,000 dynes/cm^2 gave good results. The stress-strain modulus for brain material is very strain rate dependent (Ref. 7), and for the high strain rates at impact a value in this range could be expected.

The parametric study revealed that a single value for Poisson's ratio (ν) was not adequate; the hard surface impacts require higher values. The padded impact has a long acceleration pulse (Figure 2a) while the unpadded and ineffectually padded impacts have a sharp spike shaped acceleration pulse (Figures 2b and c). A higher Poisson's ratio, to represent a more nearly incompressible material, is required to simulate these spike shaped acceleration traces. The underlying reason for this requirement is still being researched. It may be that the material constants should be varied throughout the model to more accurately represent the brain or it may be a natural consequence of the system mechanics. The pressure release mechanisms (flow out of the cranial cavity) are a function of the event time duration. During the spike shaped acceleration these mechanisms would not have sufficient time to act or effect the pressure response. Then as the pulse duration decreases the nearly incompressible character of the brain tissue would begin to predominate. Three values for ν (0.48, 0.49, and 0.499) were selected, defining Models I, II and III. Model selection is based on the shape of the acceleration trace. The value of the acceleration peak magnitude, A, divided by the average pulse width, T, is the model selection parameter. Refer to Figure 2 and Table 1. Employing the three models as specified in Table 1, the correlation between the measured and computed intracranial pressures is good. Sample correlation for the three models are shown in Figures 3, 4 and 5. When the selection procedure in Table 1 is not followed, a variation between measured and computed peak pressures results. This is demonstrated in Figure 4: a low Poisson's ratio model, Model I, was used to compute peak pressure in an unpadded impact.

Although the model can predict pressures, the research objective is the prediction of injury. Using the three models, ten additional pressurized cadaver head impact tests were simulated. In these experiments the brain injuries were graded, but pressures were not measured, Ref 4. Injury codes of 1, 2 and 3 were established to indicate minor, moderate and severe injury, respectively. Because the injuries were primarily in the frontal region of the brain the computed frontal pressures, shown in Table 2, were compared to injury severity. A bar plot of the pressures and corresponding injury severity numbers, Figure 6, shows the following relationship between pressures and injury. Pressures above 2.30×10^6 dynes/cm^2 correspond to a severe injury. Pressures between 1.80×10^6 and 2.30×10^6 dynes/cm^2 correspond to a moderate injury, and pressures less than 1.80×10^6 dynes/cm^2 correspond to minor or no injury. The same relationship exists between the magnitude of occipital computed pressures and occipital brain injury in four live animal tests, Ref. 8. The Head Injury Criterion (HIC) and Gadd Severity Index (GSI) are currently used to predict injury: a HIC or GSI of 1000 indicates a serious injury. The ability of these two indices to predict injury can be compared to the prediction based on frontal pressure. Bar graphs similar to the one used for frontal pressure were prepared for the HIC and GSI, Figures 7 and 8. These indices do not predict the serious injury which occurred in Test 29. However, the frontal pressure correctly predicts a grade 3 injury. In this particular test series the frontal pressure magnitude is a better predictor of injury than the HIC or GSI. The frequency and shape of the forcing function (head acceleration) and the system response characteristics of the brain predominate in the calculation of the intracranial pressures. The HIC and GSI are computed by using only the integration of the head acceleration raised to the 2.5 power. In this integration the contribution of the spike shaped acceleration pulse is small. But the effect of the acceleration spike on intracranial pressure is significant as shown in Test 29.

These results demonstrate the importance of adequate padding. Appropriate padding would have eliminated the spike shaped acceleration pulse and ensuing high intracranial pressures in Test 29. No injury would have occurred in Test 29 if the impactor had been adequately padded. Extrapolating to head impacts in vehicle accidents, lives could be saved by covering potential hard surface head impact sites with compressible material. Also, padding in helmets may be more important than originally believed. High magnitude short duration intracranial pressures occur when the padding is too hard, or when it compresses completely (bottoms out) under the load. Helmet padding should be designed to minimize the pressure pulses developing in the brain, by providing optimum crush rates for a range of impacts.

This combined analytical and experimental research effort has revealed important facts about the response of the brain to impact. A better understanding of frontal impact injury has resulted. But there is much more to learn. A great amount of additional experimental research is needed. The results used in this study reported in Ref. 4 and 5 constitute only a small injury sample. Additional impacts such as side and occipital need to be performed and simulated to test the models. The effects of high head rotational accelerations and velocities need to be studied. Response measures other than pressure may be important when rotation is significant. Experimentally, in situ brain material properties need to be defined for these short duration high loading rate events. (The properties probably vary throughout the brain, but with the limited information available, uniform material properties had to be assumed.) Also, the compressibility provided by flow out of the cranial cavity needs to be investigated.

CONCLUSIONS

1. Models now exist which can predict intracranial pressures for frontal head impact.

2. A relationship between frontal pressure magnitude and frontal lobe injury severity is demonstrated.

3. In this particular test series the frontal pressure magnitude is a better indicator of injury than the HIC or GSI indices.

4. Adequate padding of possible impact surfaces and in helmets could be very effective in preventing the type of injury observed in these tests. Such padding would eliminate the high magnitude impulsive type head accelerations which produce high magnitude pressure pulses in the brain.

REFERENCES

1. S. O. Lindgren, "Experimental Studies of Mechanical Effects in Head Injury". Acta Chirurgica Scandinavia, Supplementum 360, Stockholm, 1966.

2. V. L. Roberts, V. Hodgson, L. M. Thomas, "Fluid Pressure Gradients Caused by Impact to the Human Skull." ASME paper (66-HUF-1), Biomechanics Monograph, 1967.

3. C. C. Ward and R. B. Thompson, "The Development of a Detailed Finite Element Brain Model." Proc. 19th Stapp Car Crash Conf., 1975.

4. A. M. Nahum and R. W. Smith, "An Experimental Model for Closed Head Impact Injury." Proc. 20th Stapp Car Crash Conf. 1976. (SAE 760825)

5. A. M. Nahum, R. Smith and C. C. Ward, "Intracranial Pressure Dynamics During Head Impact." Proc. 21st Stapp Car Crash Conf. 1977. (SAE 770922)

6. D. S. Malkus and T. J. R. Hughes, "Mixed Finite Element Methods Reduced and Selective Integration Techniques: A Unification of Concepts." submitted to Computer Methods in Applied Mechanics and Engineering.

7. "Determination of the Physical Properties of Tissues of the Human Head - Final Report," Biomechanics Laboratories of Department of Theoretical and Applied Mechanics College of Engineering, West Virginia University. Contract # Ph-43-6 -1137, May, 1970.

8. C. C. Ward, P. E. Nikravesch and R. B. Thompson, "Biodynamic Finite Element Models Used in Brain Injury Research." Presented at Symposium on Biodynamic Models and their Applications, Dayton, Ohio, February, 1977. To be published in the Journal of Aviation, Space and Environmental Medicine.

Table 1. Model Properties

Model No.	A/T* $\times 10^8$ cm/sec^3	Type of Element	Young's Modulus dynes/cm^2	Poisson's Ratio
I	0-1.5	Split Energy	6670000.0	0.48
II	1.5-2.5	Split Energy	6670000.0	0.49
III	>2.5	Split Energy	6670000.0	0.499

Table 2. Simulated Frontal Pressures

UCSD Test No.	$\frac{A^*}{T}$ $\times 10^8$ cm/sec^3	Model Used	Pressure $\times 10^6$ dynes/cm^2
15	0.8	I	1.53
17	2.7	III	4.17
18	2.2	II	1.41
19	1.1	I	1.76
26	0.45	I	1.09
27	0.15	I	0.40
28	1.37	I	1.98
29	4.16	III	3.10
31	0.63	I	1.32
32	1.65	II	2.43

*A/T = peak head acceleration/average pulse width

——— Measured, UCSD-37
.... Simulated, 1976 Model ($E = 667000$ dynes/cm^2, $\nu = 0.48$, Solid El.)
- - - - Simulated, 1977 Model ($E = 667000$ dynes/cm^2, $\nu = 0.4975$, Solid El.)
-.-.- Simulated, 1978 Model ($E = 6670000$ dynes/cm^2, $\nu = 0.48$, Split E. El.)

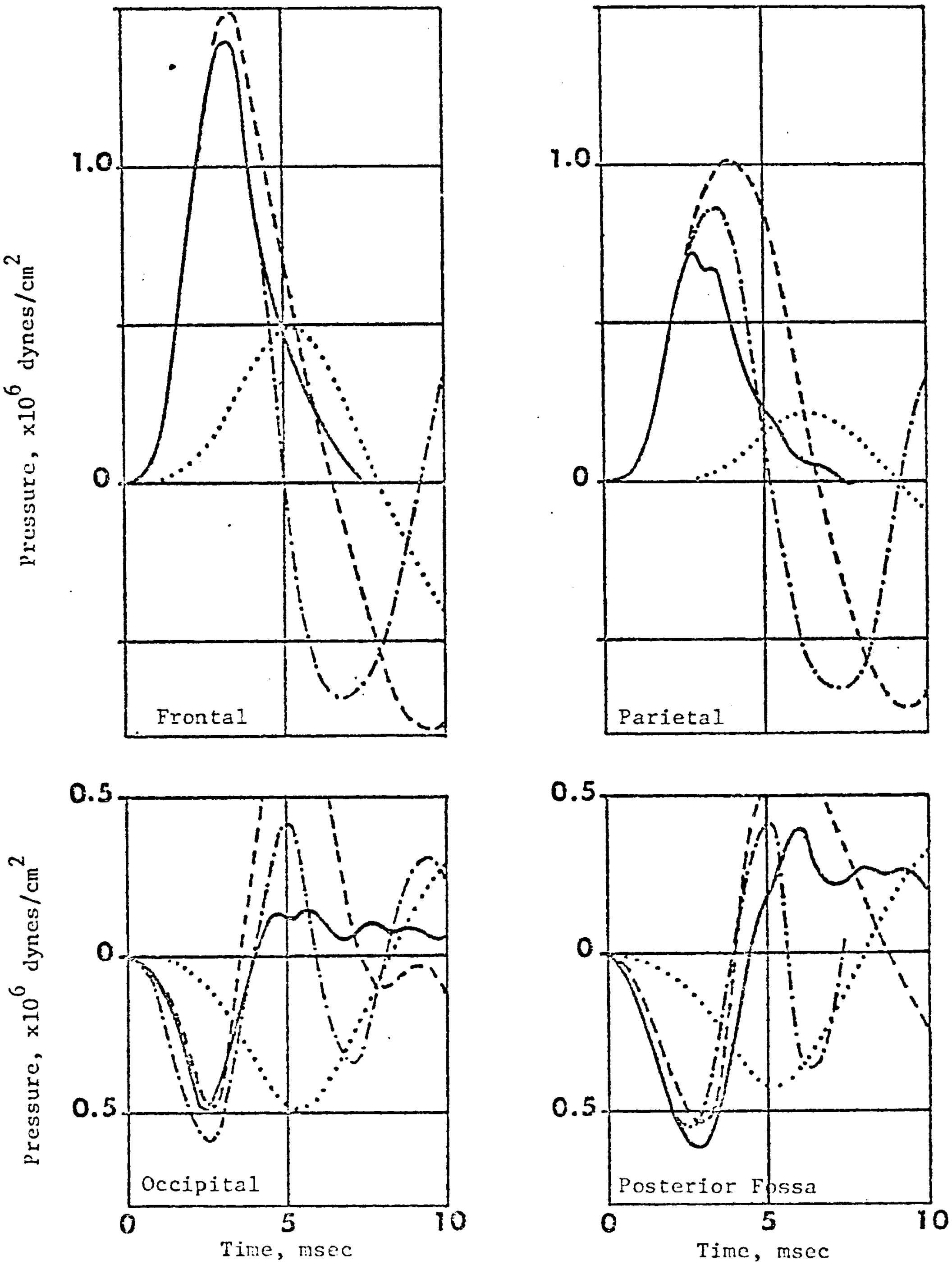

Figure 1. Measured and Computed Intracranial Pressures for UCSD Test 37

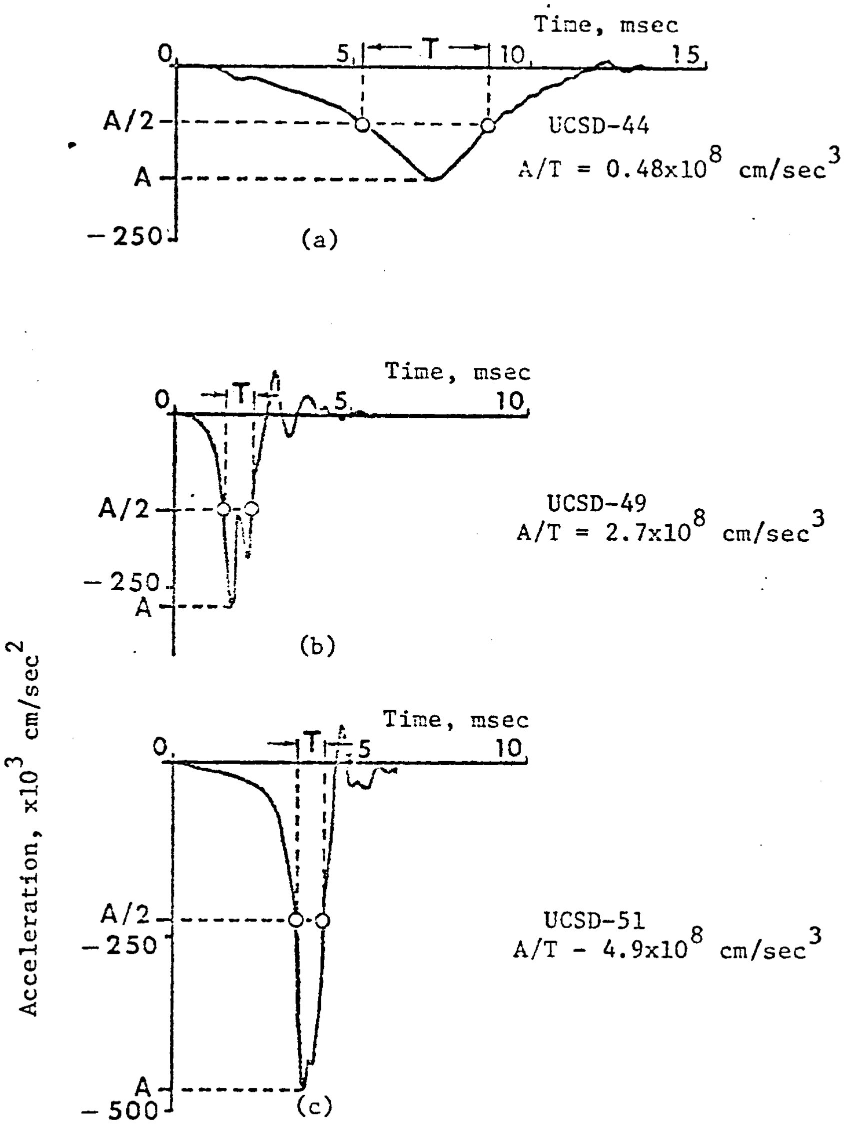

Figure 2. Measured Head Accelerations for UCSD Tests 44, 49 and 51

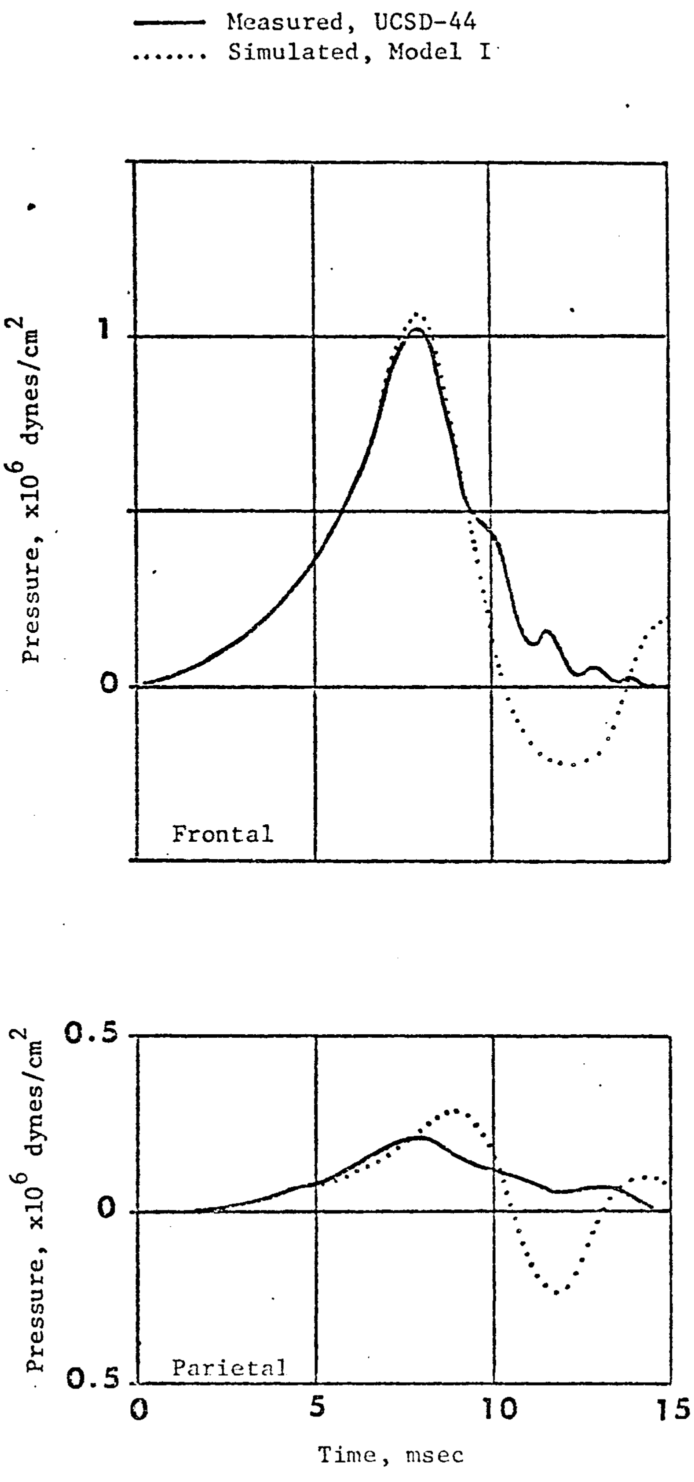

Figure 3. Measured and Model I Computed Intracranial Pressures for Test 44 (Padded Frontal Impact)

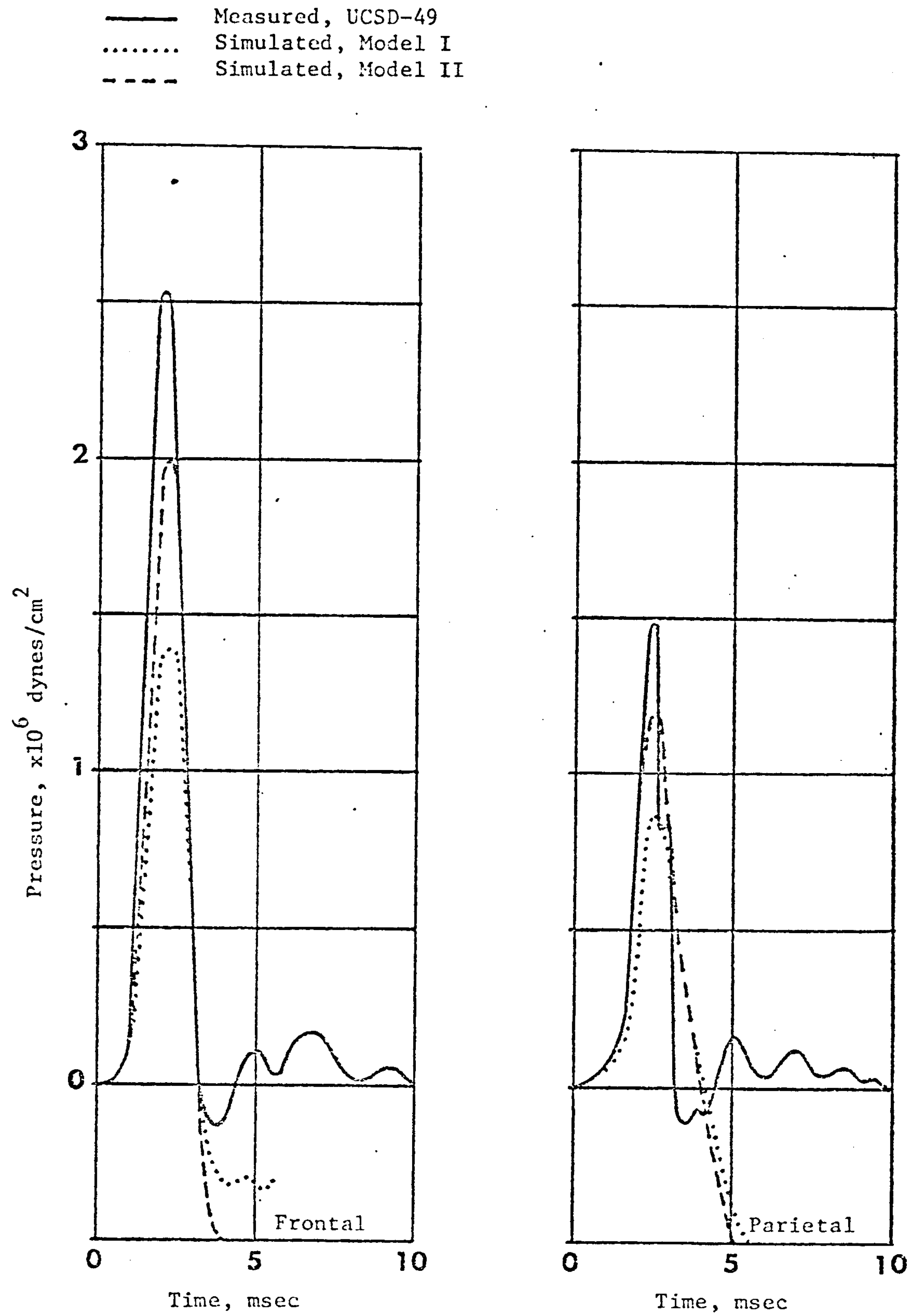

Figure 4. Measured and Model I and II Intracranial Pressures for UCSD Test 49 (Unpadded Head Impact)

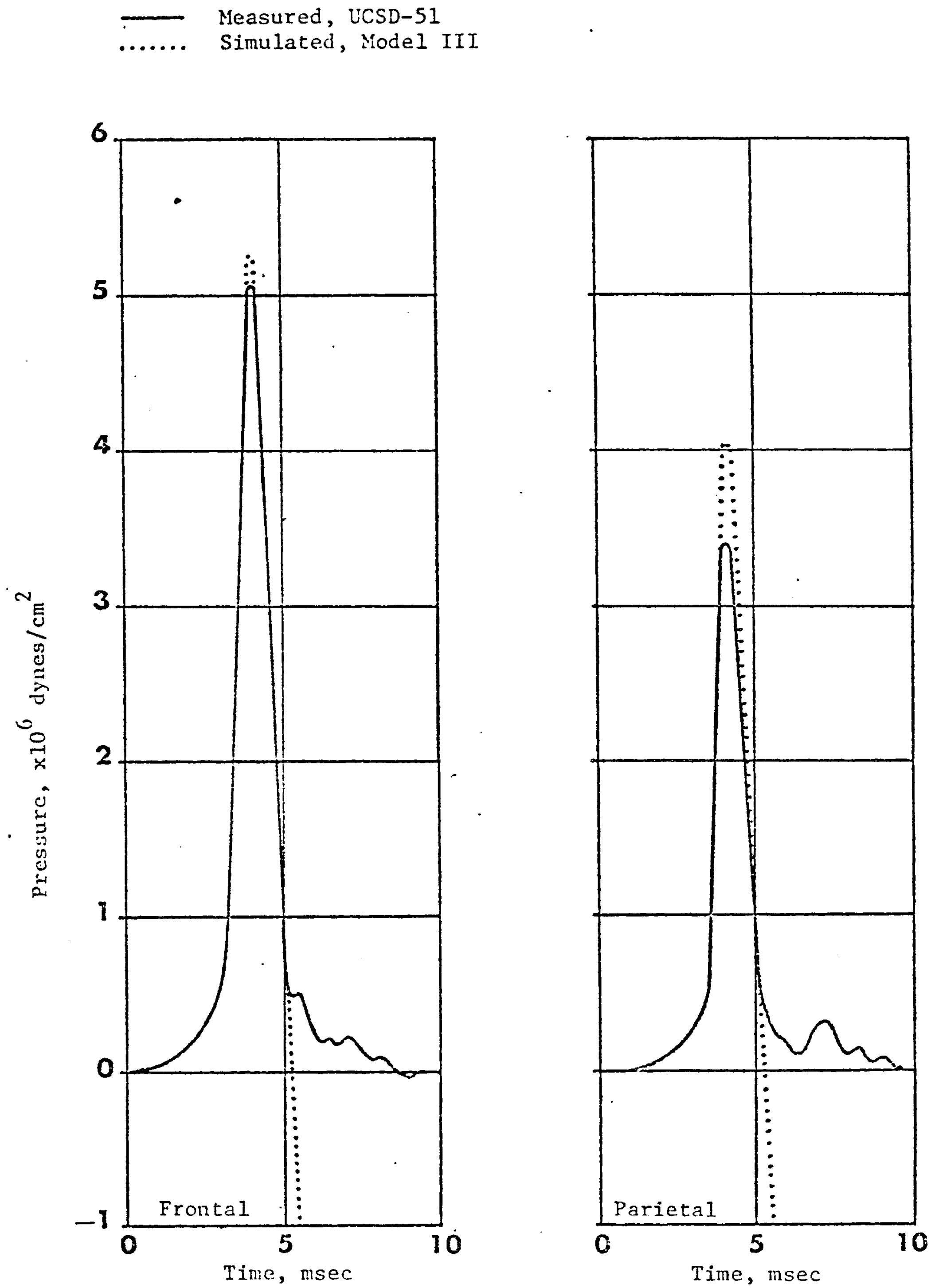

Figure 5. Measured and Model III Intracranial Pressures For UCSD Test 51 (Ineffectually Padded Head Impact)

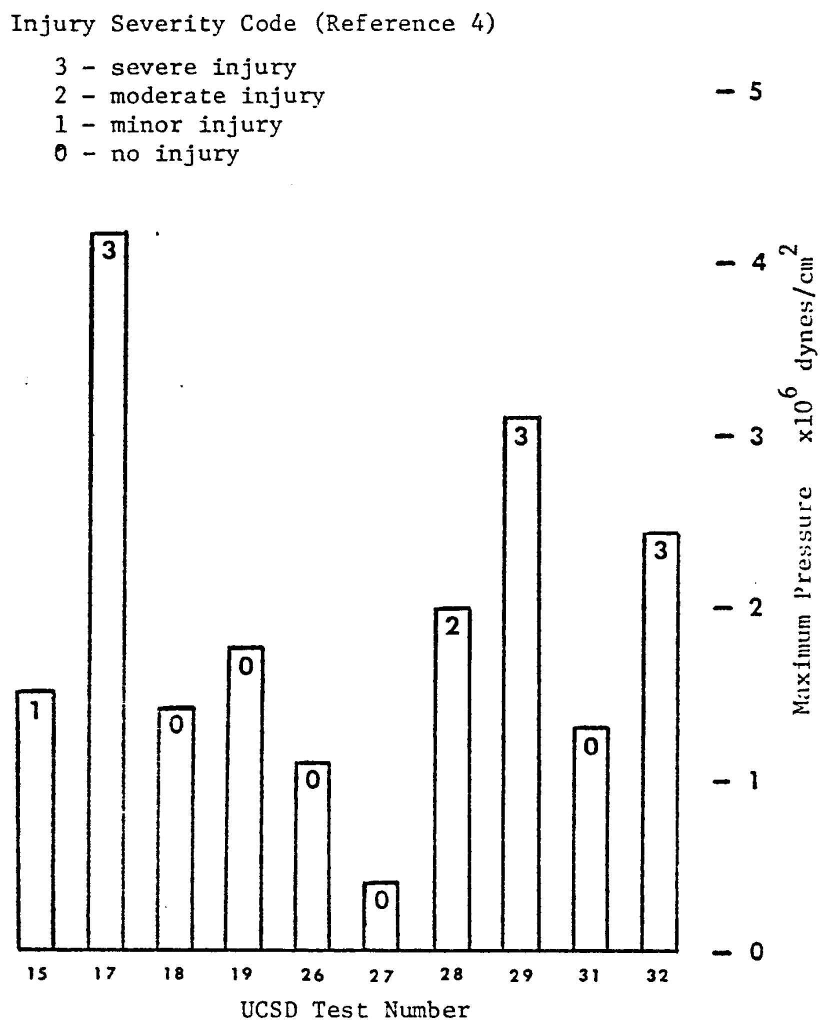

Figure 6. Comparison of Injury Severity and Computed Maximum Frontal Pressures for Ten UCSD Head Impact Tests

Injury Severity Code (Reference 4)

3 - severe injury
2 - moderate injury
1 - minor injury
0 - no injury

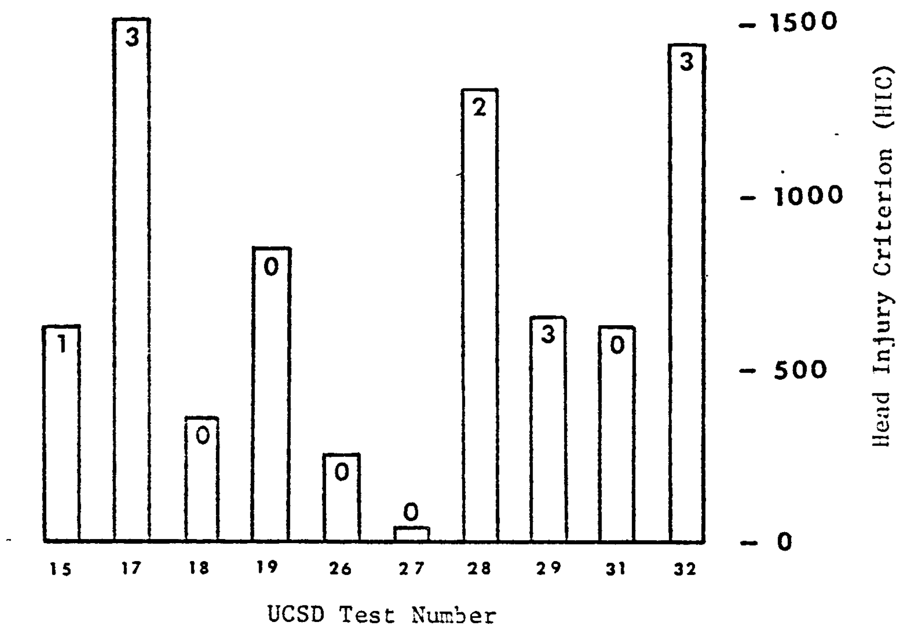

Figure 7. Comparison of Injury Severity and Head Injury Criterion Index for Ten UCSD Head Impact Tests

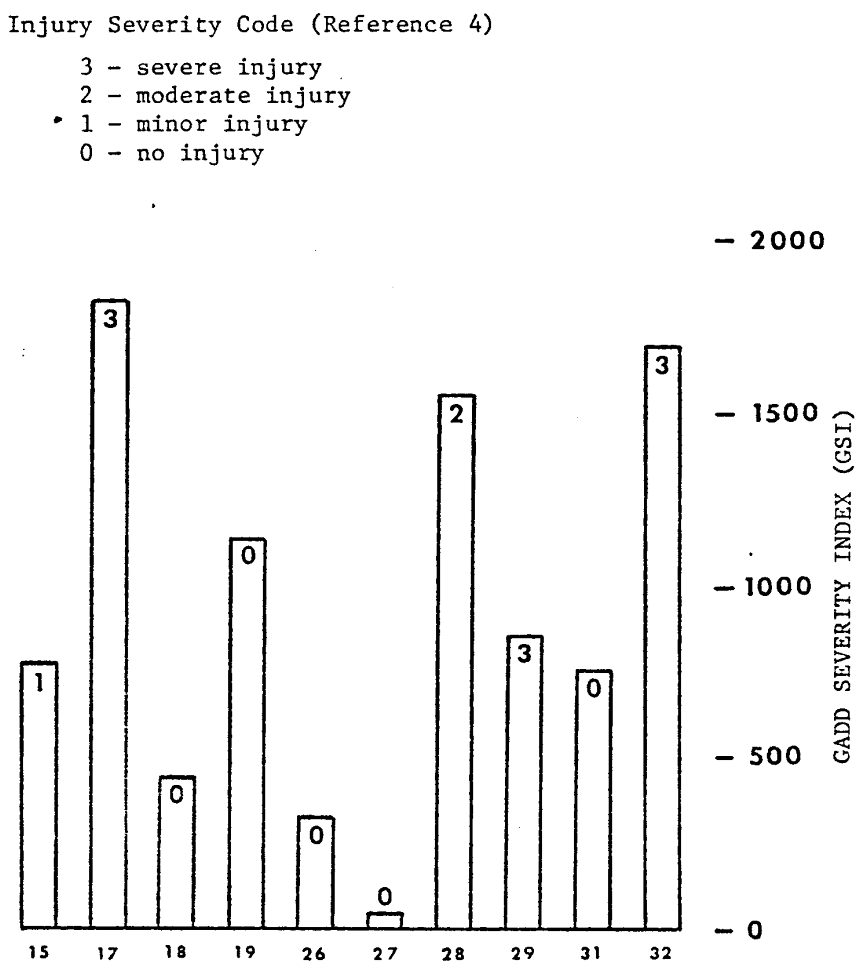

Figure 8. Comparison of Injury Severity and Gadd Severity Index for Ten UCSD Head Impact Tests