CORRELATION BETWEEN BRAIN INJURY AND INTRACRANIAL PRESSURES IN EXPERIMENTAL HEAD IMPACTS

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Background.

Since 1974, finite element models have been used to study the dynamic response characteristics of the brain. Although this critical region of the human body does not resemble a structure in the usual sense, structural analysis procedures have been successfully employed, and computer simulations of head injury events performed. Models of the human brain and experimental test animal brains (a baboon and a monkey) were developed. Because correlation with experimental results is the most important requirement for any biodynamic model, measured brain displacements, pressures and injury data were obtained from government agencies, universities, and research centers throughout the United States. During this validation phase, the models were redesigned to increase their accuracy. Also, as better numerical solution methods became available, the computerized solution procedures were improved.

The history of the development can be summarized as follows. First, elongation and compression of the brain stem were simulated for full ventroflection and extension of the neck (Ref 1 and 2). Displacements on the surface of the brain during impact and inside the cerebrum during head vibration were also investigated (Ref 3). Motion of the intact brain relative to the skull was found to be very limited, contrary to previously accepted findings. (Misleading large displacements were obtained in early lucite calvarium tests (Ref 4). In these tests the skull cap, scalp and underlying dura were replaced with a lucite calvarium. The large relative displacements observed during impact were a consequence of the altered structure, the removal of the dura and connecting tissue. They were not representative of the normal invivo state.)

Some of the most interesting results were obtained in correlating measured and computed intracranial pressures for impact events. The model accurately predicts pressure (stresses) throughout the brain. It was shown that pressures developing in the test animal and human brains for the same head acceleration are different in magnitude, frequency content, and distribution (Ref 5). As a result direct extrapolation from animal to the human is impossible. Recently, the effect of impact surface padding on intracranial pressure was investigated (Ref 6). A comparison of brain pressures in padded and unpadded impacts showed that padding reduces the pressures in the brain and minimizes brain contusions. A similar pressure difference was observed in helmeted and unhelmeted head impacts and is reported in this paper.

Analysis.

Standard linear structural analysis procedures are followed for the most part; only two modifications to accommodate the unique requirements of the brain were made. One modification was at the element level, the other was in the forcing function. Two finite element computer programs (Structural Analysis Program SAP and Engineering Analysis Corporation's program EASE2) were updated to reflect these modifications.

The models are composed of three-dimensional brick and membrane elements (Figure 1). The brick elements represent the soft tissue and contained fluids, and the membrane elements represent the partitioning internal folds of dura, the falx and tentorium. The unique shape of the brain is approximated by the external shape of the assembled elements. To simulate the opening at the base of the skull, the foramen magnum, elements representing the transition from medulla to cervical cord are free to move in the superior-inferior direction.

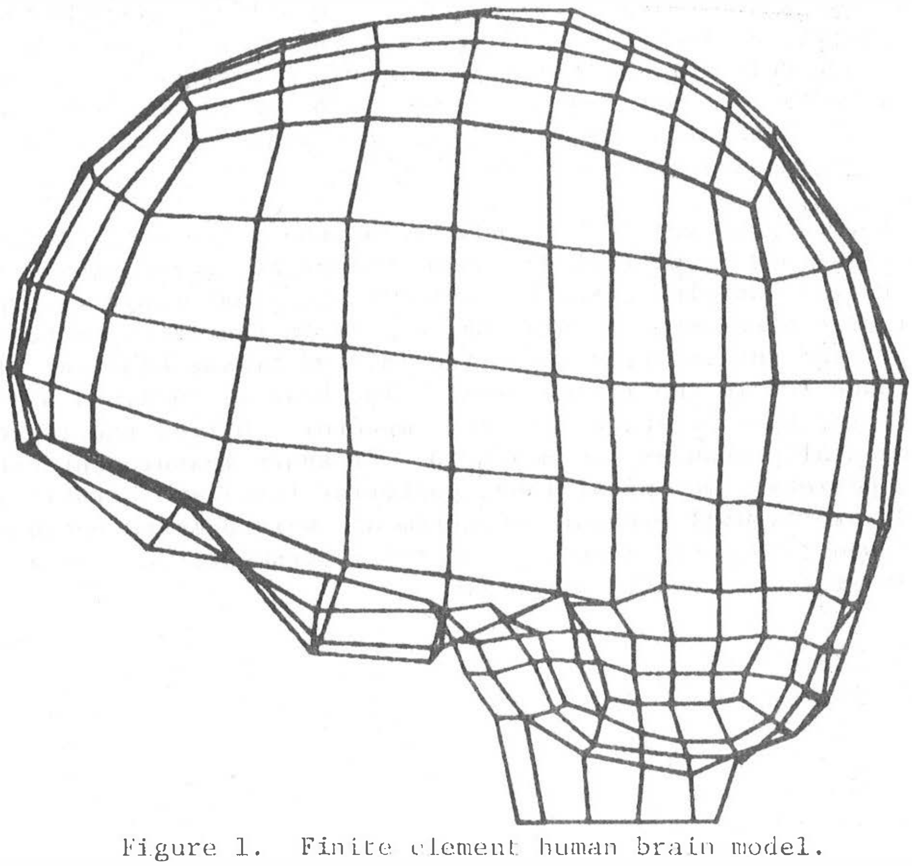

Figure 1. Finite element human brain model.

Because the brain material is nearly incompressible, a special element was developed (Ref 7). (Standard structural analysis elements are not accurate for nearly incompressible materials because the stiffness terms approach a numerical singularity.) In the new element the equations are separated, and a reduced one-point quadrature used on the dilatation energy part. Consequently, the element is referred to as the split energy element.

The system matrix for the model is compiled in the computer by developing the equations of motion for each node (element corner point). Three degrees of freedom are permitted for each interior node, while surface nodes (except at the foramen magnum) are prescribed to move with the skull. Relative coordinates are used to avoid inaccuracies caused by large head displacements and rotations. That is, nodal displacements are measured relative to a skull fixed anatomical axis. Known rigid body head accelerations are considered separately and included on the right side of the equation as inertial forcing functions. Although the structural analysis programs had to be modified to include these terms, the analysis is greatly simplified. Using this technique the skull, scalp, face and neck need not be simulated. Measured head accelerations are used instead of applied forces. Mathematically, the brain model is forced to move in space, as the brain moves in the test. External geometry of the model is maintained to simulate the skull's inner surface.

Experiments.

Two types of experiments are used, (a) head impacts on helmeted and unhelmeted unembalmed human cadavers (Ref 6) and (b) head impacts on helmeted headforms (Ref 8). All are midsagittal frontal impacts. Impact energy is varied in the cadaver tests, producing a wide range of head accelerations. Intracranial pressures are measured and used to confirm the model calculations. In the injury assessment, careful gross and microscopic pathologic studies of the cadaver brain are performed (Ref 9). The headform tests were re-enactments of actual head impacts. Helmet damage in the crash and in the test are approximately the same. The actual brain injuries sustained by the crash victims are used in the pressure-injury correlation.

Methodology for the cadaver tests conducted at the University of California Medical School at San Diego (Ref 9) can be summarized as follows. Prior to impact the cranial vascular network and cerebral spinal fluid were pressurized to nominal invivo pressures. India ink is used in the arteries to indicate brain injury. The ink solution turns the injured tissue black as bleeding turns the tissue red in the living brain. The head is inclined 45° and the blow is delivered by a constant velocity impactor. During the impact, six or eight intracranial pressures are measured. Pressure measurement sites include the frontal, parietal, occipital lobe, posterior fossa and carotid artery. In the helmeted head frontal pressure measurements were deleted because the helmet liner would contact frontally mounted instrumentation. Biaxial head accelerations were measured in all tests.

The headform tests (Ref 8) were conducted at the U.S. Army Aeromedical Research Laboratory at Fort Rucker, Alabama. Helmets involved in Army aircraft accidents were retrieved and examined. It was assumed that the damage was related to force experienced by the crash victim. In the experiment, the impact was simulated by attaching a similar helmet to a humanoid headform and dropping it onto a surface of appropriate shape. Drop heights were varied to obtain comparable helmet damage. During the test, headform accelerations and impact forces were measured. These traces resemble those obtained in the cadaver helmeted tests. Although the ability of the headform to reproduce the accelerations of the human head may be questioned, the variation does not appear to be significant in frontal impacts.

Results.

All of the impacts were simulated using the finite element model of the human brain. In the cadaver tests where pressures were recorded (Ref 10), the measured and computed intracranial pressures are compared. In tests where injury data are available (cadaver unhelmeted head and helmeted headform simulations) the peak pressure is correlated with injury. The brain responses in protected (helmeted) and unprotected head (Ref 6) impacts are compared.

Pressure distribution in a typical helmeted head impact is presented in Figure 2. Positive pressures develop in the frontal lobe, near the impact; negative pressures develop in the posterior fossa, opposite the impact. Parietal pressures are positive and resemble the frontal pressures, but have a lower magnitude. All of the pressures in the cerebrum and cerebellum are of short duration and quickly dissipate. They occur prior to any significant head rotation. Not presented are some pressures which develop after 20 msec at the skull opening (the foramen magnum) and just outside the cranial cavity in the carotid artery. Because the model does not produce corresponding pressures late in the event, these pressures must result from a response external to the brain, such as neck rotation.

Pressure gradients also develop in the unprotected heads, but the traces can be quite different from those shown in Figure 2. In an unprotected head hitting a hard surface (Figure 3), the pressures are short and spike shaped in the frontal and parietal lobes. When the impact surface is padded, the pulse is lower and of longer duration (Figure 4). In order to simulate this range of impacts the effective compressibility in the model was changed (Ref 6). Poisson's ratio is varied from 0.499 to 0.48 to represent the pressure release provided by flow into and out of the cranial cavity. A comparison of Figures 2, 3, and 4 shows that the pressures in helmeted heads are more like pressures in heads impacting well-padded surfaces. Lining in helmets, and padding on impact surfaces, tend to have the same function; they increase the acceleration rise times and eliminate spike-shaped intracranial pressures. The pressure pulses are longer and of lower magnitude.

The pressure-injury relationship in the protected head appears to be the same as that developed for the unprotected head. Injury severity versus peak intracranial pressure is shown in Figure 5 for both test series. Pressures above 34 psi are associated with head injury. Unfortunately, the injury information and number of tests is insufficient for a good comparison. Only six headform tests are available and the injuries are all at, or near, the ends of the scale - Abbreviated Injury Scale (AIS) values of 0, 1, and 5.

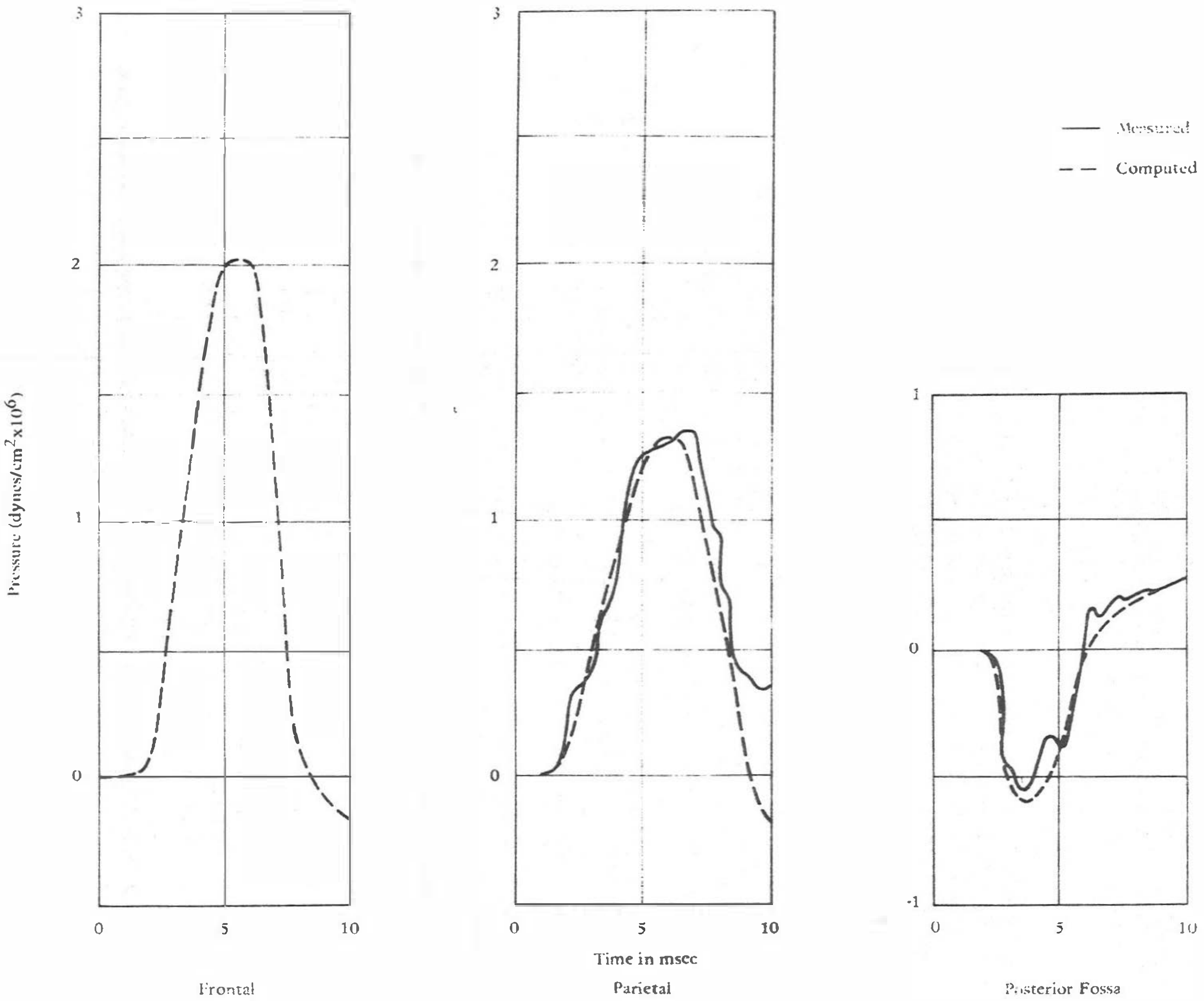

Figure 2. Intracranial pressures in a protected (helmeted) head impacting a hard surface.

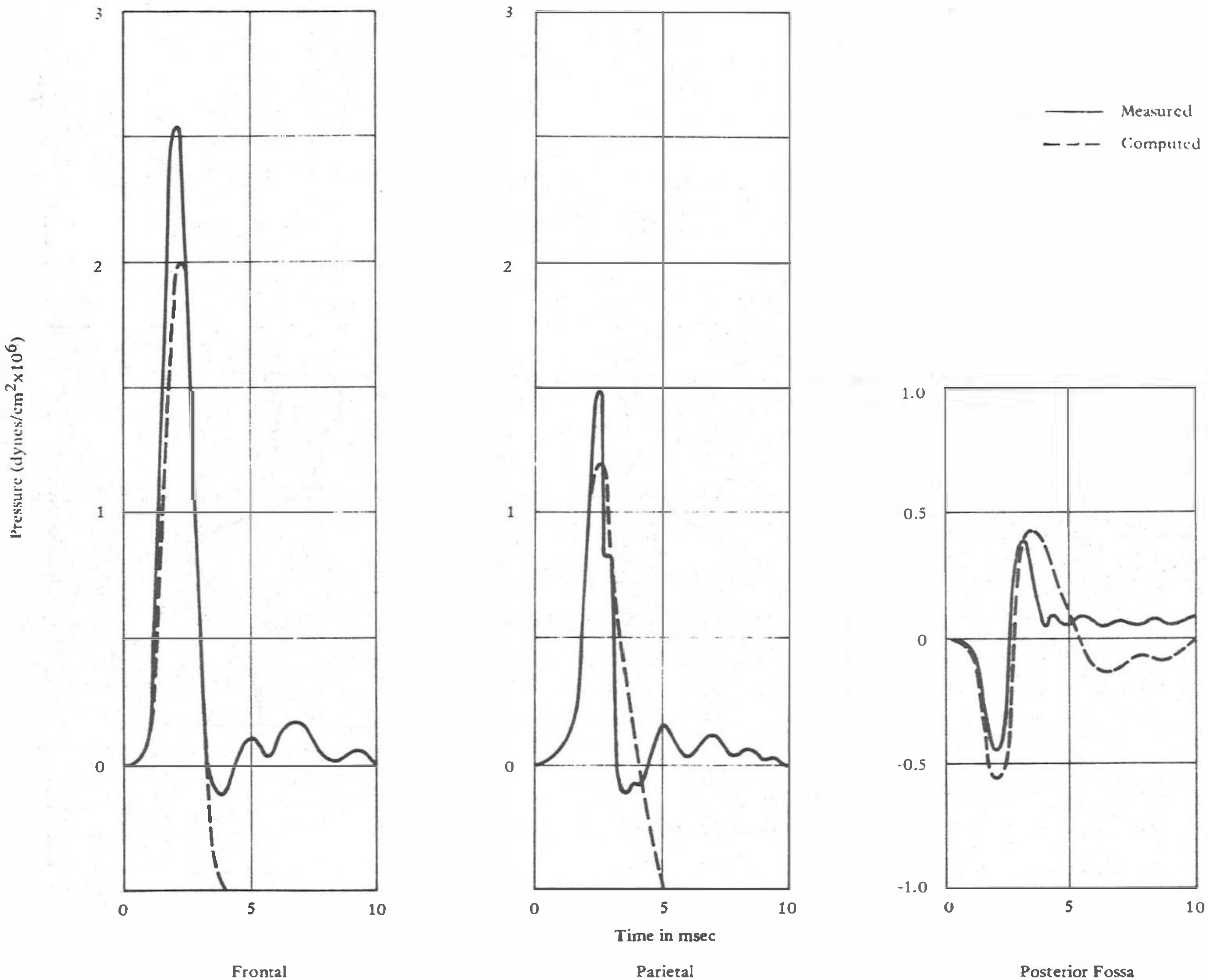

Figure 3. Intracranial pressures in an unprotected head impacting a hard surface.

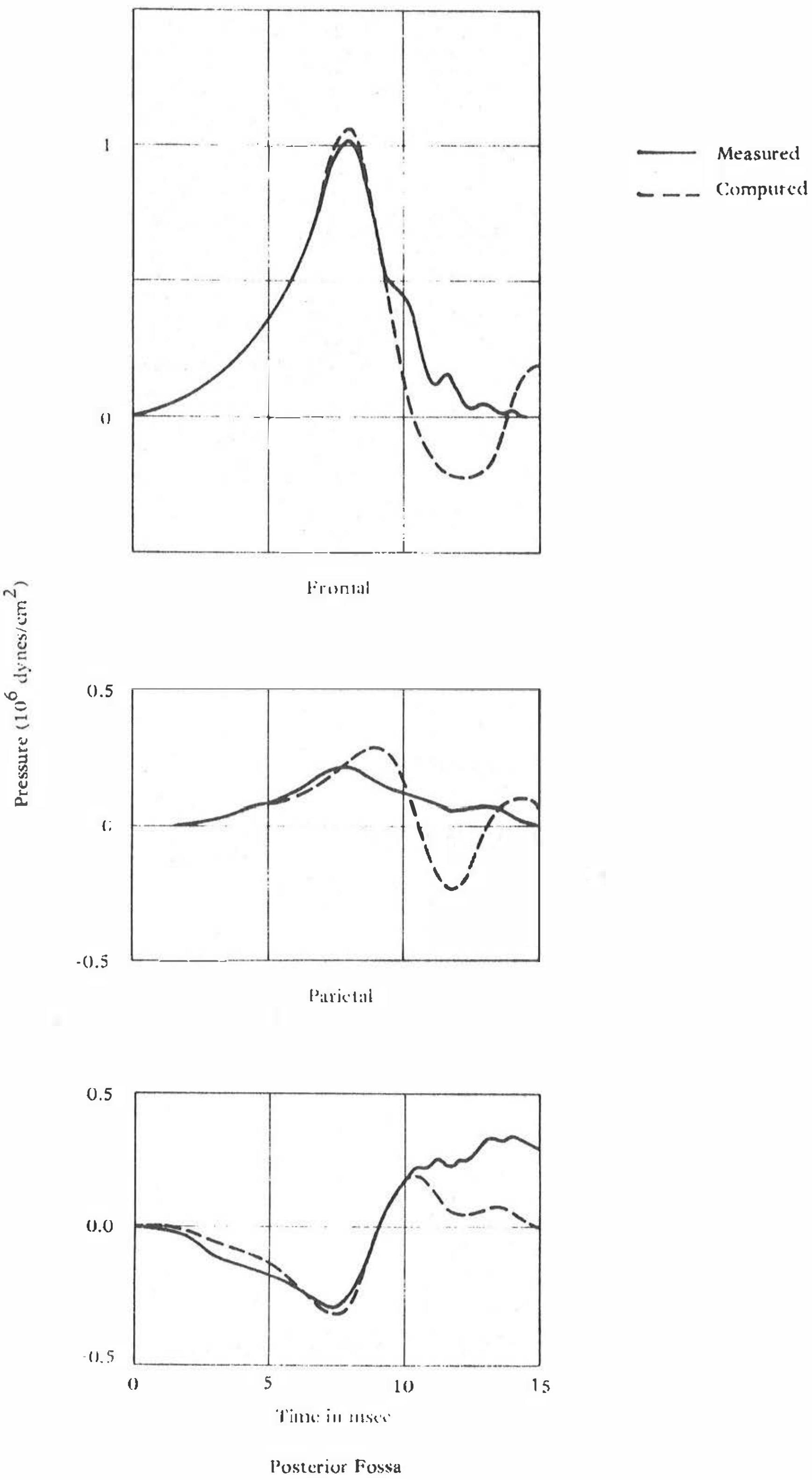

Figure 4. Intracranial pressures in an unprotected head impacting a padded surface.

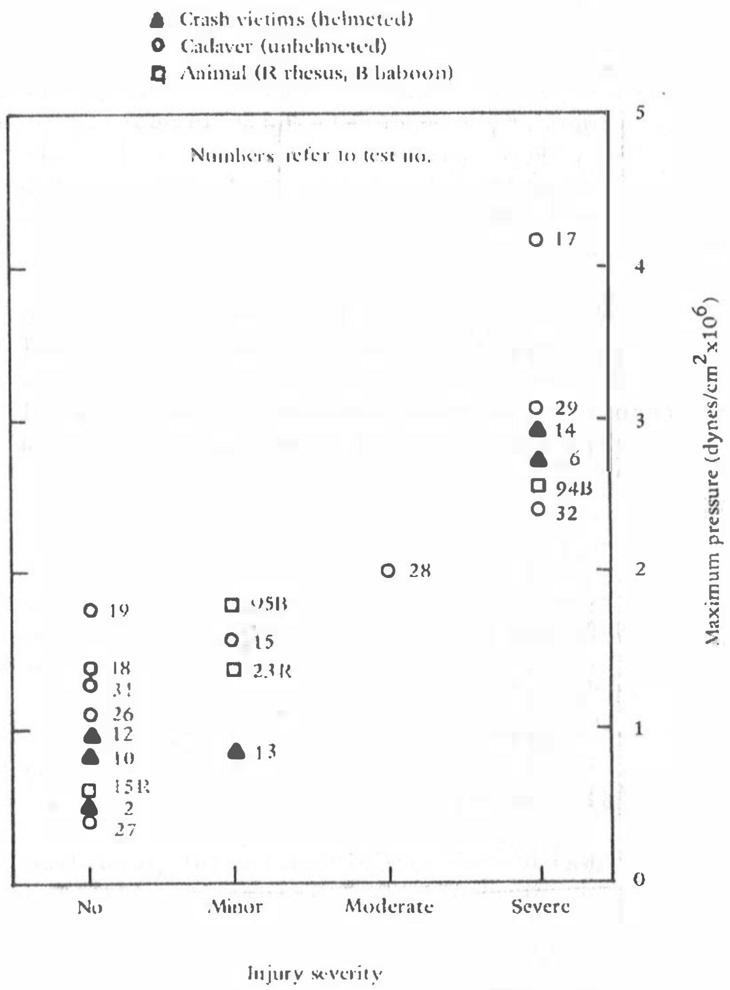

Figure 5. Comparison of injury severity and computed maximum pressures for six Army aircraft crash victims (Ref. 8), ten UCSD human cadaver head impact tests (Ref. 10), and four HSRI live rhesus and baboon head impact tests.

Additional data points in the AIS 2, 3, and 4 range are needed to confirm that the pressure injury relationship is the same. Also, in the unprotected head contusion most often correlates with high pressures. Contusions were not mentioned in the brief accident injury reports. There are several possibilities. Contusions may not have occurred, or they may have occurred and not been detected or reported. They may have been observed and thought to be of lesser importance than the injuries listed. A more detailed description of the injury is important in establishing any future injury criterion for the brain.

Conclusions.

In all head impacts (helmeted or unhelmeted head) a pressure gradient develops in the brain due to head motion; high positive pressures develop near the impact site and negative (less than atmospheric) pressures develop opposite the impact. The finite element model can accurately predict these pressures provided the Poisson's ratio is varied to accommodate pressure release in the longer duration head accelerations. Intracranial pressures in the helmeted head have a characteristic shape. They are, in general, lower and of longer duration. High frequency components have been eliminated by crushing of the helmet liner. The sharp spike-shaped head accelerations and pressures which characterize a hard surface head impact do not develop. Brain response in the helmeted head is more like the response which occurs when a head impacts a well-padded surface.

The same general pressure-injury relationship seems to apply to the helmeted and unhelmeted head impacts; however, additional simulations are required to substantiate this observation. Pressures above 34 psi are associated with injury. The results show that padding in the helmet liner, or on the impact surface, can be very effective in reducing intracranial pressures and minimizing brain injury.

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