A STUDY OF SCALING AND HEAD INJURY CRITERIA USING PHYSICAL MODEL EXPERIMENTS

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INTRODUCTION

Primate studies conducted in our laboratory have produced the spectrum of diffuse head injuries from mild concussion, through prolonged coma with diffuse axonal injury (DAI), to bilateral acute subdural hematoma (ASDH). The loading conditions employed in these experiments were restricted to well-distributed, angular accelerations with the center of rotation in the low cervical spine region for sagittal and coronal plane motions (pure rotation was used in the horizontal plane only). Although each of these pathophysiological entities constitutes a head injury, we have found that their tolerance levels are distinctly different. We believe that the variation in magnitude and topographic distribution of the brain tissue deformations associated with a specific set of loading conditions will help to explain these differences.

At the present time, criteria for head injury relate the inertial response of the head, as a result of impact or impulsive loading, to a tolerance or threshold for brain injury. Perhaps this relationship can be defined in greater detail if we understand the intermediate consequences of this complex process; specifically, the variations in the field parameters of displacement, strain or stress within the brain, and their relationship to the kinematics of dynamic loading.

This report discusses the use of some simple physical models of the skull-brain structure as an experimental tool for the measurement of the deformations experienced by the surrogate brain under inertial loading conditions. Determining this transfer function between the gross description of the head loading conditions and the response of the structure is an important step in the direction of elucidating the various mechanisms of head injury. The animal model has enabled us not only to study the pathophysiology and morphology of injury (with the goal of improved diagnosis and therapy), but also to provide useful biomechanical data relating the kinematics to the specific injury. However, in order to improve upon existing head injury criteria, scaling these subhuman primate data to man is essential. The physical model experiments discussed in this report should prove helpful in this regard.

BACKGROUND

Previous investigators have employed a variety of experimental techniques to measure the response of the in-vivo brain, cadaver specimens, or model systems to dynamic loading. Holbourn [1], in one of the earliest applications of photoelasticity to experimental biomechanics, observed the shear stress distribution in a wax-skull, gelatin-brain structure subjected to rotational acceleration. Holbourn related the degree of strain indicated by the model to the amount of stretch of the neural and vasclar fibers in the region, and thus, to injury severity. Prudenz and Sheldon [2] viewed the motions of the cerebral cortex in living monkeys through a transparent calvarium in response to angular acceleration of the head. Displacements of approximately 1 centimeter were reported. Subsequently, Hodgson, et.al.[3], and Gurdjian, et.al.[4], using monkey cadaver head hemisections and gelatin models subjected to occipital impact, reported significant strains in the region of the foramen magnum associated with this loading condition.

More recently Aldman, Thorngren and Ljung [5] conducted a qualitative study to demonstrate that brain motion is altered with changes in skull geometry and the nature of the skull-brain interface. The simplified plastic skulls were filled with a silicone gel and markers were placed on planes of interest within the transparent gel to observe relative displacements when the molds were subjected to rotational deceleration from constant angular velocity.

Although this line of research has not yet played a major role in improving upon our current expressions for head injury tolerance criteria, the findings of these investigators have clearly influenced our concepts of head injury mechanisms and have significantly impacted upon the direction of research programs. The work presented in this report is a continuation of these physical model experiments with the ultimate goal of developing models which provide reasonable approximations of the structural response of the head to dynamic loading.

METHODS

The experimental system used to provide the loading conditions for the physical models is identical to that used in our primate studies. This device is depicted in Figure 1 and consists of a custom-designed, 6-inch Bendix HYGE (R)

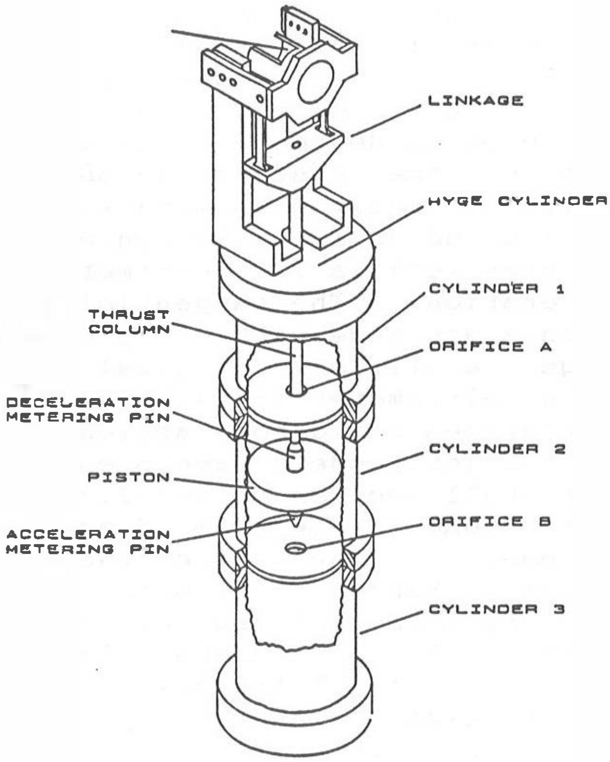

FIGURE 1

actuator and linkage assembly capable of delivering a well-distributed impulsive load to the primate head. The linear actuator is composed of a piston and thrust column which travel within the cylinder which is divided into three compartments by internal orifice plates. The design permits fine control of the acceleration and deceleration phases of the piston motion by shaping the contours of the metering pins which are located on the top and bottom faces of the piston. The relative acceleration and deceleration amplitude and duration as well as waveshape can therefore be independently controlled. The linear actuator is coupled to a linkage which converts the purely translational motion of the column into a rotational motion of the fixture which rigidly supports the primate head or physical model. Relatively simple mechanical alterations prior to an experiment permit us to independently vary the direction of rotation in three mutually perpendicular axes, along with the degree of angular excursion and the center of rotation. By careful control of these kinematic conditions it is possible to produce the range of brain injuries previously discussed. It is therefore desirable to study the response of model systems under identical conditions, utilizing well-distributed angular acceleration.

Figure 2 shows the linkage portion of the loading apparatus with a physical model fastened to the side arm of the assembly. All physical models described herein were subjected to either an angular acceleration or a pure centroidal rotation of 65 degrees. The shape of the acceleration-deceleration time history is shown to be an asymmetric profile with a largely predominate, half-sine deceleration phase. The radius from the center of rotation (CR) to the center of mass (CM) is 7.3 centimeters in all cases of angular acceleration. The tangential acceleration-time history measurements are made with a piezoelectric accelerometer/ charge amplifier/ digital recording system (Endevco). The accelerometer is fastened to the linkage assembly which undergoes rigid body motion in a plane.

A series of physical models have been constructed which consist of either skull section or idealized right circular cylindrical geometries. Figure 3 demonstrates two such model configurations. In these cases the skull is the anterior portion of a baboon skull which was separated from the posterior half by a coronal saw cut. The anterior portion is shown potted into an aluminum cylinder using Castolite Resin (Buehler, Ltd.). This cylinder is machined so as to adapt onto the kinematic linkage. The edge of the skull is recessed from the transparent cover plate forming a space, 3 mm deep, which is ultimately filled with water. The material used as the surrogate brain is Silicone Gel System (Dow Corning) which is mixed in a 50-50 ratio of catalyst to polymer. This mix has been shown to yield a material whose modulus of elasticity is approximately equal to that of brain. The gel is optically transparent and can be cast in layers which will adhere to themselves. Consequently, it is possible to pour a layer of gel to a level which represents a plane of interest and allow the gel to cure at that point. A black enamel paint is then used to construct an orthogonal grid of approximately 5 mm square spacing. The next layer of gel is then applied and attaches to the painted surface as well as other boundaries. Water serves to permit the free surface to move relative to the transparent cover plate.

In the case of the right circular cylindrical geometry the gel is poured directly into the aluminum container. In both instances the grid is located approximately 2 cm below the edge of the cylinder, and although the cylinder and skull are finite in length, we feel that the edge effects can be neglected in these geometries.

In their present configuration the models have a no-slip boundary condition on all surfaces with the exception of the transparent cover plate, as noted previously, and the material porperties of the gel are approximately linear elastic, homogeneous and isotropic. The material properties of the gel at a 50-50 mix vary somewhat from model to model which is most probably a function of the temperature and humidity during the cure cycle. In order to normalize for these var-

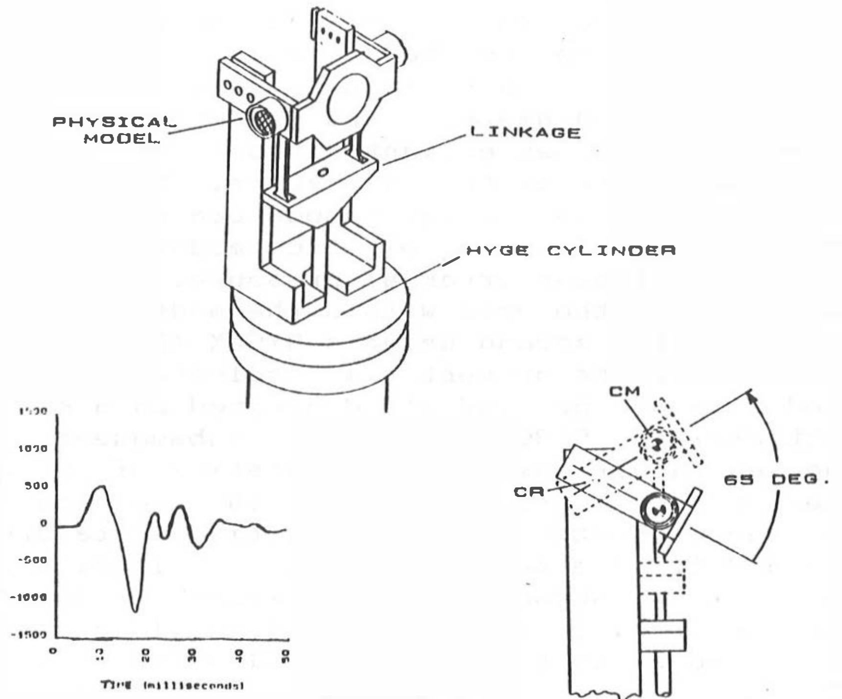

FIGURE 2

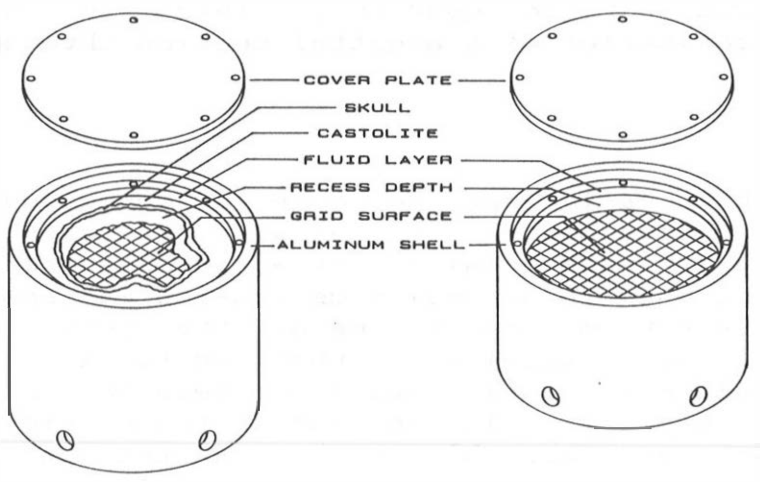

FIGURE 3

iations, mechanical indentation of the gel surface is performed non-destructively. The load-deflection characteristics of the gel are recorded from an isometric force transducer in series with a circular-faced indentor (approximately 1.5 mm in diameter) and a micrometer drive. The material modulus is then estimated based upon the solution for local indentation of the half-space. Similar studies on the cortical surface of the baboon brain yield a value of approximately 1 psi for the elastic modulus, which is a linear function of mean arterial pressure.

Deformation of the grid within the model is photographed at 4000 frames per second using a HYCAM (R) camera (Redlake Corporation). At the present time, selected frames from the high speed film are printed and digitized on a bit-pad using a Hewlett-Packard 9836 computer. Subsequent analysis, including an algorithm for computation of the non-linear strain tensor is also accomplished with the lab computer. Computer reconstructions of the digitized data are shown in Figures 4 and 5. The selected frames are labeled where "t" is time in milliseconds. Large shear strains can be observed in both the coronal plane half-skull as well as the cylindrical model at t=14 and 17 milliseconds respectively, which corresponds to the temporal occurrence of peak tangential deceleration. Rigid body motions of the skull or cylindrical container were subtracted from the data during the digitization process.

This report presents the results of five different physical models which include: the anterior portion of the baboon skull with the coronal cut made approximately 1.5 cm posterior to the edge of the formamen magnum, three right circular cylinders of different gel diameters (11.12, 7.94, and 6.35 cm), and one right circular cylinder (diameter = 11.12 cm) with a rigid quadrant. This last model is a simple approximation of a sagittal section through the head.

RESULTS

From his early work Holbourn [1] concluded that since the shear modulus of brain tissue was orders of magnitude lower than its bulk modulus, it would be subject to larger deformations when rotational acceleration was applied to the head. Figure 6 is a comparison of the peak deformations experienced by a model when subjected to: a) pure translational acceleration, b) angular acceleration, c) pure rotational acceleration. In the case of translation, the peak deceleration was comparable to the peak tangential deceleration of the center of mass of the model loaded in angular deceleration. Likewise, in rotation the peak rotational deceleration was matched to the case of angular deceleration as closely as possible. The acceleration-time histories are indicated below each figure. Although the rotational acceleration experiment indicates somewhat larger deformations

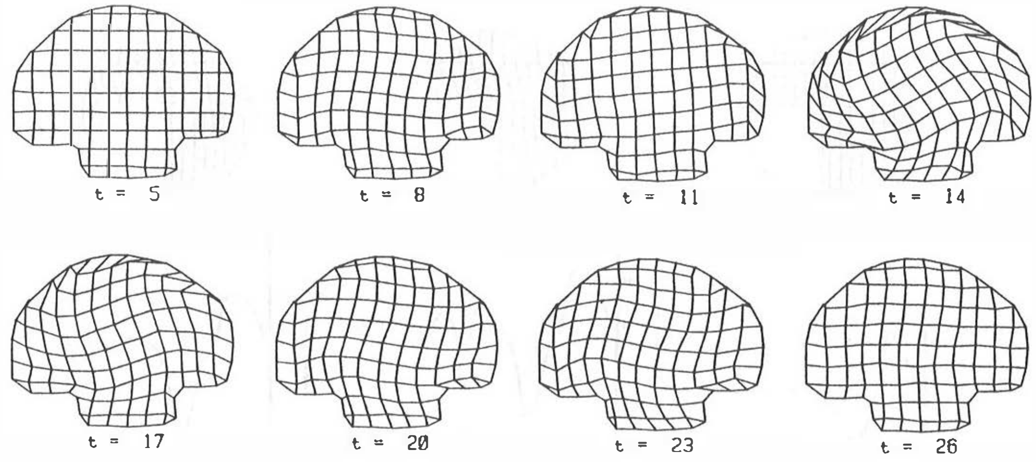

FIGURE 4

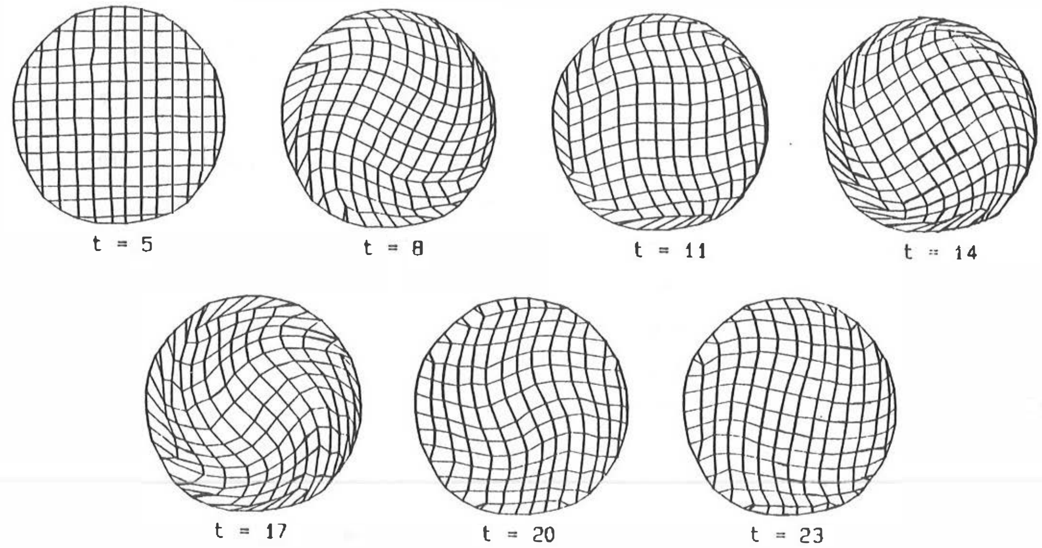

FIGURE 5

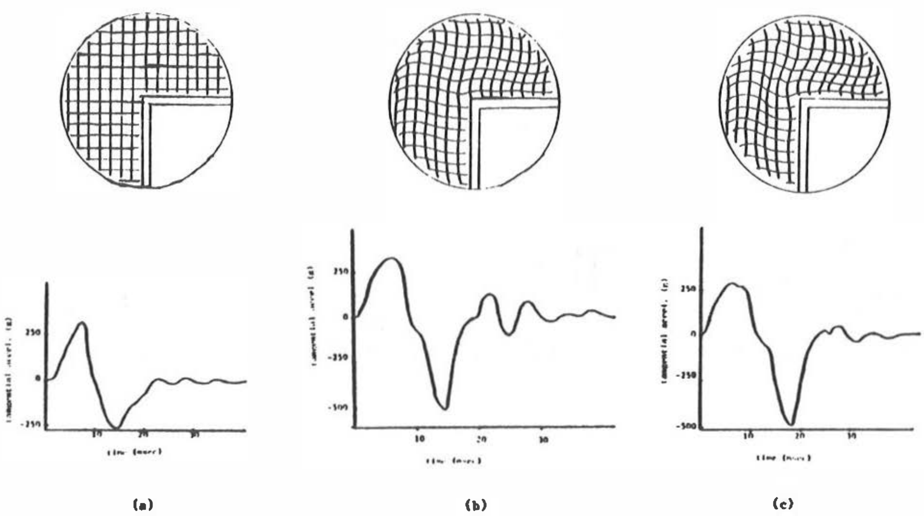

FIGURE 6

EXPERIMENT	MODEL RADIUS(cm)	PEAK ROTATIONAL DECELERATION (rad/sec x10^{-4})	e^{L}	$e^{L} * \frac{E}{E_B}$
A-100	5.560	4.64	.145	.210
A-200	5.560	9.01	.304	.440
B-50	3.970	2.00	.043	.070
B-100	3.970	4.14	.077	.127
B-200	3.970	9.28	.176	.290
C-100	3.175	4.65	.090	.109
C-200	3.175	9.01	.166	.201
C-300	3.175	12.02	.250	.302

TABLE I

than the angular case, it is clear that the distribution is virtually identical. The translational case does not differ from the unloaded conditions within the resolution of the measurement scheme; i.e., the strains are approximately zero. It is also clear as in the previous example that shear waves are generated at the skull-brain interface in the case of rotational acceleration and our example of angular acceleration. When the models do not contain the solid fourth quadrant there appears a rigid-body type rotation of the central core within the gel. The presence of the boundaries in the fourth quadrant disturb this rigid-body behavior and we believe that this geometric constraint plays an important role in producing injury to the deeper structures within the brain.

Shown in Table 1 are the summary data obtained from three right circular cylindrical models. Experiments are designated as A, B, and C with a suffix which indicates the pressure in pounds per square inch used to accelerate the models. The rotational accelerations range from approximately 2 to 12 $\times 10^4$ radians per second squared. Lagrangian strains of the peripheral elements are computed from the single frames of high speed film. The data are then normalized for variations in the moduli of the material. The data described are plotted in Figure 7. As expected, the deformations and consequently, the strains, increase with increasing models size as well as increasing levels of rotational acceleration. The load may be thought of as a torque exerted on the no-slip periphery of the model which is then proportional to the moment of inertia and the rotational acceleration. It may have been this line of reasoning which led Holbourn to conclude that the rotational acceleration which produces some critical level of tissue deformation in a model can be scaled from data obtained in a prototype according to the relation:

$$\frac{\ddot{\theta}_m}{\ddot{\theta}_p} = \left(\frac{M_p}{M_m}\right)^{2/3}$$

where M_p = mass of the prototype
M_m = mass of the model
$\ddot{\theta}$ = rotational acceleration

Since the torque is proportional to the moment of inertia then the rotational acceleration is proportional to $1/r^2$ or to $1/(\text{mass})^{2/3}$. However, strains are generated within the tissue when the skull moves relative to the brain. If the tangential displacement of the skull is given as δ then for a constant rotational acceleration $\ddot{\theta}$ with duration t_D we can write,

$$\delta = \frac{1}{2}\ddot{\theta}t^2\left[R\left(1 - \frac{r}{R}\right)\right]$$

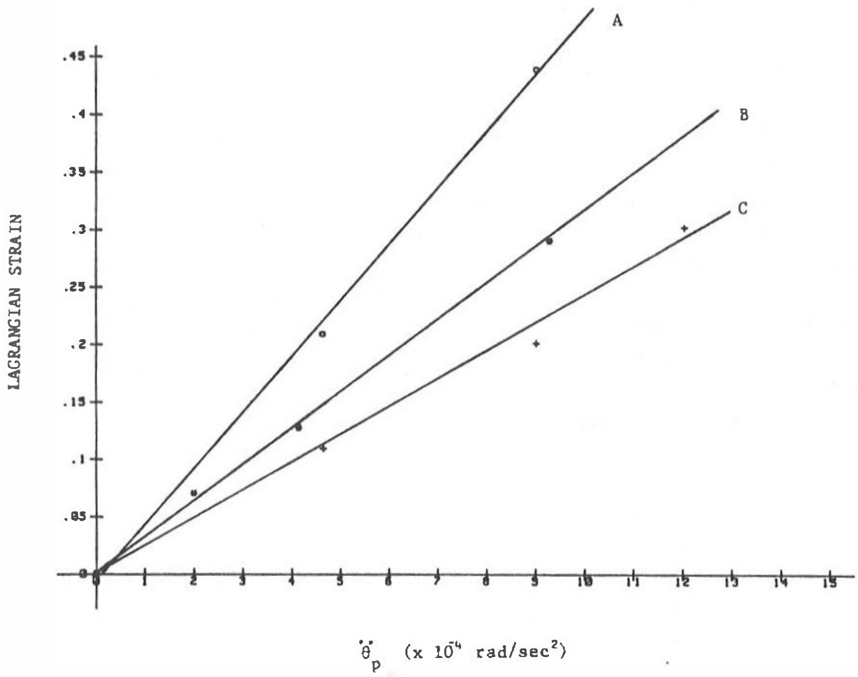

FIGURE 7

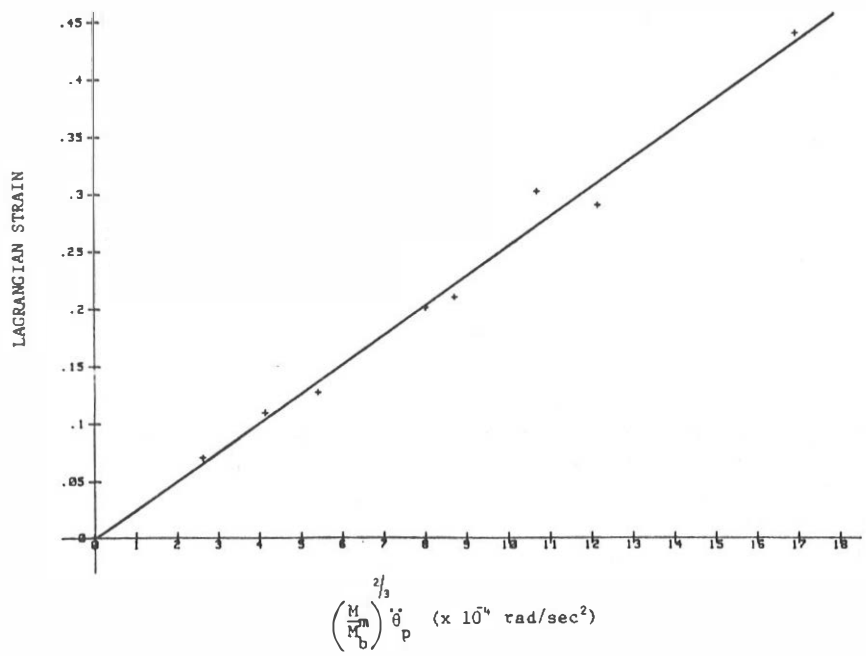

FIGURE 8

where R is the brain radius and r=0 at the skull-brain interface. Each subjacent element of brain tissue will then displace according to the relation,

$$\delta = \frac{1}{2}\ddot{\theta}\left[R\left(1 - \frac{r}{R}\right)\right]\left[t - \frac{r}{V}\right]^2$$

where V is the local tangential velocity of brain.

The shear strain can then be obtained and evaluated at r=0 and $t=t_D$ where it is a maximum,

$$\left.\frac{\partial\delta}{\partial r}\right|_{r=0,\ t=t_D} = \frac{R\ \ddot{\theta}t_D}{V}$$

The shear strain can then scale provided the rotational acceleration goes as $1 / (Rt_D)$. If the pulse duration is proportional to R then $\ddot{\theta} \sim 1 / R^2$ or to $1 / (\text{mass})^{2/3}$ as Holbourn suggested. This relationship is true provided the shear wave is short compared to the brain diameter.

Using this scaling relationship, Figure 8 shows the rotational accelerations of the models scaled to the mass of the baboon brain (since this species has been the one used most extensively in our primate studies). The line shown has a slope of 2.57×10^{-7} and a correlation coefficient of .996. Given all of the idealizations assumed in these model studies, it is this type of relationship which will enable us to compare thresholds for specific injuries as determined by the loading conditions, assuming that tissue deformation is responsible for injury. In order to accomplish that goal additional research is necessary and will be discussed in the summary.

SUMMARY

Dynamic mechanical loads applied either directly or indirectly to the head result in deformation of the intracranial contents. Associated with these deformations are strains and stresses which are generated within the neural and neurovascular tissue. If these field parameters of displacement, strain, or stress exceed some critical level of tolerance, then the discrete neural or vascular tissue elements will respond adversely from a structural or functional point of view. The complicated geometry of the brain and the cranial vault, the constitutive relations for the materials, and the kinematics concomitant with the loading will play major roles in determining the response of the head to direct impact or impulsive loads.

A system has been developed which enables one to study the deformations occurring in model skull-brain systems where the kinematics of the loading conditions are identical to those which are demonstrated to produce a range of specific brain injuries in a subhuman primate model. The simple models presented here suggest that the strains can be determined and the rotational accelerations can be scaled as a function of brain mass to yield a linear relationship between the two.

Models under development in our laboratory at the present time possess the following features:

a. Realistic skull geometries
b. Pure-slip boundary conditions
c. Internal structures including the falx and tentorium
d. A compliant foramen magnum

It is hoped that these features will provide a more realistic representation of the magnitude and topographic distribution of the deformations associated with these loading conditions.

Our goal is to improve upon our current understanding of head injury tolerance criteria. Prudent use of animal and mathematical models, the development of material constitutive properties, the use of experimental stress or strain analysis of physical (inanimate) models, and the study of the response of isolated neural and vascular tissue elements to mechanical trauma will, in concert, help to shed light on this difficult problem.

ACKNOWLEDGEMENTS

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