A LUMPED PARAMETER APPROACH TO SIMULATE THE ROTATIONAL HEAD MOTION

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ABSTRACT

Studies conducted by the University of Pennsylvania have shown that subdural hematoma, diffuse axonal and other shearing type brain injuries are the most common causes of death and disability seen in human head injury, and that angular acceleration is the primary cause. It is, therefore, the main interest of this paper to study and relate the coronal or angular acceleration-induced shear strain that is exerted on the brain and its connective tissue to the tolerance, or threshold of the above mentioned head injuries.

In this study, the human head is modelled as a 3-degrees-of-freedom mechanical system using a lumped parameter approach. The model consists of masses, springs and dampers and has been validated with experimental data obtained by the University of Pennsylvania.

The model was exercised with various loading conditions commonly seen in car crash environments. In addition, the effect of helmet and airbag on rotational head injury were investigated. The resulting dynamic responses of the model were utilized for computing the shearing strain and providing a way to establish some injury criteria for estimating the potential of rotational head injury in actual car crash simulation.

INTRODUCTION

It is estimated that in the United States, approximately four million head related injuries were reported annually. Out of these, 50,000 cases were known to be fatal [1]. Since the human brain is a part of the body which not only controls itself but also the flow of information as a whole, it is not difficult to realize why head-related injuries are often fatal, and are the cause of persisting neurological disabilities among the survivors. To study these problems, research on head injuries, head injury mechanisms, and head injury criteria are necessary. By understanding more in these areas, physicians would have better knowledge in rehabilitating the injuries and engineers would have a better basis for designing effective head protective devices.

The basic mechanisms which produce a head injury may be classified into three categories, namely: contact phenomena, stress wave propagation,

Numbers in [ ] designate references at end of paper.

and acceleration (inertial) effects. As opposed to the contact phenomena and stress wave propagation which are the main cause of focal injuries, inertial effects are the major cause of diffuse brain injuries when the head is subjected to impact or impulse loading. It has been indicated that, although injuries due to angular acceleration comprise only one-third of all injuries, they are the primary cause of subdural hematoma, diffuse axonal and other shearing type brain injuries which are responsible for two-thirds of all mortalities and severely disable survivors [2].

It is for these reasons that the main purpose of this paper is to study the angular acceleration induced shear strain that is exerted on the brain and its connective tissue and relate it to the tolerance and threshold of the above mentioned head injuries. To accomplish these objectives, a rotational head injury model was created. The model represents the human head as a physical system using a lumped-parameter approach. Since most head injury hypotheses suggest that the relative motion of the brain to the skull and the shearing motion within the brain are of critical importance, they are given primary consideration in the proposed model.

BACKGROUND

Over the years, numerous head models have been developed in validating a specific hypothesis or theory of head injury. Amongst all, the earliest analytical structural model was developed by Anzelius who modelled the head as a rigid spherical shell filled with inviscid compressible fluid [3].

More recently, Thibault, et al., have developed some simple physical models using the skull-brain structure as an experimental tool to study the relationship between the rotational/angular acceleration induced shear strain to the acute subdural hematoma and other shearing injuries [4]. The physical models, which represent primate heads, have been constructed to include a skull and surrogate brain materials with a 'no-slip' boundary condition at the skull-brain interface. This structural model was then subjected to various amplitudes and pulse shapes or waveforms of angular acceleration. The shearing strain was computed based on the deformation observed on the surrogate brain. Such strain will be used in validating the proposed head injury model presented in this paper.

METHODOLOGY

From the comprehensive anatomic description of the human head, the brain is connected to the skull by dura, bridging veins, and other connective tissues. Their material properties are similar to those of the brain which are heterogeneous, viscoelastic, and nonlinear in nature. If a model is created such that it closely duplicates the actual physical behavior and the structural characteristics of the head, it will have to involve fewer assumptions and result in a more complex model. This,

however, requires a highly sophisticated and lengthy mathematical technique to obtain the solution of the model, if indeed the model has a solution at all. Therefore, in order to achieve a practical model that is capable of simulating the desired injury mechanisms, many approximations and idealizations have to be made. To consider such a practical model, the following assumptions are made in the proposed head injury model:

1. The skull is assumed to be a rigid body.
2. The brain is considered as a discrete system. It is symmetrical to the axis of rotation with homogeneous, isotropic and linear material properties.
3. The boundary condition between the skull-brain interface has a non-slip condition, and the connecting tissues have viscoelastic properties that are represented by groups of torsional Kelvin elements.
4. The rotation of the brain is restricted to the sagittal plane.

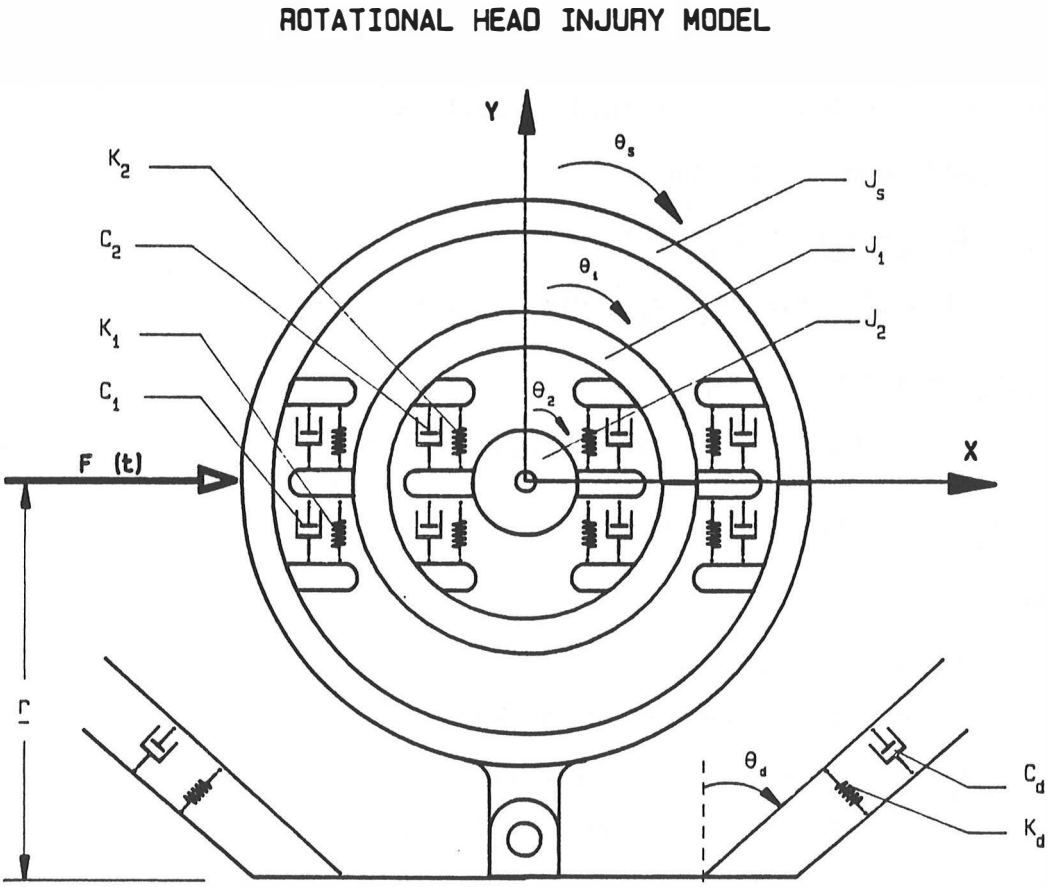

Fig. 1 - The Proposed Rotational Head Injury Model

For the purpose of head injury simulation, the head is considered as a two dimensional, three-mass system as shown in Figure 1. The brain is thought of as a discrete system and is represented by two masses with mass moments of inertia J_1 and J_2, respectively. They are coupled by four torsional Kelvin elements which signify the material properties of the neural tissues. Each Kelvin element consists of a spring and a damper, connecting in parallel between the two masses. The brain is modelled in such a manner that the shearing strain exerted within itself can be detected as it is deformed angularly with respect to the skull due to inertial loading. The brain is coupled to the skull with another group of Kelvin elements that represent the material properties of the dura and other connective tissues, and enable us to determine the shear causing differential displacement between the skull-brain interface.

The head can also contact outside surfaces. The inclined springs and dampers connected to the reference frame represent the material properties and physical characteristics of the contacting object, such as an airbag or a steering wheel. They are expressed in a delayed fashion to emulate various impact conditions. Before the model is in contact with the above mentioned objects, no model responses from the impact will be observed.

To describe the motions of the system, Newton's method was selected; the equations of motion are expressed as follows:

With impact force as loading:

$$T_\iota - K_1(\theta_s - \theta_1) - C_1(\omega_s - \omega_1) - \left[K_d(\theta_s - \theta_d) + C_d \times \omega_s \right] \bullet U(\theta_s - \theta_d) = J_s \times \alpha_s \tag{1}$$

$$K_1(\theta_s - \theta_1) + C_1(\omega_s - \omega_1) + K_2(\theta_2 - \theta_1) + C_2(\omega_2 - \omega_1) = J_1 \times \alpha_1 \tag{2}$$

$$K_2(\theta_1 - \theta_2) + C_2(\omega_1 - \omega_2) = J_2 \times \alpha_2 \tag{3}$$

where

T_ι = $F(t) \times r$, mN
= Torque exerted on the head

K_ι = Spring stiffness, mN/rad

C_ι = Damping coefficient, mN.sec/rad

ι = 1,2

1 = skull-brain connecting tissues
2 = brain/neural tissues

J_ι = mass moment of inertia, $Kg.m^2$

α_ι = Angular acceleration, rad/sec^2

ω_ι = Angular velocity, rad/sec

θ_ι = Anuglar displacement, rad

ι = s, 1 and 2

s = skull
1 = brain - 1st layer
2 = brain - 2nd layer

$U(\theta_s - \theta_d)$ = Unit delayed step function

θ_d = The angular displacement required for the model to be in contact with the objects, rad.

∴ The dynamic responses of the model can be determined from the following differential equations:

$$\theta_s = (1/J_s)\cdot\iint\Big(T_\iota + K_1(\theta_1 - \theta_s) + C_1(\omega_1 - \omega_s) - \Big[K_d(\theta_s - \theta_d) + C_d \times \omega_s\Big]\bullet U(\theta_s - \theta_d)\Big)d\tau^2 \quad (4)$$

$$\theta_1 = (1/J_1)\cdot\iint\Big(K_1(\theta_s - \theta_1) + C_1(\omega_s - \omega_1) + K_2(\theta_2 - \theta_1) + C_2(\omega_2 - \omega_1)\Big)d\tau^2 \quad (5)$$

$$\theta_2 = (1/J_2)\cdot\iint\Big(K_2(\theta_1 - \theta_2) + C_2(\omega_1 - \omega_2)\Big)d\tau^2 \quad (6)$$

If a well distributed angular acceleration input is given to the model as initial loading, the equations of motion reduce as follow:

$$K_1(\theta_s - \theta_1) + C_1(\omega_s - \omega_1) + K_2(\theta_2 - \theta_1) + C_2(\omega_2 - \omega_1) = J_1 \times \alpha_1 \quad (7)$$

$$K_2(\theta_1 - \theta_2) + C_2(\omega_1 - \omega_2) = J_2 \times \alpha_2 \quad (8)$$

i.e.,

given α_s,

$$\omega_s = \int \alpha_s \, d\tau$$

and

$$\theta_s = \iint \alpha_s \, d\tau^2$$

Differential equations identical to equations (5) and (6) can be solved to determine the dynamic responses of the model under such loading.

The analysis of the proposed head injury model was divided into two phases. In phase one, the main objective was to establish the empirical values for the parameters used in the lumped system. In the second phase, the model was exercised under various impact conditions. The responses obtained from these simulations will be correlated with the appropriate injury criterion in future studies, thus elucidating the tolerance and threshold for rotational head injury.

PHASE I - VALIDATION OF THE INJURY MODEL

Most of the properties related to the model such as mass moments of inertia for the human head have been directly measured. Due to the wide variation in the biological properties and observed responses between individual human heads, it is difficult to ascertain the statistical significant of the measured properties. To contend with this problem, average values of the mentioned properties have been obtained. However, such properties only partially describe the dynamic behavior of the head; additional information on the viscoelastic properties of the connective and neural tissues are needed. Much information on such tissues is contained in the form of load versus deflection curves under quasistatic or dynamic loading, relaxation and creep tests and other similar tests. As the Kelvin elements are utilized to represent the physical properties of the neural and connective tissues, the constants that are assigned to the spring and damper of the Kelvin element can be determined from such data. However, such values only serve as initial estimates. The model was subjected to inertial loadings with angular acceleration traces similar to those observed in experiments conducted by Thibault, et al. [4]. The empirical constants for the Kelvin elements were established via iterations till the generated Lagrangian strains between the skull-brain interface, and the brain itself matches well with the experimental data.

Figures 2 to 5 are the typical responses of the head model under inertial loading with the brain mass moment of inertia kept constant at 2.78×10^{-4} $Kg.m^2$ (in Fig. 5, RD1 = $|\theta_s - \theta_1|$ and RD2 = $|\theta_1 - \theta_2|$, and are the relative displacements between the skull-brain interface and the deformations within the brain, respectively).

After several iterations, it was found that the responses of the proposed model agreed well with the experimental data when the Kelvin elements had stiffnesses of 0.32 and 0.25 for K_1 and K_2, and damping coefficients of 0.11 for C_1 and C_2. This correlation is shown in Figure 6. In addition, 37.5% of the total brain mass moment of inertia is occupied by J_1, and the remainder is shared by J_2 [5].

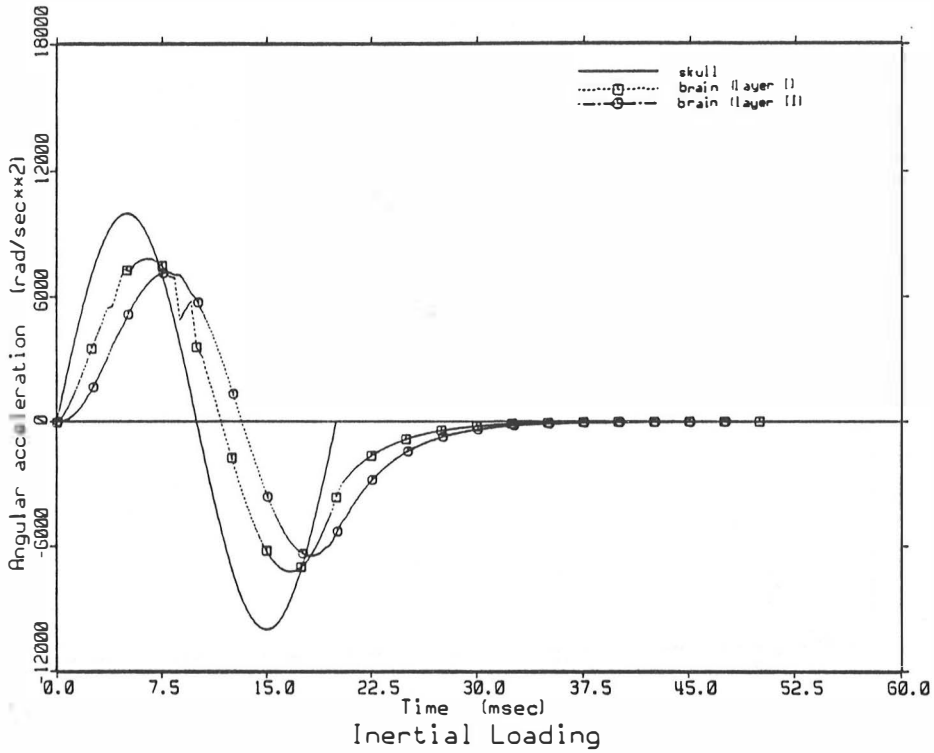

Fig. 2 - Angular Acceleration
(Inertial Loading)

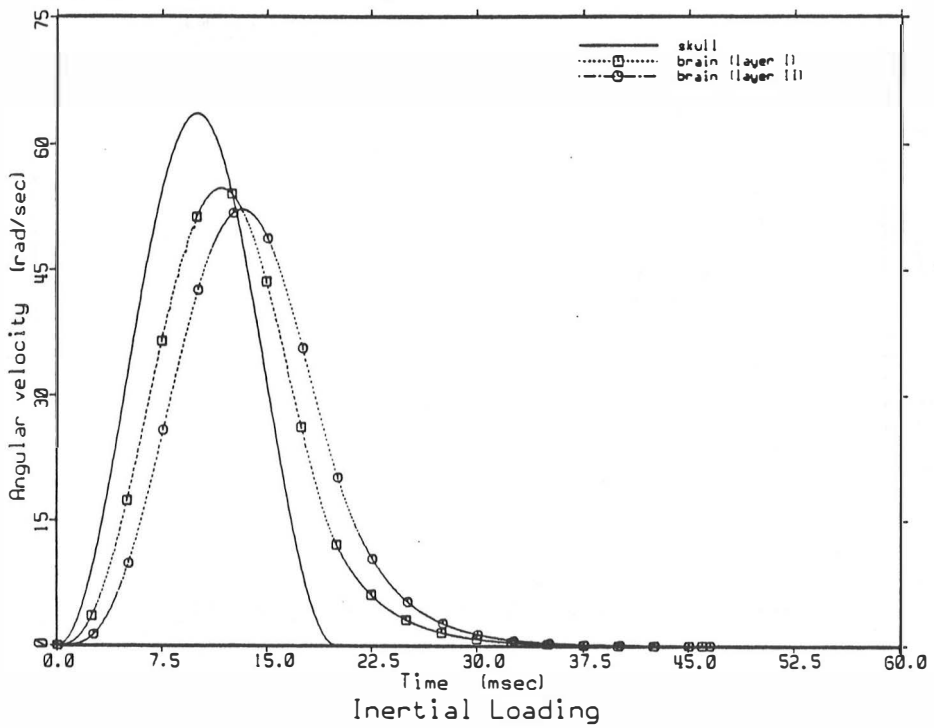

Fig. 3 - Resultant Angular Velocity

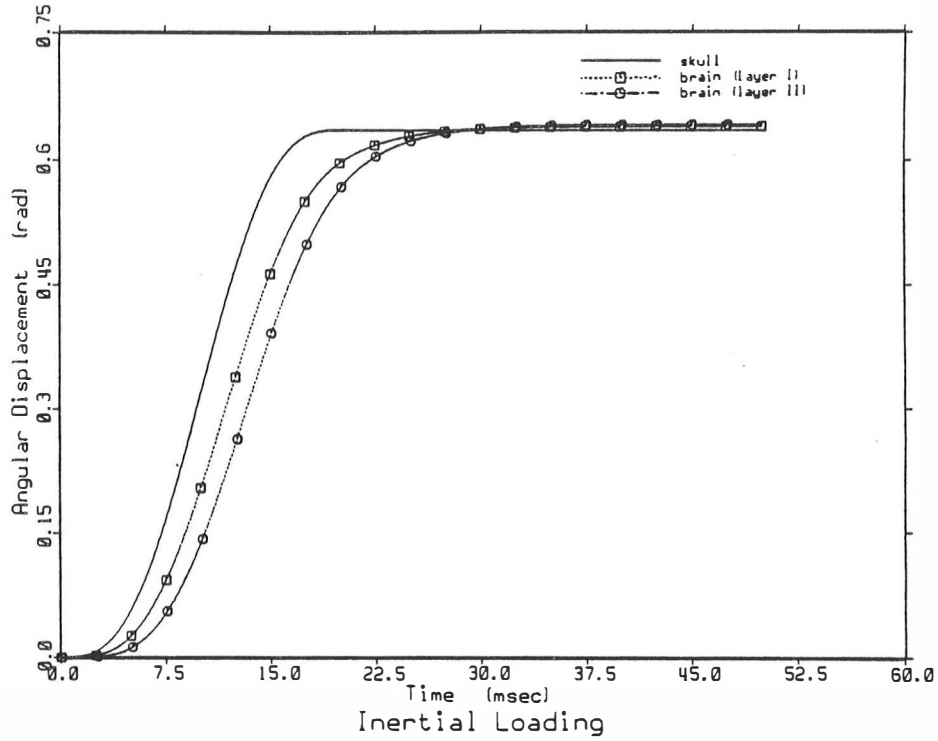

Fig. 4 - Resultant Angular displacement

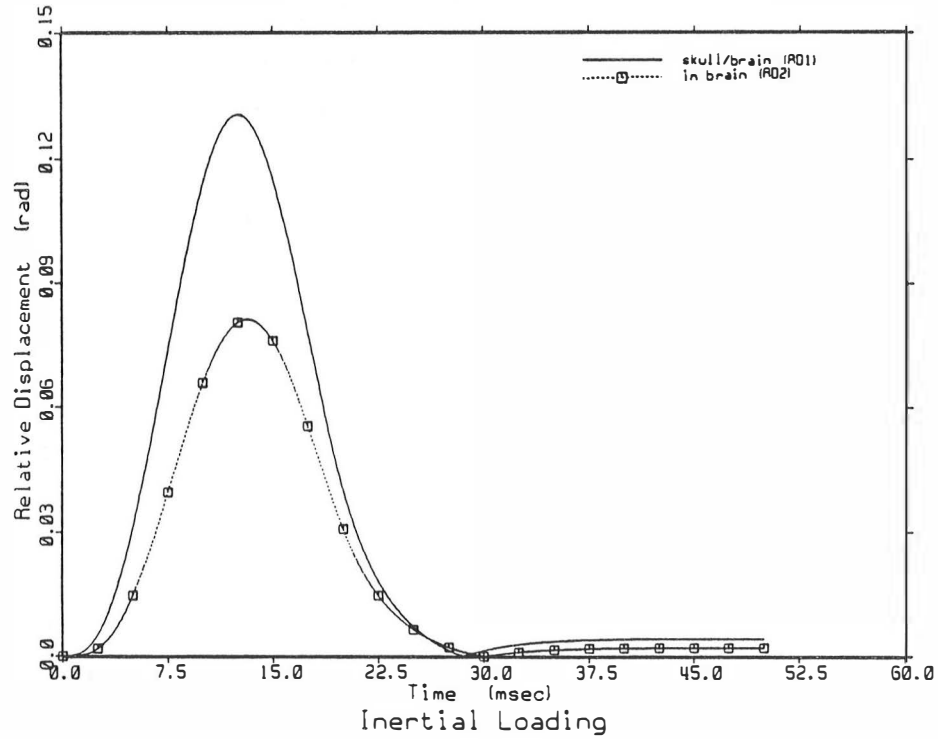

Fig. 5 - Relative Displacement

PHASE II - APPLICATIONS

The model developed can be used for various dynamic analyses for many car crash and other impact simulations. Two examples will be presented in this paper to illustrate the versatility of the head model.

Response of the model - with helmet attached:

Here, the primary objective was to determine the model response to impulsive loading with a helmet attached. These exercises were conducted with the assumption that the helmet is rigidly affixed to the head. The helmet size was varied but the mass was held constant at 0.9 Kg. For these simulations, the J_s term in equations (1) and (4) will include the mass moment of inertia for the helmet. Figure 7 illustrates the sensitivity of the model response (in Lagrangian strains) to various helmet size.

The Lagrangian strains were computed from the peak relative displacement traces similar to those in Figure 6. Figure 7 inferred that for a given impact condition, the severity and possibility of rotational head injury can be reduced with the increased of helmet size.

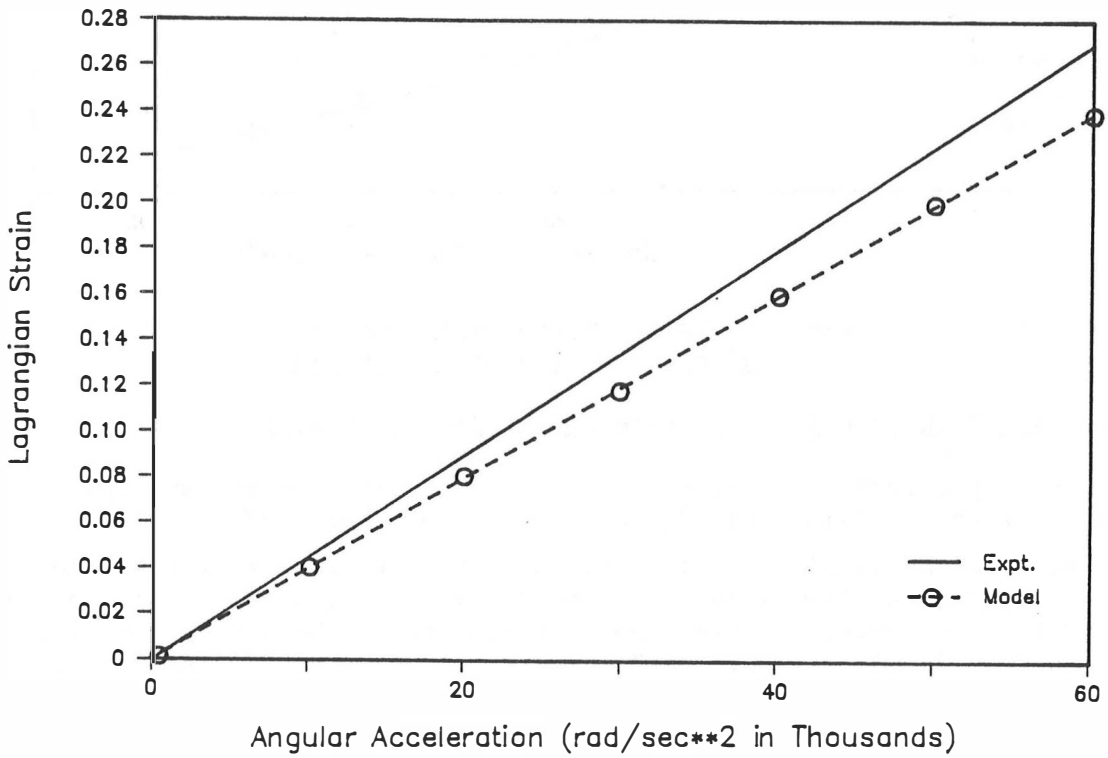

Fig. 6 - Lagrangian Strain vs Angular Acceleration
(Injury model and experimental data)

This is reflected in the reduction of Lagrangian strain between the skull-brain interface (R1) and in the brain itself (R2). Such strain is proportional to the amount of stretch in the neural and vascular tissues. Caution should be exercised that for a given mass, increase in helmet size does reduce the possibility of rotational head injury, however, such increment in size will inevitably reduce the helmet wall thickness, making it more vulnerable to blunt impact, increasing the possibility of other forms of injuries.

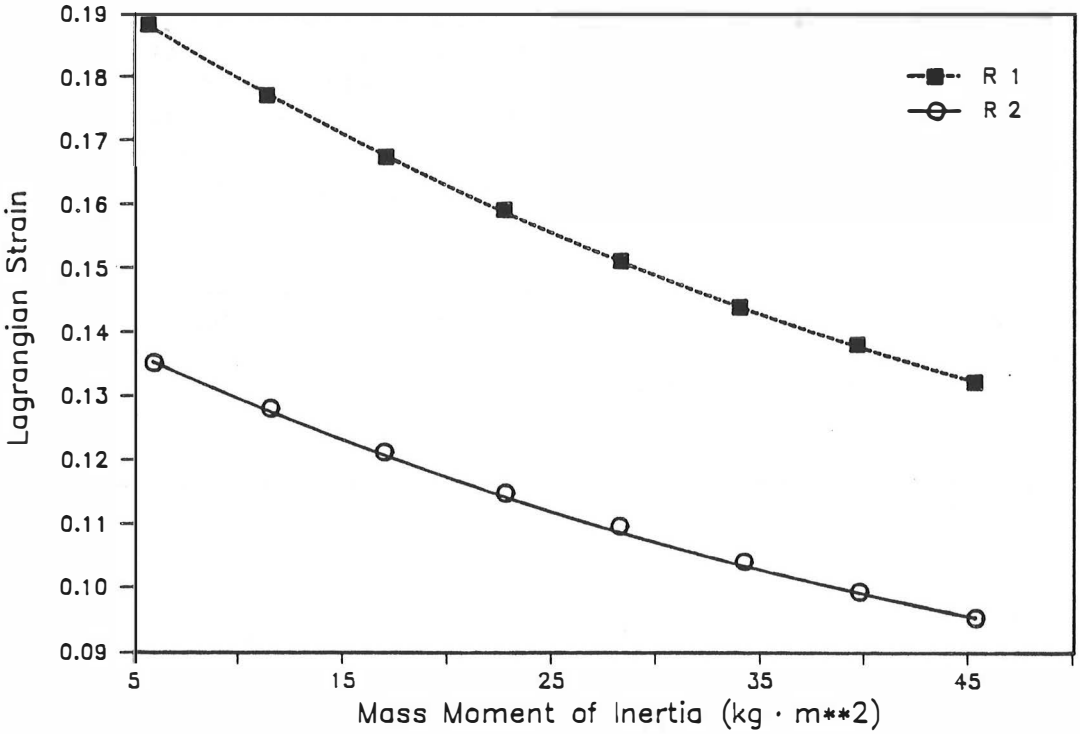

Fig. 7 - The response of the model with increasing size of helmet

Response of the model to a supplementary Restraint System - Airbag

In this parametric study, the airbag was modelled as two springs and a damper with a delay of θ_d which is the placement of the airbag with respect to the model initial position. The mathematical formulation of this simulation is similar to those derived in equations (1) to (3). However, the delayed terms now represent the mechanical properties of the airbag (K_d is replaced by K_{a1} and K_{a2}, and C_d by C_a, respectively). K_{a1} is the stiffness of the airbag which is linearly proportional to the airbag deflection $(\theta_s - \theta_d)$. K_{a2} is the stiffness that is related to $(\theta_s - \theta_d)^3$ and characterizes a pneumatic spring. Both the spring

constant and damping coefficient can be estimated from pneumatic spring and damper approach, with operating pressure varying between 0 - 350 KPa.

The main objective of this study was to examine how the stiffness and damping coefficient of the airbag have affect on the response of the model.

Figure 8 illustrates the general response of the model in terms of relative displacement when impacted into an airbag. The peak displacements were used to compute the Lagrangian strains exerted in the head . As indicated, higher strain values will increase the possibility of severe head injury. Figure 9 indicates that for a given impact condition, increasing the airbag stiffness will reduce the amount of strain seen in the model and thus lower the possibility of severe head injury. A similar phenomenon was observed with the increase in airbag damping coefficient, as illustrated in Figure 10. It is interesting to note that the improvement on head injury prevention is in the form of exponential decay with increase in airbag stiffness and the damping coefficient. This indicates that there may exist an ultimate value of how stiff and how energy absorbing the airbag should be.

CLOSURE

The values obtained from the above parametric studies should not be coded directly for any designing purposes. They are only illustrations

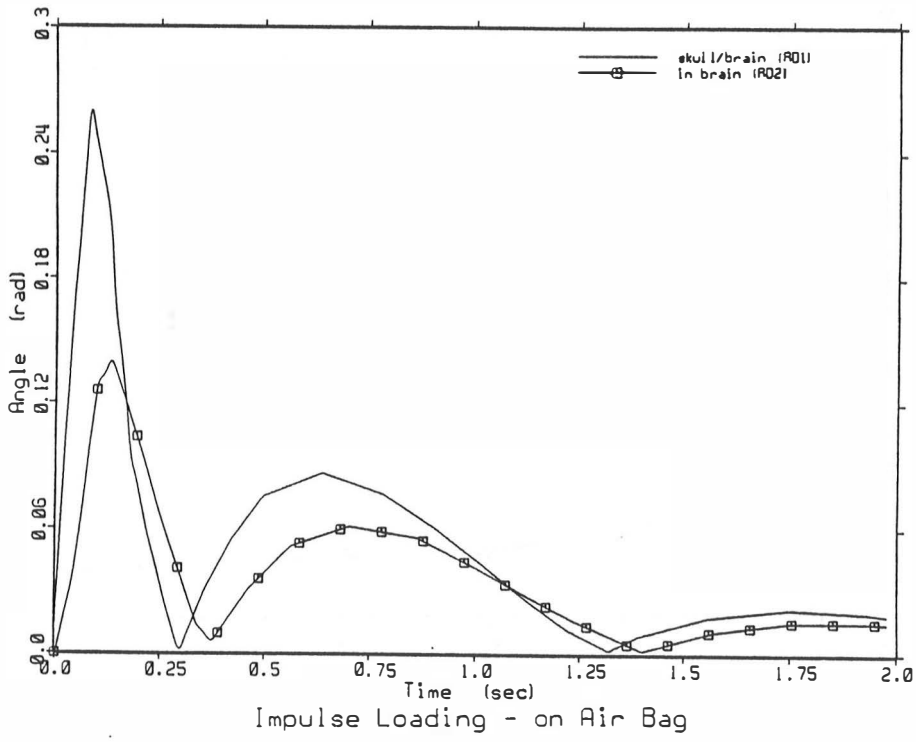

Fig. 8 - Relative Displacement

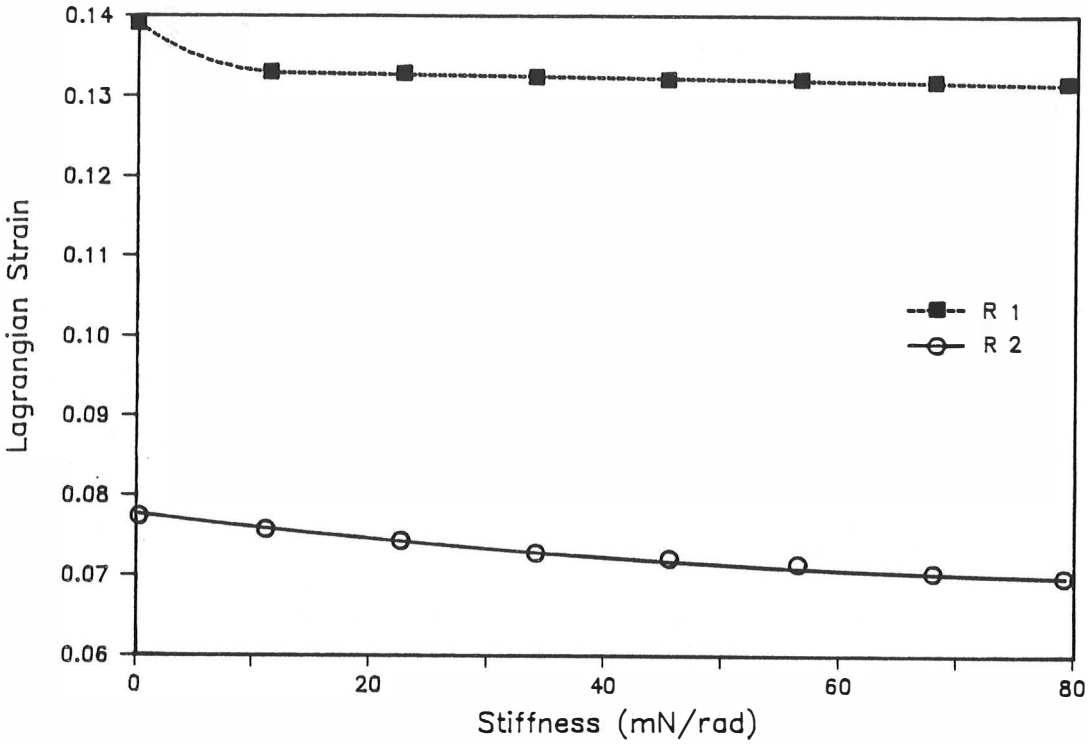

Fig. 9 - The response of the model to various airbag stiffness

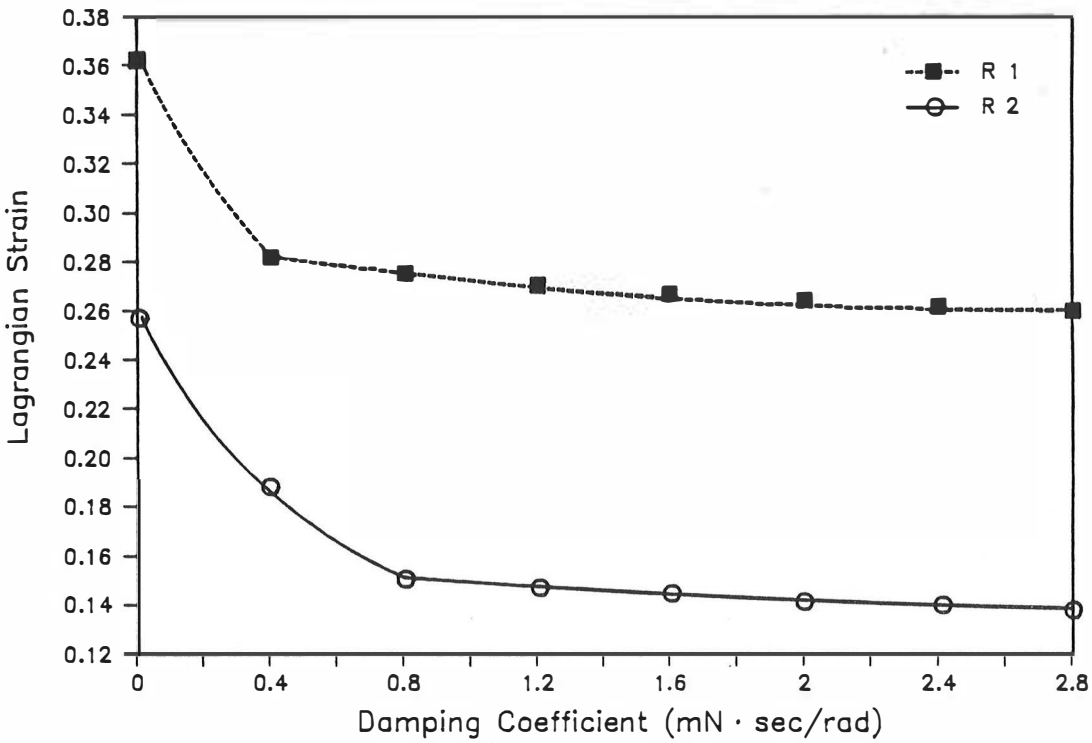

Fig. 10 - The response of the model to various airbag damping coefficients

on how the model can be applied to various impact simulations. It is evident that the developed injury model is very flexible in performing various dynamic analyses. By varying the stiffness and damping coefficients of the delayed elements (K_d and C_d) many different impact conditions can be simulated (such as impact on a steering wheel or a windshield). If the impact is given in the form of angular acceleration (as most experimental data are presented), such acceleration traces can be applied to the model directly without prior knowledge of the impacting surface or any other external protective devices that are attached to the model.

To successfully utilize the proposed injury model as a valid experimental tool in determining the injury mechanism and criteria, more extensive parametric studies should be conducted to account for the nonlinear properties of the neural and skull-brain connective tissues. A tolerance curve for evaluating the injury severity of a given impact can be established, so as to correlate the computed Lagrangian strains with the known AIS scale.

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