The model consists of alternating rigid and deformable bodies simulating the vertebrae and th discs respectively and accounts for the natural curvature of the spine. The rigid-elements are allowed three degrees of freedom in the sagital plane of the body, and the mass of the system is assumed to be concentrated in them. Engineering beam analysis is employed to characterise the disc force-deformation behaviour.

The dependent variables are the axial force, shear and bending moment resultants at the interfaces of the discs and the vertebrae as well as the displacement of the vertebral bodies; time is the independent variable. The base of the spine is constrained against rotation and the pulse is applied through the pelvis and the sacrum. The following further assumptions are incorporated in the model; (i) all inertial resistance to motion is attributed to rigid elements, having geometry corresponding to that of individual vertebrae and capable of executing three rigid body motions (two translational and one rotational, in the sagital plane) (ii) deformations are sustained only by the discs, which respond to axial, shear and bending deformations in accordance with specified consistutive equations, (iii) the head is simulated by a rigid link with appropriate values of translational and rotational inertial resistance and (iv) the stresses and strains in the spine under its own static (1g) load are not considered.

In the initial configuration of the system (prior to loading), the axis of any disc is coincident with the axis of the vertebrae, immediately below. Also, each disc is assumed to be of uniform thickness. A translational acceleration pulse is applied at the base of the spine through the sacrum at an angle to the longitudinal (z) axis. Rotation of the pelvis is assumed to be restrained by its large mass as well as by the presence of lap-type seat belts, thereby simulating a 'fixed beam' and condition at the base of the spine.

The model determines changes in the spinal configuration due to the application of the acceleration and the axial force, shear force and bending moments at the interfaces of disc and vertebrae; these force and moment resultants characterise the nature and the extent of damage to the vertebrae. In order to obtain the solution, the following (digital computerised) procedure is adopted: (1) Following the elapse of some small time, after the pulse is applied, the deformation and the rates of deformation of the discs are computed as function of the instantaneous configuration. (2) When these are known, using the constitutive equations, the axial force, shear force and bending moments acting on the discs are calculated which inturn provide the reaction on the vertebral bodies. (3) The equations of motion for each vertebra are solved by means of Runge-Kutta (or an equivalent algorithm) which enable determination of the new configuration of the system. (4) The procedure is repeated each time with the updated configurations, until the spinal response for a large span of time has been determined.

Thus, due to the dynamic coupling between the axial and bending modes, the problem is nonlinear. In delineating the kinematics of a typical spinal element, the position of points D and A (terminal points of the rigid link, (Fig-1) in the plane are defined by the coordinates: u_i (t), w_i (t), ϕ_i (t) and the link length l_i. At t=0, the axis of the disc (of length g_i) is assumed to be collinear with DAC. At the time t (>0), let the terminal points of the disc be designated by the position A and B. The disc is subject to lateral deformation δ_{1i}, axial elongation δ_{2i}, and rotation δ_{3i}at end sections, relative, to each other. These deformations, are as follows:

$$\delta_{i2} = (\overline{BC})_i \qquad (1)$$

$$\delta_{2i} = (\overline{AC})_i \qquad (2)$$

$$\delta_{3i} = r_i - r_i^0$$

$$= (\phi_i - \phi_{i+1}) - (\phi_i^0 - \phi_i^0{}_{+1}) \qquad (3)$$

wherein (i) the 'sero' superscripts refer to the unstressed state, (ii) the distances $(\overline{BC})_i$ and $(\overline{AC})_i$ are

$$(\overline{BC})_i = (\overline{AB})_i \sin \beta_i \qquad (4)$$

$$(\overline{AC})_i = (\overline{AB})_i \cos \beta_i \qquad (5)$$

The chord length $(\overline{AB})_i$ being given by

$$(\overline{AB})_i = \{ [u_{i+1} - (u_i + l_i \sin \phi_i)]^2 + [w_{i+1} - (w_i + l_i \cos \phi)]^2 \}^{\frac{1}{2}} \qquad (6)$$

(iii) β_i, the angle between the axis of the rigid link l_i and the chord $(\overline{AB})_i$ is given by

$$\beta_i = \propto_i - \phi_i \qquad (7)$$

$$\text{and } \propto_i = \sin^{-1} \left[\frac{u_{i+1} - (u_i + l_i \sin \phi_i)}{(AB)_i} \right] \qquad (8)$$

(iv) r_i^0, the initial angle between adjacent vertebrae in the unstressed state, is

$$r_i^0 = (\phi_i^0 - \phi_i^0{}_{+1}) \qquad (9)$$

and angle r_i, between adjoining vertebrae eat some subsequent time, is,

$$r_i = (\phi_i - \phi_i + {}_1) \qquad (10)$$

so that at time the rotational deformatian between two end plates is $(r_i - r_i^0)$.

To compute the time rate of change of δ_{2i}, for future use in the governing equations of motion, we have

$$\dot{\delta}_{2i} = (\dot{\overline{AC}})_i = (\dot{\overline{AB}})_i \cos \beta_i - \dot{\beta}_i (\overline{AB})_i \sin \beta_i \qquad (11)$$

which, in view of equation (4) can be written as

$$\dot{\delta}_{2i} = (\dot{\overline{AB}})_i \cos \beta_i - \dot{\beta}_i (\overline{BC})_i \qquad (12)$$

where in

$$\dot{\beta}_i = \dot{\propto}_i - \dot{\phi}_i \qquad (13)$$

$$(\dot{\overline{AB}})_i = [(u_{i+1} - u_i - l_i \sin \phi_i)(\dot{u}_{i+1} - \dot{u}_i - l_i \dot{\phi}_i \cos \phi_i) + (w_{i+1} - w_i - l_i \cos \phi_i)(\dot{w}_{i+1} - \dot{w}_i + l_i \dot{\phi}_i \sin \phi_i)]/(\overline{AB})_i \qquad (14)$$

and the derivative $\dot{\propto}$ is obtained, from equation (8) as follows:

$$\dot{\propto} = [(\dot{AB})_i (u_{i+1} - u_i - l_i \phi_1 \cos \phi_i) - (\overline{AB})_i (u_{i+1} - u_i - l_i \sin \phi_i)]/(\overline{AB})_i^2 \cos \propto_i \qquad (15)$$

We note that although in functional notation,

$$\dot{\delta}_{2i} = f(u_i, \dot{u}_i, w_i, \dot{w}_i, \phi_i, \dot{\phi}_i, u_{i+1}, \dot{u}_{i+1}, w_{i+1}, \dot{w}_{i+1}, l_i, g_i)$$

it is calculated sequentially, by stepwise determination of its constinuent terms as prescribed by equations (12) – (15) and (16).

Constitutive equations (force-deformation characteristics) of the discs

The lateral, axial and rotational deformations of the disc are denoted by δ_1, δ_2, δ_3 as shown in Fig 2. F_1, F_2, F_3 denote the shear, axial force and moment, respectively. The lateral displacement δ_1 is comprised of shear deformation δ_s and a bending deformation δ_b. We also assume plane sections remain plane after deformation, and the discs are short, uniform beam segments.

The intervertebral disc behaves like a viscoelastic solid under axial load. Tests have shown that, when subjected to constant axial strain, an exponential relaxation of axial stress occurs with time strain response to a suddenly applied and subsequently maintained axial stress consists of an initial elastic response (strain) with an asymptotic limit. These observations suggest that, a three parameter viscoelastic solid model may be used, at least for phenomenological purposes. Such a model is equivalent to either, a spring in series with a Kelvin-Voigt model, or a spring in parallel with a Maxwell model, either of which can be put in the form

$$\sigma + p_1 \dot{\sigma} = q_0 \epsilon + q_1 \dot{\epsilon} \qquad (16)$$

where p_1, q_0, q_1 represent the three material parameters. If we choose to mechanically represent the system by a spring in series with a Maxwell model, then the following relations exist between these parameters and the mechanical model constants E_1 (parallel elasticity) E_2 (series elasticity) and μ, the viscosity coefficient:

$$\left.\begin{aligned} q_0 &= E_1 E_2 / (E_1 + E_2) \\ E_2 &= q_1 / p_1 \\ p_1 &= 1 / \mu(E_1 + E_2) \\ q_1 &= E_2 / \mu(E_1 + E_2) \end{aligned}\right\} \qquad (17)$$

wherein, q_0 represents the static stiffness of the model and is equivalent to the two springs E_1, and E_2, in series. E_2, on the other hand represents the stress corresponding to a sudden application of a unit strain because the dashphot tends to 'lock-up' at very high strain rates.

When applying equations (16) to a uniform rod of cross-sectional area A and length l, the constitutive equation for axial loading, with force deformation as a basis, can be put in the following form:

$$F_2 + p_1 \dot{F}_2 = A(q_0 \delta_2 + q_1 \dot{\delta}_2)/l \qquad (18)$$

wherein

$$\sigma = F_2 / A = \text{axial stress} \qquad (19)$$

and

$$\epsilon = \delta_2 / l = \text{axial strain} \qquad (20)$$

A shear behaviour is assumed similar to the axial deformation behaviour, defined by equation (18), in the form

$$F_1 + p_1 \dot{F}_1 = (A/Kl)(r_0 \delta_s + r_1 \dot{\delta}_s) \qquad (21)$$

where K is the shape factor associated with the shear.

The constitutive equations for bending are similarly given by

$$F_1 + p_1 \dot{F}_1 = (12 I/l^3)(q_0 \delta_b + q_1 \dot{\delta}_b) + (6 I/l^2)(q_0 \delta_3 + q_1 \dot{\delta}_3) \qquad (22)$$

$$F_3 + p_1 \dot{F}_3 = (6 I/l^2)(q_0 \delta_b + q_1 \dot{\delta}_b) + (4 I/l)(q_0 \delta_3 + q_1 \dot{\delta}_3) \qquad (23)$$

where I represents the area moment of inertia about an axis perpendicular to the sagital plane. We also have

$$\delta_l = \delta_s + \delta_b \qquad (24)$$

which can be substituted in equation (22), to solve for F_1 and substitute in equation (21) to obtain

$$\dot{\delta}_s = \frac{\dfrac{Kl}{A}\left\{F_1 + s_1\left[-F_1 + \dfrac{12 I}{l^3}\left(q_0(\delta_1 - \delta_s) + q_1 \dot{\delta}_1\right) + \dfrac{6 I}{l^2}(q_0 \delta_3 + q_1 \dot{\delta}_3)\right]\right\} - r_0 p_1 \delta_s}{r_1 p_1 + \left(\dfrac{12 I}{l^3}\right)\dfrac{Kl}{A} s_1 q_1} \qquad (25)$$

which gives the rate of change of shear deformation $\dot{\delta}_s$ as a function of the state variables $\delta_1, \delta_s, \delta_3, \dot{\delta}_1, \dot{\delta}_3$ and other known parameters. From equation (21) we now obtain

$$\dot{F}_1 = [-F_1 + (A/Kl)(r_0 \delta_s + r_1 \dot{\delta}_s)]/S_1 \quad (26)$$

where $\dot{\delta}_s$ is obtained from equation (25).

Similarly, from (18) and (24) we obtain,

$$\dot{F}_2 = (1/p_1)[-F_2 + (A/l)(q_0 \delta_2 + q_1 \dot{\delta}_2)] \quad (27)$$

$$\dot{F}_3 = (1/p_1)\{-F_3 + (6I/l^2)[q_0(\delta_1 - \delta_s) + q_1(\dot{\delta}_1 - \dot{\delta}_s)] + (4I/l)(q_0 \delta_3 + q_1 \dot{\delta}_3)\} \quad (28)$$

The constitutive equations (25) to (28) are a system of first order equations, in a manner, readily solvable by numerical solutions such as Runge-Kutta procedure.

At present, however, the material behaviour of the discs in shear and bending is not defined properly, and hence the use of the three perameters s_1, r_0, r_1 to describe the expected viscoelastic behaviour must be considered premature. Also, the complete construction of a disc may lead us to speculate that the nucleus pulposus, because of its central location within the disc, may be providing its viscous damping action mainly in the axial mode since in the bending and shear it is comparatively undisturbed. Incorporating this conjecture, it is reasonable to assume that the discs are viscoelastic in axial compression and, elastic in shear and bending. Thus, by putting $p_1 = q_1 = 0$ and $q_0 = E$, equations (25) to (28) become respectively

$$F_1 = (Ar_0/Kl)\,\delta_s \quad (29)$$

$$F_2 = (12\,I\,q_0/l^3)(\delta_1 - \delta_s) + (6\,Iq_0/l^2)\,\delta_3 \quad (30)$$

$$F_3 = (6\,Iq_0/l^2)(\delta_1 - \delta_s) + (4\,Iq_0/l)\,\delta_3 \quad (31)$$

$$F_2 + p_1 \dot{F}_2 = (A/l)(q_0 \delta_2 + q_1 \dot{\delta}_2) \quad (32)$$

On combining equations (29) and (30) we get

$$F_1 = [(12 q_0 I/l^3)\,\delta_1 + (6\,q_0\,I/l^2)]/[1 + (12 q_0 I/l^3)(Kl/r_0 A)] \quad (33)$$

and on combining equations (29) and (31) we get

$$F_3 = (6\,Iq_0/l^2)[\delta_1 - F_1(Kl/Ar_0)] + (4\,Iq_0/l)\,\delta_3 \quad (34)$$

From equation (32) we obtain

$$\dot{F}_2 = [-F_2 + \frac{A}{l}(q_0 \delta_2 + q_1 \dot{\delta}_2)]/p_1 \quad (35)$$

Substitution of (33) in (34) and simplification yield,

$$F_1 = \frac{12\,I\,q_0}{l^3(4\xi - 3)}\delta_1 + \frac{6\,I\,q_0}{l^2(4\xi - 3)}\delta_3 \quad (36)$$

$$F_3 = \frac{6\,I\,q_0}{l^2(4\xi - 3)}\delta_1 + \frac{4\xi\,I\,q_0}{l(4\xi - 3)}\delta_3 \quad (37)$$

wherein

$$\xi = 1 + \frac{3\,I\,q_0}{l^3}\left(\frac{Kl}{r_0 A}\right) \quad (38)$$

Now, the shear force F_1 and bending moment F_3, given by equations (36) and (37) respectively, are explicit expressions in terms of the deformations δ_1 and δ_3. Equation (35) represents the axial behaviour, and it can be integrated numerically.

Reactions of the disc on the vertebral bodies

The force components acting on the i^{th} disc (the i subscript has been omitted, however, for clarity) shown in figure 2, constitute the reactions on the rigid body. Before delineating these forces, it is useful to note that the consideration of the moment equilibrium of the disc indicates the moment acting on the lower (left) end as

$$B_i = F_{3i} - F_{1i}(AC)_i \quad (39)$$

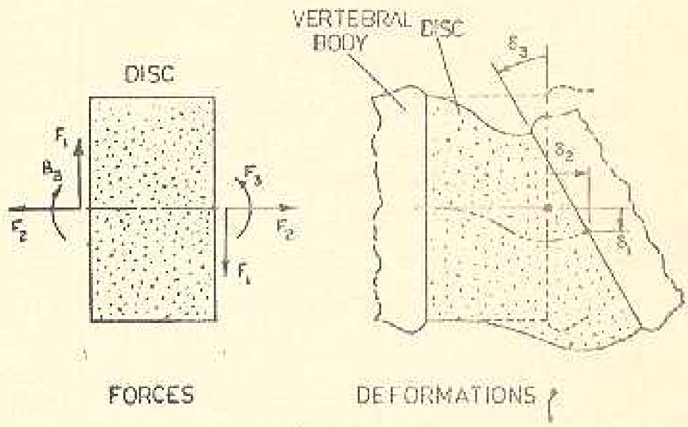

Fig. 2.

Now, turning to the reaction forces on the vertebral body, the transformation of the shear and the axial reaction forces to components acting in the X and Y direction (fig. 3) provides

$$\left.\begin{aligned} Q_i &= F_{1i}\cos\phi_i + F_{2i}\sin\phi_i \\ P_i &= F_{1i}\sin\phi_i + F_{2i}\cos\phi_i \end{aligned}\right\} \quad (40)$$

where ϕ_i denotes the instantaneous orientation of the vertebral body, shown in Figure 3. The forces acting on the vartebral bodies are also delineated in Figure 3 wherein (i) the mass moment of inertia, about an axis normal to the sagital plane and through the centre of mass, is denoted by J_i and (ii) the eccentricity of the centre of mass with respect to the axis of the vertebral body is e.

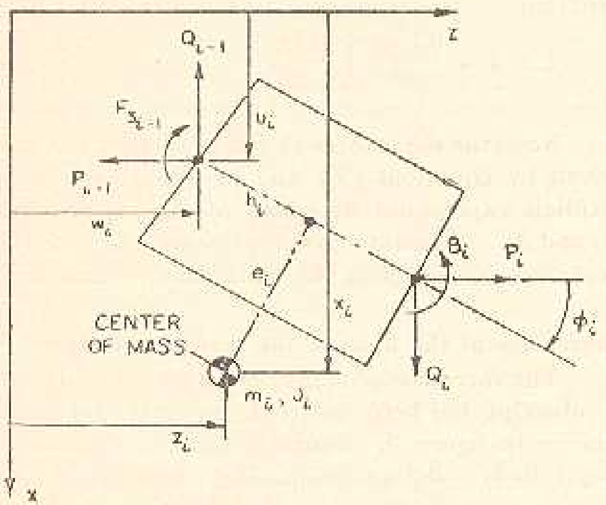

Fig. 3.

Equation of motion of the vertebral body

If x_i and z_i denote the coordinates of the mass centres, then from figure 3, we have,

$$x_i = u_i + h_i \sin\phi_i + e_i \cos\phi_i \quad (41)$$

$$z_i = w_i + h_i \cos\phi_i - e_i \sin\phi_i \quad (42)$$

wherein u_i, w_i and ϕ_i are the Lagrangian coordinates.

Differentiation of equations (41) and (42) twice with respect to time gives the acceleration of the mass centre (with respect to u_i, w_i and ϕ_i) as follows :

$$\ddot{x}_i = \ddot{u}_i + C_{1i}\dot{\phi}_i^2 - C_{2i}\ddot{\phi}_i \quad (43)$$

$$\ddot{z}_i = \ddot{w}_i + C_{2i}\dot{\phi}_i^2 + C_{1i}\ddot{\phi}_i \quad (44)$$

where

$$C_{1i} = -(h_i \sin\phi_i + e_i \cos\phi_i) \quad (45)$$

$$C_{2i} = -(h_i \cos\phi_i - e_i \sin\phi_i) \quad (46)$$

The equations of translatory motion of the vertebrae can be now written as

$$m_i \ddot{x}_i = Q_i - Q_{i-1} \quad (47)$$

and

$$m_i \ddot{z}_i = P_i - P_{i-1} \quad (48)$$

Substitution of equations (43) and (44) into the above equations give

$$\ddot{u}_i = [(Q_i - Q_{i-1})/m_i] - C_{1i}\dot{\phi}_i^2 + C_{2i}\ddot{\phi}_i \quad (49)$$

and

$$\ddot{w}_i = [(P_i - P_{i-1})/m_i] - C_{2i}\dot{\phi}_i^2 - C_{1i}\ddot{\phi}_i \quad (50)$$

Taking moments about the mass centre, we obtain the equation of rotational motion as

$$J_i \ddot{\phi}_i = (F_{3i-1} - B_i) - C_{2i} Q_{i-1} + C_{3i} Q_i + C_{1i} P_{i-1} - C_{4i} P_i \quad (51)$$

wherein

$$C_{3i} = h_i \cos\phi_i + e_i \sin\phi_i \quad (52)$$

and

$$C_{4i} = h_i \sin\phi_i - e_i \cos\phi_i \quad (53)$$

The terms Q_i, P_i, B_i etc. in equations (49) - (51) are the force and moment reactions of the discs on the vertebra, Herein, ϕ_i is first obtained from equation (51) and then substituted into equations (49) and (50) to give $\ddot{u}_i$ and $\ddot{w}_i$ respectively.

For the boundary conditions, we treat the system as a continuum. Then at the terminal end, the head, we have force boundary conditions

$$B_i = P_i = Q_i = O \quad (54)$$

where i refers to the head mass in equations (49) – (51). At the base of the spinal column, displacement boundary conditions are imposed.

As shown in Figure 1, the sacrum supplies an acceleration pulse, (F(t) at some angle η to the Z — axis. The pelvic angle is hence assumed to be fixed and the (z and x) components of the acceleration pulse are given by

$$\ddot{z}(t) = \cos\eta\, F(t) \quad (55)$$

$$\ddot{x}(t) = \sin\eta\, F(t) \quad (56)$$

This completes the mathematical description of this discrete parameter model of the spinal column.

Solution and Results

With twenty-five masses in the system each having three degrees of freedom in the plane, numerical integration is used to solve the governing equations (35)—(37) and (49)—(51). In solving them, the second order equations (49)—(51) are converted to six first order equations.

The results reveal that (Orne and Liu[7].)

(1) In the pilot ejection problem, a dominant factor is the bending and shear deformations, in the gross response of the body. Hence, any injury criterion should be based on combined axial, shear and bending forces.

(2) The peak response of axial force, shear and bending moment remains invariant when the applied acceleration pulse rise time is varied between 5 and 14 msec.

(3) The high incidence of vertebral fractures in the thoracic region may be explained from the fact that the bending moments and axial compressive forces produce compressive stresses which are additive at the anterior portion of the thoracic vertebrae.

(4) The shear and moment response in the lumbar region is appreciably increased by the horizontal component of acceleration for a pulse angle of 10°. Simultaneously, this component reduces the axial response along the vertebral column.

In order to obtain more useful information on spinal injury mechanisms and states it is relevant to get and incorporate in the model further experimental information on. (i) the variation of rotary inertia and mass eccentricity along the spinal column (towards this end, Liu et al[5,6] have been working) (ii) the variation of the behaviour of the vertebral and disc materials along the spine when subjected to axial, shear and bending loads, (iii) the failure mechanism of vertebrae under combined states of loading and (iv) mathematical description af possible pilot postures at ejection.

Eventually we would want to characterise and study the response of the system as a continuum. The discrete-parameter models, when incorporated with more details on geometry variations and inhomogenieties within an element and from element to element, may approach continuum models. The current continuum models of spine are rather simplistic to help predict the locations of and circumstances under which fractures occur.

Continuum model for spinal injury

Let us consider a uniform, viscoelastic, one-dimensional continuum model, studied by Terry and Roberts and determine its suitability by comparing the accelerations attained at certain points with experimentally applied accelerations at the corresponding points of the cadaver spine. The material considered is that of a Maxwell type viscoelastic medium.

A ramp input acceleration pulse is applied to the theoretical rod (simulating the spinal column) at one end to conform to the acceleration input. The analysis concerning the determination of the resulting acceleration at the far end of the rod (which corresponds with the head acceleration) consists of using the governing equation of motion in the form of a one-dimensional wave in a uniform rod ; this equation is

$$\sigma_{XX} = \rho U_{tt} \qquad (57)$$

wherein σ and U are longitudinal stress and displacement components and the subscripts x and t denote diferentiation with respect to these variables.

We know, by definition, that

$$\varepsilon = U_x \qquad (58)$$

The constitutive equation is

$$\varepsilon_t = (1/E)\,\sigma_t + \mu\sigma \qquad (59)$$

wherein E is the Young's modulus of the rod material and μ^{-1} is its coefficient of viscosity.

Eliminating ε between equations (58) and (59) and then differentiating the resulting expression with respect to t and rearrangement give

$$a_t = (1/\rho)\,S_x \qquad (60)$$

$$a_x = (1/E)\,S_t + \mu S \qquad (61)$$

wherein S is the stress rate and a is the instantaneous acceleration.

Equation (60 and (61) are partial differential equations of the hyperbolic type and are solved by the method of characteristics using a computer. The chosen boundary conditions are

$$a(o,t) = (A/Y)\,t,\ t \leqslant Y,\ x = O$$

$$a(o,t) = A,\ t \geqslant Y,\ x = O \qquad (62)$$

$$S(L,t) = O,\ x = L$$

wherein A is the plateau acceleration of a ramp input, Y = the rise time for ramp input and L is the length of the viscoelastic rod.

To obtain the optimal model, the values for E and $(1/\mu)$ are obtained by matching the analytically obtained acceleration time curves with the corresponding experimental curves for different onset rate and acceleration level characteristics. Whereas the optimal value of E is 2.5×10^4 lb/ft², the value of μ (in ft²/lb-sec.) is given by exp (− 4.6 + 0.00827) (onset rate).

Observations

The following observations are made from the model results. (i) For low input acceleration levels upto say 4.5 g, the values of the theoretical acceleration-time curves corresponded with the experimental results.

(ii) the error (difference between the theoretical and experimental curves) increases with the increase in the acceleration level, (iii) the evaluated response frequency of the theoretical model is a constant, whereas a similar observation cannot be made in regard to the experimental curves.

Some improvements on this model, as suggested by Terry and Roberts[9] entail modification of the theory to employ a viscous element in which the viscosity is a function of the strain rate. This would permit the experimentally obtained strain rate properties of the bone to be incorporated in the model and provide a nonlinear response model of the spine. Of course, any changes in the basic theory would also, affect the method of solution and as the sophistication of the theory increases, so do the complications in the solution. The best practically utilitarian model would, however, be the one which combines maximum accuracy with maximum simplicity.

REFERENCES :

1. EVANS, G.F. and LISSENER, H.R. Biomechanical studies on the lumbar spine and pelvis: J of Bone and Joint Surg. 41-A: 2. 278-290, 1959.

2. EVANS, G.F., LISSNER, H.R. and PATRICK, L.M. — Accelration-induced strains in the intact ver tebral column, J. App. Physiology 17, 405-409, 1962.

3. HIRSCH, C — The reaction of intervertebral discs to compression forces,: J of Bone and Joint Surg. 37-A : 6, 118-1196, 1968.

4. KING, A.I., and VULCAN, A.P. Elastic deformation characteristics of the Spine: J. Biomechanics, 4, 413-429, 1971.

5. LIU, Y.K., LABORDE, M.J. and VAN BUSKIRK, W.C. Inertial properties of a segmented cadaver trunk: their implications in acceleration injuries. Aerospace Med, 42 (6), 650-657, 1971.

6. LIU, Y.K., WICKSTROM, J.K. and CHAN, H.S. Estimation of the inertial property distribution of the human torso from segmented cadaver data: Research report. Biomechanics Laboratory, Tulane University School of Medicine and Engineering, New Orleans, 1972.

7. ORNE, D, and LIU, Y.K. A mathematical model of spinal response to impact, Dept of Engineering Mechanics, the University of Michigan, July 1969.

8. PATRICK, L.M., LISSNER? H.R. and EVANS, G.F. Effects of controlled acceleration on stress-strain phenomena in the intact vertebral column. Societe' International de Chirurgie Orthopedique et de Traumstologie, Buitieme Congres, New York, 71.

9. TERRY, C.T. and ROBERTS, V.L. A viscoelastic model of the human spine subjected to +gz accelerations. J. Biomechanics 1, 161-168, 1968.