THE DEVELOPMENT OF A PHYSICAL MODEL TO MEASURE STRAIN IN A SURROGATE SPINAL CORD DURING HYPERFLEXION AND HYPEREXTENSION

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ABSTRACT

There have been many attempts to characterize the motion of the head and neck under direct head impact or external acceleration loading. However, little information is available which describes how these motions result in deformation of the spinal cord within the spinal canal and, in turn, damage to the spinal cord.

In this study, we examined the mechanical characteristics of the head and neck structure, with special emphasis on the spinal cord, in order to gain insight into the deformation of the spinal cord during traumatic spinal cord injury. Mechanical tests of fresh human cadaver cervical and thoracic spinal cords and cervical dura mater were conducted in order to characterize the time dependent properties of these tissues. Once characterized, the results from these tests were used to evaluate appropriate materials for a surrogate spinal cord and brain. Materials deemed as suitable surrogates were used for the construction of an articulating model of the head and cervical spine. Quasistatic flexion tests of the completed physical model indicated reasonable agreement between vertebral motion and available information on the cervical spine kinematics. Additionally, motion of the spinal cord surrogate was similar to previous cadaver and human volunteer studies.

With more validation, this model may be used to measure the strains in a surrogate spinal cord during dynamic hyperflexion and hyperextension experiments. The results from these experiments, together with isolated tissue experiments conducted in parallel in our laboratory, will allow us to correlate the stretch parameters measured and the functional response of neural and vascular tissue. Relating the functional failure criteria for isolated central nervous system tissue to estimates of the spinal cord deformation, during appropriate loading conditions, will allow estimation of the spinal cord lesions expected from hyperextension, hyperflexion and subluxation of the cervical spinal column.

BACKGROUND

Traumatic spinal cord injury is a significant medical problem throughout the developed world. Although these injuries make up less than 1 percent of the hospitalized injuries every year, each injury carries a very high cost in hospitalization, therapy, rehabilitation, and lost earnings [22]. To date, the exact nature of the insult and the resulting cellular processes are not fully understood, although some therapeutic measures have been shown to somewhat ameliorate the permanent neurological damage from a spinal cord injury. In particular, the mechanical events which give rise to functional or structural failure of the spinal cord are not adequately described in the literature. Currently, some studies are available that describe the in situ spinal cord motion during full flexion and full extension for human subjects, [5,15] but none have reported the dynamic motion of the spinal cord in conditions such as hyperflexion, hyperextension, and subluxation. Moreover, no data is available describing the deformation of the spinal cord occurring during injurious loading, and the manner in which these

deformations relate to functional impairment of nervous tissue measured in vitro [9,23].

In response to these needs, this study was conducted to develop a system to quantify the deformation of the spinal cord due to common modes of injury. The mechanical behaviour of the spinal cord and dura are tested using human cadaveric material, and suitable surrogate materials for these structures are identified and characterized. Lastly, an anatomically accurate physical model of the skull-brain-cervical spine structure is constructed using these surrogate materials and evaluated for its biofidelity and consistency with earlier investigations [16-19, 20-21, 24, 27].

METHODS

1. Mechanical Testing

(a) Human Cadaver Cervical and Thoracic Spinal Cords

Cervical and thoracic specimens of human spinal cord were obtained at autopsy from adults aged 30-84, under the Hospital of the University of Pennsylvania Anatomical Gifts program[1]. Candidate donors were screened, and rejected if there was any history of brain or spinal cord trauma, or if more than 24 hours had elapsed since death. The samples were excised gently in order to avoid damage prior to testing. The dura mater was also harvested at this time. Samples were placed immediately on a bed of saturated gauze, sprayed lightly with Ringer's solution, and sealed in a container, maintaining a 100 percent humidity environment to avoid degradation of mechanical properties caused by dehydration of the tissue.

In preparation for testing, the ends of each sample were mounted between two plastic plates and fixed in position with cyanoacrylate adhesive, as shown in Figure 1. During this procedure, and throughout testing, the sample was constantly irrigated with Ringer's solution in order to keep it moist. The dimensions and gauge length of each sample were recorded.

The samples were then transferred to a materials testing machine (Instron Corp, model 1011) attached to a personal computer (IBM Corp, PC-AT). The computer performs on-line data acquisition and software allows the maximum extension, extension rate and type of test to be carefully controlled. Relaxation tests were conducted using a variety of loading rates (0.02-0.3s^{-1}) and maximum strains (10-20%). To investigate time-dependent properties of the cord, force and displacement data was recorded by the personal computer, while dimensions and gauge length of each sample were recorded manually. From these data, stress relaxation curves and stress-strain curves were constructed for each sample. Additionally, average tangent moduli were calculated for that portion of each stress-strain curve above a threshold stress of 0.025MPa.

(b) Human Cervical Dura Mater

Dura mater samples were obtained at autopsy from adult human cadavers, using procedures similar to those described above for the spinal cord samples. The samples were sprayed with Ringer's solution to maintain hydration and then immediately stored in a sealed container on a bed of saturated gauze.

The samples were then dissected into small strips approximately 40mm in length and 7mm in

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1. The rationale and use of human cadaveric material has been reviewed and approved by the Institutional Review Board. All testing procedures and data analysis have been conducted under the guidelines of the Declaration of Helsinki.

width, with the long axis of the specimen aligned with the primary fibre direction. Samples were then mounted between two thin plastic plates, using cyanoacrylate adhesive to ensure a secure grip, and held in pneumatic grips on a materials testing machine (Instron Corp., model 1011). Specimens were subjected to a ramp displacement test, at either 100mm/min ($\dot{\varepsilon} \approx$ 0.03-0.05s^{-1}) or 500mm/min ($\dot{\varepsilon} \approx$ 0.36-0.5s^{-1}) to a maximum extension of 5mm ($\varepsilon \approx$ 0.1-0.2), while extension and force data were acquired by the computer.

(c) Surrogate Materials

The test samples of candidate surrogate materials were prepared from Sylgard silicone gel material (Dow Corning, Inc., MI). The primary motivation for this choice of material was that the mechanical properties may be varied over a wide range by altering the relative proportions of polymer and catalyst in the mixture, and by altering the type of catalyst. For the purposes of this study, a number of samples were poured in shallow rectangular moulds, yielding samples 20mm wide, 3mm thick and approximately 80mm in length. The samples were mounted and tested in the same manner described above for the spinal cord tissue to ascertain stress-strain behaviour, and to measure their time-dependent properties.

2. Construction of an Anatomically and Mechanically Accurate Physical Model

The physical model of the head/neck complex was constructed as follows. A plastic reproduction of a human skull (Carolina Biological Supply Co., NC) was sectioned sagittally 3cm left of the midsagittal plane, preserving the foramen magnum and atlanto-occipital articulation. Each vertebra in a matching cervical vertebra set had a 1cm section cut from the lamina, between the left facet and spinous process, to allow visualization of the cervical cord while maintaining the normal kinematics and stability provided by the facet joints. Polymer reproductions of the intervertebral disks, whose the compressive mechanical properties were within the range reported for the human intervertebral disk [13] were placed between the vertebrae. Surrogate ligaments, constructed from household elastic, were appropriately pretensioned to match the mechanical properties of human longitudinal ligaments [12] and attached on the most anterior and posterior portions of the vertebrae with epoxy and metal studs. These surrogate ligaments and disks act to stabilize the model, limit the range of motion, and maintain kinematic biofidelity during loading [14].

The surrogate spinal cord model was constructed in a nearly cylindrical lucite mould (1cm diameter, 15cm long) with a gentle flare at one end (diameter=2.5cm) to represent the brainstem. The cylinder was divided longitudinally in two halves. Each part of the mould was sealed at the superior and inferior ends and coated with a releasing agent (petroleum jelly). Sylgard 527 and 184 gels (Dow Corning Inc., MI) were mixed in two concentrations to match the mechanical properties of the human cervical spinal cord and brain tissue at high strain rates [4]. The brain surrogate was poured into each mould near the flared end and the spinal cord surrogate at the opposite ends. The two gels were then gently mixed at the interface. This technique produced a smooth gradient of mechanical properties from the stiffer spinal cord to the softer brain tissue and avoided a mechanical mismatch at the spinal cord-brain interface.

To visualize cord deformation, a grid of enamel dots with 3.0 mm spacing was painted on the mid-sagittal plane surface of the cured gel and allowed to dry thoroughly. The two cord/brainstem surrogate halves were then assembled and more gel poured to join the two halves, yielding the final cord/brainstem model. This was then removed from the mould, placed into the vertebral canal and aligned so that the full grid was visible through the

removed section of lamina. Additional silicone gel simulating brain tissue was poured into the skull portion of the model. A grid was painted on the exposed mid-sagittal plane surface of this brain surrogate and allowed to dry. The remainder of the exposed cranial cavity was filled with silicone gel material, and a Plexiglas plate sealed the intracranial cavity prior to testing. In our model, the cord was left free within the canal, tethered at the distal end by a flexible block of gel, which was able to deform giving limited motion in both superior and inferior directions. The T1 vertebra of the spine/cord model was potted in an aluminium can using Castolite epoxy resin (Buehler, Ltd.) and fixed to a stationary point on a modified HYGE linear actuator (Consolidated Vacuum Corp.). A custom designed linkage converts the linear HYGE motion to an angular motion with allowance for up to 20mm extension of the spinal column, where the degree of angular excursion may be varied by adjusting the linear displacement of the HYGE piston. The skull of the physical model was also potted in a larger aluminium can, which was mounted on to the movable arm of the linkage. The position of the T1 vertebra was adjusted to simulate the normal lordosis of a cervical spine. Figure 2 is a schematic of the finished model, mounted on the actuator.

A series of quasistatic tests were conducted using the model to ensure that the motion of the head/neck system is consistent with normal human motion, and that the motion and quasistatic stretch experienced by the surrogate cord in the vertebral canal is consistent with those seen *in vivo*. The model was moved in 7-10 degree steps by manually positioning the linkage and cylinder. At each position, three colour photographs were taken. The photographs were printed, and each image was then examined to calculate the relative angular rotation of adjacent vertebrae. Further, the length of the anterior and posterior surfaces of the spinal canal and the relative motion between the embedded grid points in the surrogate cord and the spinal canal were measured.

RESULTS

1. Mechanical Testing

Tensile tests of human cadaver cervical and thoracic spinal cord and dura mater yielded stress relaxation curves which demonstrate the viscoelastic nature of the materials, with the dura being approximately 10-15 times stiffer than the spinal cord. Similar tests were conducted on a series of silicone based gels and household elastic in order to select appropriate materials for use as surrogate spinal cord and longitudinal spinal ligaments in a physical analogue of the human head/neck complex.

(a) Human Cadaver Spinal Cord Mechanical Properties

A total of 19 tensile tests were conducted on cervical and upper thoracic spinal cord tissue, at constant strain rates in the range $0.02\text{-}0.3s^{-1}$, with average tangent moduli found to be in the range 0.92-2.8MPa. Figure 3 shows a typical series of stress relaxation curves for three spinal cord samples tested at different strain rates, delineating clearly the time-dependent nature of the tissue subjected to a ramp displacement.

Figure 4 shows the associated family of stress-strain curves for the same samples, demonstrating the characteristic "J-shaped" curve seen in all the samples tested, and the increase in the stiffness of the spinal cords with increasing strain rate. In six tests, significant slippage was noticed between the spinal cord and the grips, and those samples were excluded from further analysis. The calculated tangent moduli for the remaining samples are summarized in Figure 5, where they are plotted against strain rate. This figure clearly demonstrates the increase of the tangent moduli with increasing strain rate, and the apparent plateauing toward the upper end of the testing range. As yet, no strong relationship has been

observed between modulus and donor age, sample location, or time post mortem. Further testing may elucidate such relationships.

(b) Human Cadaver Dura Mater

A small number of samples (n=4) of human cervical dura mater were also tested, at strain rates in the range $0.04-0.5s^{-1}$. The average tangent moduli of these samples increased sharply with increasing strain rate, where the modulus for low strain rate samples was approximately 55MPa. Figure 6 shows the stress relaxation curves for two dura mater samples at different strain rates.

(c) Surrogate Spinal Cord Materials

In contrast to the spinal cord samples, the surrogate materials were found to be essentially linearly elastic, even at large deformations and high strain rates. There was no dependence on loading speed and negligible amounts of relaxation at peak extension. However, variations in the type of catalyst and the polymer to catalyst ratio used in specimen fabrication dramatically altered the stiffness of the gel, in the range 0.003-3.0MPa. A family of sample stress-strain curves of varying composition is shown in Figure 7.

In choosing an appropriate surrogate material made from of this material, it is important to tailor the modulus to match the effective modulus of the human spinal cord at the desired loading rate. As our primary interest is in injury conditions, where it is presumed that the loading rate is high, (duration of loading is usually of the order of 10-100ms) it is therefore logical to select a modulus corresponding to the higher strain rate behaviour of the spinal cord. However, when selecting a linear elastic material whose modulus is approximately constant over the 0-50% strain range to mimic the strongly J-shaped curve of the spinal cord over the same range, a compromise must be made between the effective modulus at high and low strains so that the deformations of the surrogate material are not grossly distorted from those in the human spinal cord under dynamic loading. As mentioned above, the high strain tangent modulus of the spinal cord increases with strain rate, but appears to reach a plateau value of approximately 3-4MPa, at strain rates of $0.3-0.5s^{-1}$. This modulus was best achieved with a gel mix of 29% Sylgard 527 polymer, 68% Sylgard 527 catalyst, with 3% Sylgard 184 catalyst, by volume. Similar considerations guided the selection of an appropriate surrogate material for the brain tissue.

2. Physical Model Validation

In addition to ensuring that each individual element of the physical model mimics adequately the matching human element, it is important to ensure that the model as a whole is kinematically similar, and that appropriate boundary conditions are placed on the model, particularly the surrogate spinal cord. A simple 65° flexion motion was simulated quasistatically using the assembled physical model, and a number of changes in the head/neck structure were measured, including an elongation of the spinal column, stretch of the surrogate ligaments on the dorsal portion of the model, rotation of the vertebrae relative to their neighbours, motion of the spinal cord within the canal, and stretch of the upper spinal cord and lower brainstem surrogates.

During flexion the spinal canal of the model lengthened such that the anterior portion was only slightly elongated, but the posterior length elongated by about 24mm. This elongation of the posterior portion of the surrogate spinal column, along with a rotation of the skull relative to the atlas at the atlanto-occipital joint induced a stretch of the posterior surrogate ligamentous tissues. This is consistent with normal human motion, where both the ligamentous and muscular tissues are stretched in flexion [5].

As part of this motion, each surrogate vertebra rotates between 3 and 7.5 degrees relative to the adjacent vertebrae. Table 1 shows a comparison between the angles measured at given cervical levels for the model to those reported in the literature, as summarized by McElhaney et al [16] for a similar motion. Our model was somewhat stiffer in the lower cervical spine than normal, which was largely due to mounting constraints on the T1 vertebra necessary to fix it on the actuator linkage.

Cervical Level	Model Rotation Angle (deg)	Literature Rotation Angle (deg) [16]
C1/C2	7.5	2-11
C2/C3	6.5	3-8
C3/C4	6.5	3-10
C4/C5	4.5	3-13
C5/C6	4	4-15
C6/C7	3	4-13

Table 1 - Vertebral Rotation Angles for 60° flexion

In the quasistatic flexion tests of the model, the surrogate upper cervical cord was seen to move approximately 2.3mm caudally within the canal, causing the surrogate brainstem to move out through the foramen magnum, since the surrogate brain and spinal cord are a continuous structure.

DISCUSSION

In this study, an anatomically and mechanically accurate physical model of the human head neck complex was developed. Firstly, the mechanical properties of the soft tissue elements of the human cervical spinal column were determined, with particular emphasis on the spinal cord and dura mater. Both the spinal cord and dura mater tissues were found to be viscoelastic in nature, with samples from a given donor stiffening with increasing loading rate. Secondly, candidate surrogate materials for use in the construction of our physical model were also tested, to identify appropriate materials for the spinal cord and longitudinal ligaments. Thirdly, the individual motion of each surrogate vertebral element in the assembled physical model was measured and found to be within the range reported for a similar movement in humans. Finally, the motion of the surrogate spinal cord in the spinal canal was measured and found to correlate with previous studies on the *in situ* motion of the spinal cord [5,15].

Several factors must be considered when evaluating the results from this study. When determining the mechanical properties of human cervical spinal cord and dura mater, all of the samples were obtained from cadavers. This does not guarantee that the mechanical properties thus determined are a completely accurate representation of the *in vivo* properties of the tissues, which in turn limits the accuracy of the mechanical properties of the chosen surrogate materials used in the physical model. However, our results are comparable to *in vivo* results reported for tensile tests of animal spinal cord tissues[6,10,11], which suggests that they are a reasonable estimate of the true values for human spinal cord tissue *in vivo*. The mechanical properties of the surrogate materials were selected to match the average tangent moduli of the human cord at moderate strains under high strain rate loading in order to simulate injurious loading, and because of their linearly elastic stress-strain behaviour, do not mimic exactly the non-linear stress-strain behaviour of the spinal cord across the full range of strains and strain rates used for the validation tests. Finally, the exact motion used to validate the physical model against existing data is prescribed by a simple linkage whose trajectory limits motion to an arc with some longitudinal motion at the T1 vertebra, in the sagittal plane. For a pure

frontal simulated crash (see [5,15]), this sequence is a reasonably accurate representation, however, this cannot imitate all of the complexities of normal flexion movement, or multiplanar motions associated with out of plane loading forces seen in real automotive crash situations.

Previous studies of the mechanical properties of spinal cord tissue in animals by Chang et al [6] and Hung et al [10,11] using low strain rate elongation, and Bilston et al [3], using moderate strain rate testing conditions, may be compared with the data presented here. Our data is not inconsistent with values predicted by both of these studies for these animals, although the data reported by Chang et al and Hung et al [6,10,11] falls at the lower end of the range measured here. This may be due to the *in vitro* nature of our testing, the lack of vascular perfusion pressure, species differences, and especially the much higher strain rates used here. Interestingly, the data for quasistatic testing of a single human cadaveric spinal cord reported by Brieg [5] is similar to that measured here, although no precautions were used in Brieg's testing to avoid dehydration of the tissue.

In the aforementioned *in vivo* MRI study [15], the upper spinal cord, at the C4 level, was found to move 1.5-2.5mm caudally within the spinal canal (in a flexion of 60 degrees), and in quasistatic tests of the physical model described here, the upper cervical cord was found to move caudally a comparable amount. On a more detailed level, the rotation of each surrogate vertebra relative to its neighbours was found to be within the range given by McElhaney et al [16]. This favourable comparison between the motion of the cord and model kinematics in this study and previous investigations suggests that future studies using the model in dynamic loading tests will be a useful tool for estimating the deformations of the spinal during injurious loading.

Central to these future dynamic tests will be an accurate simulation of the head/neck kinematics in automotive crash situations (for example [1,2,7,8,18,26]). In particular, we will focus on the tests conducted by Ewing and Thomas [8] which indicated that the trajectory of heads of human volunteers relative to the T1 vertebra during a simulated frontal impact follows an arc of approximately 55-70 degrees, coupled with an extension of the spinal column along its axis of 15-30mm. The linkage design used here to load our model rotates the skull 65 degrees relative to T1 and allows approximately 20mm of longitudinal extension of the spinal column, which is an adequate representation of the displacement, but for future dynamic tests, an appropriate dynamic loading condition must be applied to represent the true time course of the injury event .

Finally, it is interesting to note the similarities between findings in this report and earlier isolated tissue studies. Saatman and Thibault [23] and Galbraith [9] have shown that isolated myelinated or unmyelinated axons display a response to uniaxial dynamic stretch which is strongly dependent on the strain rate. The mechanical tests of spinal cord tissue reported here are consistent with these findings, in that the spinal cord at higher strain rates is markedly stiffer, and thus may be more susceptible to injury at strain magnitudes which would not, quasistatically, result in functional or structural failure. This mechanical similarity of between the complex spinal cord unit and individual elements of the tissue suggests that isolated tissue studies may have important implications and relevance for understanding the mechanisms of injury to the cervical spinal cord, particularly in those situations where there is little gross structural damage to the spinal column. In addition, the axons and blood vessels within the relaxed spinal cord are wavy, and when the whole structure is elongated, they must first straighten before they begin to stretch and carry load. This may explain the strongly "J-shaped" stress-strain curve seen here when the spinal cord is elongated. Initially, the cord appears soft, up to about 7-10% strain, at which point we hypothesise that the axons and

vascular tissue have "straightened", and become taut, and begin to contribute to the load carrying capacity of the whole structure, resulting in a sharply steepening stress-strain curve.

This study represents one step towards a better understanding of the relationship between macroscopic loading of the head and spinal column and the resulting deformation of the spinal cord during injury. The development and validation of an anatomically and mechanically accurate physical model described here, will, in the next phase of this ongoing research effort, allow a direct measurement of strains in the spinal cord during simulated traumatic loading. In turn, this will permit a more detailed analysis of the mechanical events which initiate the cascade of cellular processes that result clinically in transient and permanent neurological deficits in the spinal cord injured patient.

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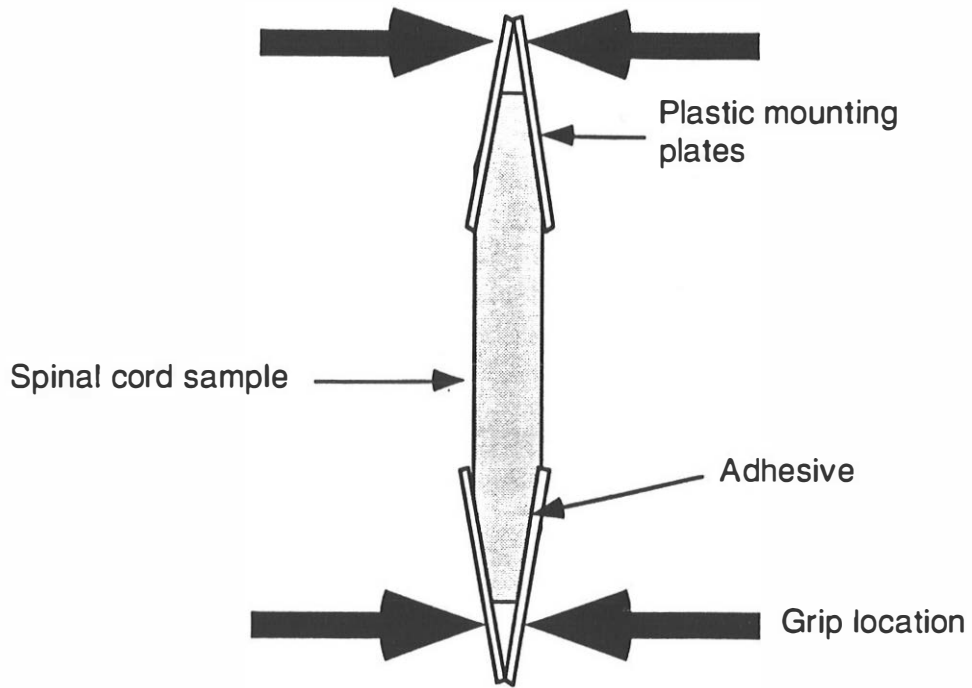
Plastic mounting
plates

Spinal cord sample

Adhesive

Grip location

Figure 1 - Mounting of a spinal cord sample

Mounting Can
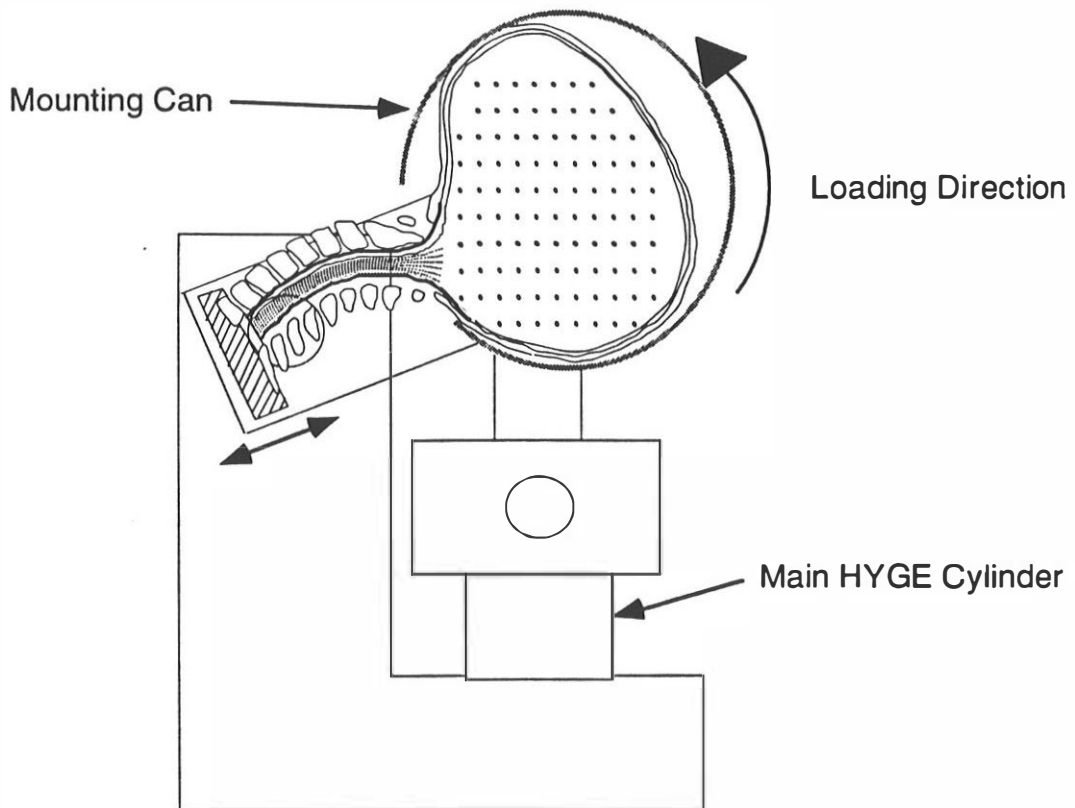
Loading Direction

Main HYGE Cylinder

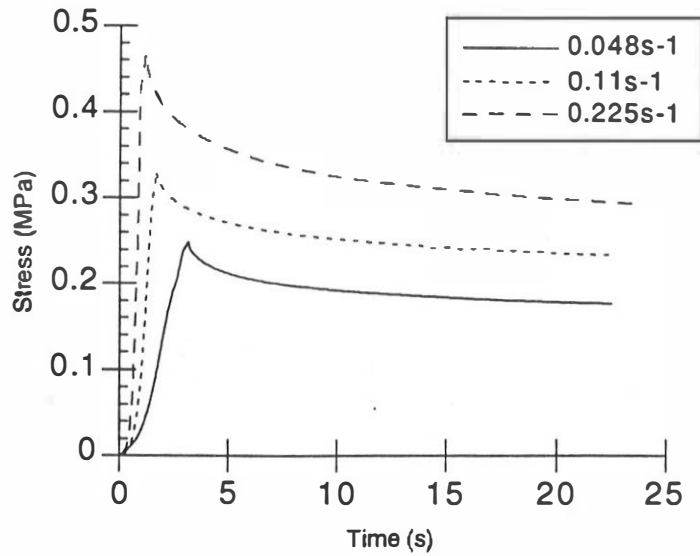Figure 2 - Schematic of the completed model

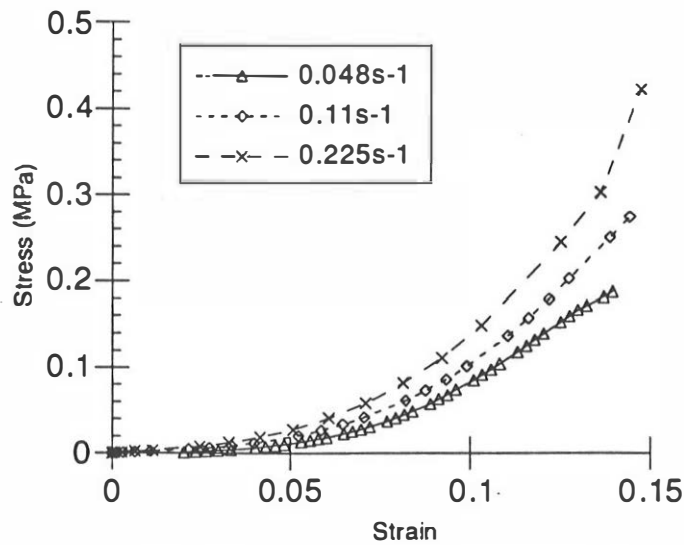Figure 3 - Sample stress-relaxation curves for a spinal cord (high and low strain rates)

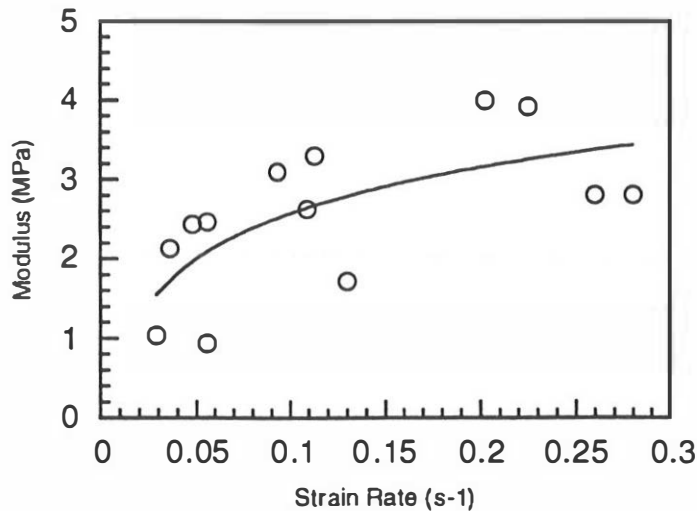Figure 4 - Sample Stress-Strain Curve

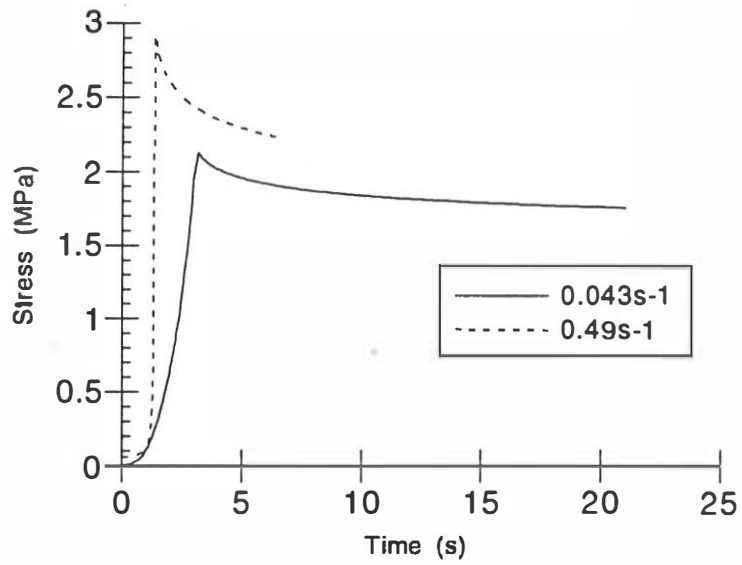Figure 5 - Summary plot showing peak tangent modulus versus strain rate

Figure 6 - Sample stress-relaxation curve for cervical dura mater at a high and a low strain rate
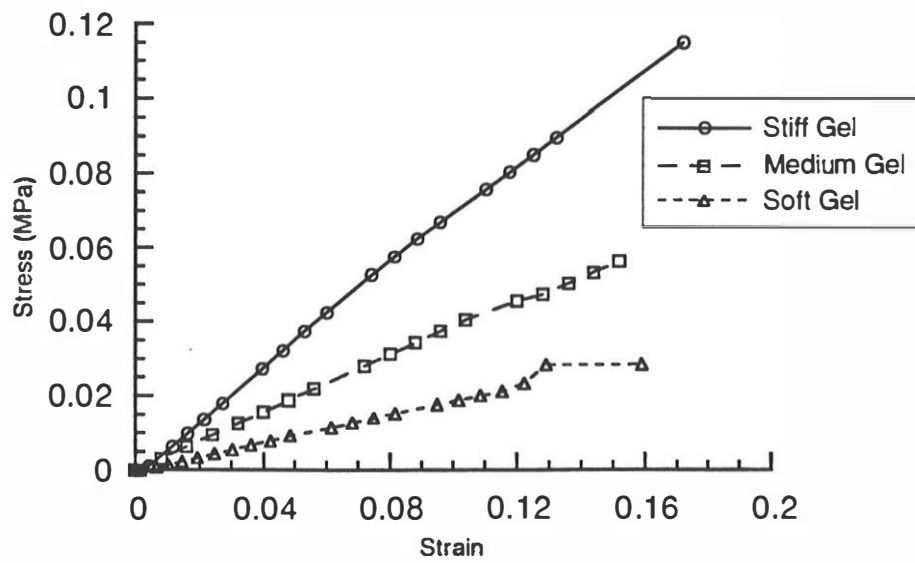

Figure 7 - Sample stress-strain curve for surrogate spinal cord materials