Computer Simulation Studies of Cervical Spine Extension Mechanics

Thomas H. Jansen, Denis J. DiAngelo, Ph.D.
School of Biomedical Engineering, University of Tennessee - Memphis
899 Madison Avenue, Suite 801
Memphis, TN 38163 USA

Abstract—A computer simulation was developed in which the cervical spine was modeled as a biomechanically equivalent linkage system. The model was driven by data from *in vitro* tests of fresh human cadaver spines. The goal of the model was to iteratively determine appropriate rotational stiffness values by comparing the loading vector with that of the experimental data. With a rotational stiffness of 2.0 Nm/deg, the loading vector was 46.7N compared to an experimental value of 45.7N for 35 degrees of rotation.

I. Introduction / Background

The mechanical behavior of human tissue is difficult to model due to complexities such as non-linearity, visco-elasticity, and hysterisis. Models should mimic *in vivo* physiological responses, but most biomechanical models must rely on *in vitro* measurements to obtain accurate data for validation. The protocols for *in vitro* testing should attempt to replicate the complex coupled motions of a joint. Displacement-controlled biomechanical tests allow the motion to be controlled within physiological ranges while measuring the load response of the tissue. In certain situations this technique may tend to complicate the analysis of the mechanics of a tissue structure, but the use of computer simulation software can help alleviate this problem.

In our lab, we use computer simulations to help gain a more thorough understanding and to visualize the complex kinematics and mechanics of the cervical spine during extension tests. We modeled the stiffness of the spine such that it had similar load and motion responses to our tissue tests.

Most computational models of the spine are finite element models which look at the mechanical properties of the spine such as stresses and strains [9]. These types of models do not always produce the most useful information, especially when investigating the cervical spine. Since the cervical spine is a highly mobile region and has multiple bodies, it becomes more useful to examine the kinematics of the spine, and to try to develop a model which can predict its mechanics based simply on stiffness values and geometric parameters [2].

Different researchers have reported different stiffness values for single functional spinal units (FSU) in extension. The following values are given in N-m/deg: Liu et al. reported stiffness values of 1.24 for the middle cervical spine and 2.58 for the lower cervical spine; Moroney et al. reported 0.73, while Shea et al. reported 1.74, and Zidel et al. reported 0.21; Coffee et al. tested 2 FSUs and reported values of 2.29 for the middle cervical spine, and 1.87 for the lower cervical spine [7] [8].

II. Methods

A. *In vitro Spine Testing*

Fresh human cadaveric spines were tested using a flexible Single Actuator Adaptable Programmable Testing Apparatus (SAAPTA). The spines were part of a study investigating the effects of anterior plating systems and strut grafts on sagittal plane spine mechanics [1]. The testing protocol was established after a previous investigation on various end conditions of the cervical spine for *in vitro* testing [3].

The spine was inverted and potted with a bismuth alloy at C2 and T1. C2 was rigidly fixed to a universal force sensor, which was attached to the base of SAAPTA. T1 was connected to a shaft which served as the moment arm for loading the spine. The other end of the shaft slipped through a linear bearing which was mounted with a pinned connection to a vertically oriented actuator as shown in Fig. 1. A load cell was placed in line with the actuator to record the axial force incurred from loading the spine. Rotational transducers were mounted at the pinned linear bearing, as well as at C3 and C7. Translational transducers captured the travel of the shaft through the linear bearing as well as the compression between C3 and C7 within the spine. There was a 200mm offset between the base of the spine and the actuator, such that the full deflection of the actuator produced 35 degrees of extension.

B. *Simulation Spine Model*

For our models, we used the software package Working Model 2D™ by Knowledge Revolution Inc. It incorporates Lagrangian mechanics with animation such that the user has full control of the simulation environment. The numerical integration can be controlled by specifying an animation step, an integrator error, and a numerical method of integration. The Kutta-Merson integration method was used for increased accuracy.

The cervical spine from C2 to T1 was modeled as a linkage system with each vertebral body representing a link. The geometry of each vertebrae was approximated as trapezoidal polygons with dimensions taken from anthropometric studies [4]. The placement of each vertebra

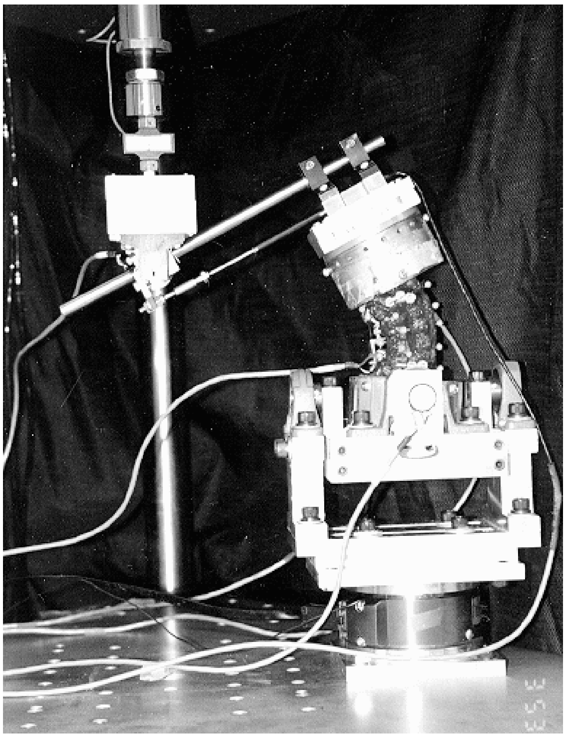

Fig. 1. Experimental setup for cervical spine extension testing

relative to one another was determined from sagittal plane radiographs of the cadaver spines.

The mechanical properties of the intervertebral discs were modeled as rotational springs having a linear stiffness such that the torque acting in the spring was directly proportional to the rotation of the spring. Each spring was assigned the same stiffness value k. The rotational axis for each spring was placed at the center of the subjacent vertebral body [5].

The mechanical joints of SAAPTA were also simulated in Working Model. Motion was induced using a vertically aligned actuator coupled to the cervical spine at a 200mm offset with a shaft and linear bearing as shown in Fig. 2. As the actuator traveled down, the joint coupling the actuator to the shaft and linear bearing allowed free rotation and free translation leaving the spine unconstrained in the sagittal plane.

III. RESULTS

By varying the rotational spring stiffness value and comparing the magnitude of the loading vector with the experimental data, the stiffness values were found to be 2.0 N-m/deg at each vertebral level. For 35 degrees of rotation, the loading vector was found to be 46.7 N, which correlated to the experimental value of 45.7 N.

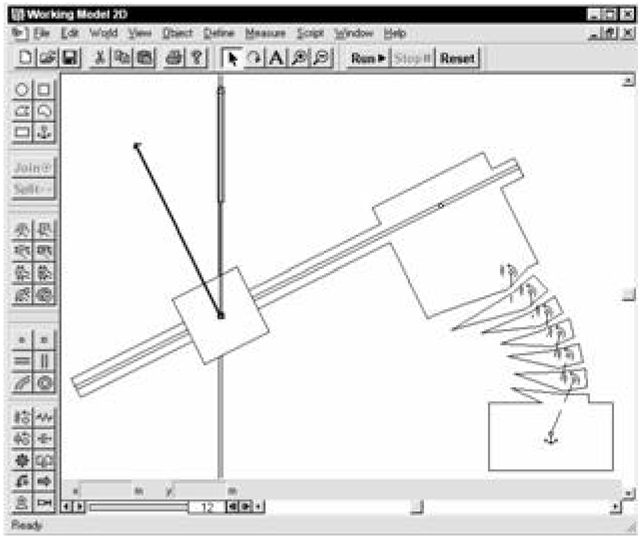

Fig. 2. Working model simulation to verify experimental protocol for cervical spine extension tests.

The limitations of the model were revealed by the moment data which was 9.2 N-m, where as the experimental data showed values of 5.4 N-m.

IV. DISCUSSION / CONCLUSIONS

By modeling the spine as a system with rotational stiffness, and subjecting it to a combined loading vector, it became apparent that the force vector reported in the vertically oriented actuator was not the full loading vector, but only a projected component of it. The true magnitude of the loading vector must be found by realization that the linear bearing of the slider mechanism will only support a force normal to the moment arm. The angular orientation of the moment arm changes as the actuator loads the spine as shown in Fig. 3. This causes the line of action of the loading vector to move closer to the base of the spine, which can cause an increasing moment distribution as you move up the spine.

Other researchers who have similar *in vitro* loading protocols should be cautioned against this, and should design their testing hardware to appropriately correct for such a situation. Studies which test one or two functional spinal units sometimes cite their moment calculation as the vertical force component, F_V times the horizontal distance, d as shown if Fig. 4. For angular deflections less than 20 degrees (which would be expected for one or two functional spinal units) the error in this calculation would be negligible.

The error in moment calculations of the Working Model simulation was inevitably due the moment arm value, which was dependent on the free translation of the slider mechanism. This translation, compared to the experimental setup, was limited in the computational model. This was likely a side effect of disabling the functional spinal unit's ability to translate at the local vertebral level.

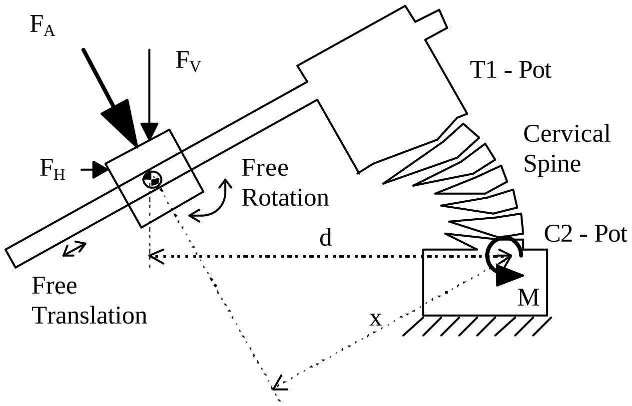

Fig. 3. Force vector diagram showing correct way to calculate the moment at the base of the spine. $M = F_A * x$

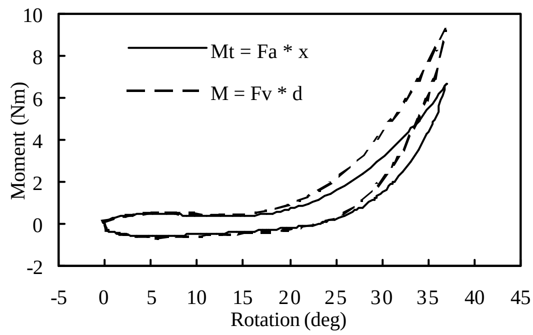

Fig. 4. Stiffness curve taken from experimental data showing the potential error in calculating the moment at the base of the spine.

REFERENCES

[1] DiAngelo, D., Foley, K. T., Jansen, T. H., Dull, S.T., Eckstein, E. C., "Biomechanical analysis of a multi-level corpectomy model: a comparison of strut graft reconstruction alone or concurrent with anterior cervical plating" in press.

[2] Dimnet, J., Pasquet, A., Krag, M.H., and Panjabi, M.M. "Cervical spine motion in the sagittal plane: kinematic and geometric parameters," *J Biomech*, vol. 15, no. 12, pp. 959-969. 1982.

[3] Faber, H. B., DiAngelo, D. J., Foley, K. T., "Development of an experimental testing protocol to study cervical spine mechanics" in press.

[4] Gilad, I., Nissan, M. "Sagittal radiographic measurements of the cervical sand lumbar vertebrae in normal adults," *The British Journal of Radiology*, vol. 58, no.695, pp. 1031-1034. November 1985.

[5] Jansen, T., DiAngelo, D. "Influences of the location of vertebral 'IAR' on cervical spine kinematics" in press.

[6] Lind, B., Sihlborn, H., Nordwall, A., Malchau, H. "Normal range of motion of the cervical spine," *Arch Phys Med Rehabil*, vol. 70, pp. 692-695. 1989.

[7] Moroney SP, Schultz AB, Miller JA, Andersson GB, "Load-displacement properties of lower cervical spine motion segments," *J Biomech*, vol. 21, no. 9, pp. 769-79. 1988.

[8] Shea, M., Edwards, W.T., White III, A.A., Hayes, W.C. "Variations of stiffness and strength along the human cervical spine," *J.Biomechanics*, vol. 24, no. 2, pp. 95-107. 1991.

[9] Yoganandan, N., Kumaresan, S., Voo, L., Pintar, F. "Finite element applications in human cervical spine modeling," *Spine*, vol. 21, no. 15, pp. 1824-1834. 1996.