DEVELOPMENT OF A ROBUST THREE-DIMENSIONAL MATHEMATICAL MODEL OF THE CERVICAL SPINE.

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INTRODUCTION

The present paper highlights some of the objectives of an ongoing investigation using a three dimensional mathematical parametric finite element model of the cervical spine. A single functional unit (FU, C5-C6) as well as a complete model of the cervical spine (C3-C7) were validated based on existing experimental data. As a result of the validation, several other variables were able to be analyzed, including stresses, non-linear behavior and preload effects.

The use of parametric finite element analysis permits several advantages over CT scan based models, which are limited to the geometry of the patient from which they were extracted. Parametric models, on the other hand, reconstruct the biological structures by averaging the dimensions of the general population and establishing a basic set of parameters that define the geometry. Parametric models also allow the identification of the main parameters that influence the solution and permit patient specific models. The model combines anatomical data, intricate mathematical description of vertebral body facets and surfaces and CT scans when desirable.

METHOD

A parametric finite element model of the cervical spine from C3 to C7 was developed. A set of parameters were defined after thoroughly researching anatomical and morphological descriptions of the cervical spine available in the current medical literature.

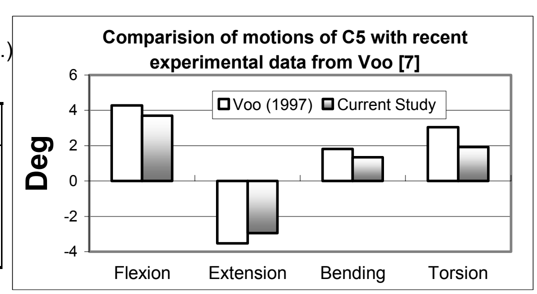
Due to the wide variety of descriptions in the literature and the lack of coherence between existing data, several mathematical relations were formulated to fully describe the geometry of the cervical spine. Using the aforementioned parameters, a CAD and finite element models were created by means of a unique assembly technique. For a full description of the procedure, see Lopez-Espina [1].

The validation of the model was performed in two stages. In the first stage, a single FU was analyzed with each load case: flexion, extension, lateral bending and axial torsion. In each load case, the load was applied on the upper endplate of C5 and ranged from 0 to 3Nm in steps of 0.2Nm. Each load case was analyzed under three different preloading conditions set at 0, 50 and 100 N perpendicular to the upper endplate of C5. This was done to simulate the effect of the self weight of different fixtures used in vitro testing [2]. Thus, in total, 12 different loading cases were studied. All the loads were applied according to a coordinate system with the origin in the posterior-most point of the superior endplate of C5 in the midsagital plane. The positive x-axis was pointed forward and the z-axis perpendicular to the endplate. In the second stage, the validation was performed on the entire cervical range, from C3 to C7. The same load cases were used but without preloading. The forces ranged from 0 to 1.6 Nm in steps of 0.4 Nm. The load was applied in the upper endplate of C3, according to a coordinate system equivalent to the previous one, but in C3. To simulate the loading that a rigid fixture will produce in the loaded vertebra, the upper endplate of this one was made virtually rigid.

The results from the validation process were post-processed to obtain several values for each step in each load case. For the kinematics, the absolute rotation and translation of each vertebra with respect to the fixed coordinate system, and the relative rotation and translation of each vertebra with respect to the vertebrae directly beneath them was calculated. The maximum Von Misses stress for each of the following parts in each vertebra was also calculated: upper and lower endplate, cortical and cancellous bone, facet lateral masses, posterior elements and disc annulus and nucleus. Finally, the number of fibers in tension

Table 1 Comparison of a C5C6 Functional Unit with various experimental data(Deg for 1.8Nm.)

	Current				Moroney				Goel				Maurel				Voo			
	FLE	EXT	BD	TOR	FLE	EXT	BD	TOR	FLE	EXT	BD	TOR	FLE	EXT	BD	TOR	FLE	EXT	BD	TOR
Rx	0.0	0.0	**1.35**	-0.07	0.34	0.11	**4.71**	-0.9	0	0.11	**3.03**	-1.58	0	0.11	**6.3**	0.8			**1.81**	
Ry	**3.7**	**-2.9**	0.26	0.35	**5.55**	**-3.52**	0.25	0.3	**5.7**	**-3.69**	0.92	0.65	**7.8**	**-8.7**	-0.6	0.9	**4.28**	**-3.5**		
Rz	0.0	0.0	0.12	**1.93**	0.34	0.04	-1.5	**1.8**	0	0.04	-1.55	**2.42**	0	0.04	0.9	**10.4**				**3.05**



and the percentage of contact in the facet joint were obtained.
Special attention was given to the calculation of the translation and rotation of each vertebra. The vertebral body is assumed to be quasi rigid and movement of each point is governed by

$$y = R \cdot x + d$$

where y is the position of the point after rotation, x is the original position of the point, R is the rotation matrix and d is the displacement. The position and displacement of four points in each vertebra were obtained and then used to compute R and d, following the algorithm proposed by Soederqvist [4]. From the rotation matrix the Euler angles where obtained. The previous method gave the rotation matrix in the fixed coordinate system, but since this matrix was known for each vertebra, the rotation of each vertebra with respect to the one directly beneath was also calculated.

RESULTS

The validation was done by comparison of the kinematical behavior of the model under several loads, with the kinematical results obtained from in vitro test [1] and finite element models [5,6,7]. Table 1 shows this comparison. In most of the cases the model response is in good agreement with the literature, specially the model presented by Voo []. The only case, in which the model provides different values with respect to the in vivo test, is for the case of lateral bending. Several reasons justify this difference. In the in vivo test, only the angle of the fixture attached to the endplate was recorded; while the present study takes in account the rotation of the vertebra as a whole, computing an averaged rotation/translation that best fit the observed displacement. It is also been shown that small changes in the material properties, as well as in the geometrical parametrical parameters, can lead to significant changes in the response of the spine [8]. In a side analysis done by the authors, the principal response in bending can be doubled by doubling the inter facet gap distance, which corroborates the variability of the results.
The whole cervical segment was validated using the results presented by Panjabi et al [3]. A good fit is observed, and the same aforementioned comments apply for these results. Figure 1 shows these results for the principal motions.
More importantly the model developed presents some interesting insight into the behavior of the disc fibers during loading. A common observation in spine studies is its non-linear behavior [3,6,7,9] as is the case shown in (Figure 2). Observe how the number of fibers in tension displacement (strain) rapidly increases for small loads, then levels off when it reaches 2.5 lb. This is behavior is common to all the loading cases tested. This effect might be responsible for the effect known as neutral zone, which is an area of load stiffness that appears at small loads [9].
Furthermore, Figure 2 shows how the percentage of contact area in the facets computed as function of the number of nodes in contact, increases. This can be explained by: a) the increase in the stiffness of the segment depicted in the load-displacements curves, b) the stress-load curves change the slope when the load increases. Some of the parts of the structure increase the slope; others decrease, depending in whether an increase in facet contact is elevated due to the increase in load share. This is shown to be true for all load cases except the flexion where there is no contact and still changes in slope of stress-load curves occur. Therefore, some of these effects have to be related to the fibers or to the non-linearity behavior of the ligaments. These changes can also be seen in the whole spine results. The effects of preload seam to be small on the kinematics of the cervical spine yet can be significant in the case of stresses, fibers in tension or facet contact.

CONCLUSION

A mathematical parametric cervical spine model was developed, using a limited number of design parameters. These parameters are identified and selected based on their critical function and their role in the anatomical condition of the cervical spine. A total number of 212 loads were applied for this study. The present study focuses in deterministic responses of the cervical spine, and does not quantify for the effects of uncertainties in the responses derived from material properties and geometry variability. Several other related responses where analyzed and related to further understand the biomechanics of the cervical spine.

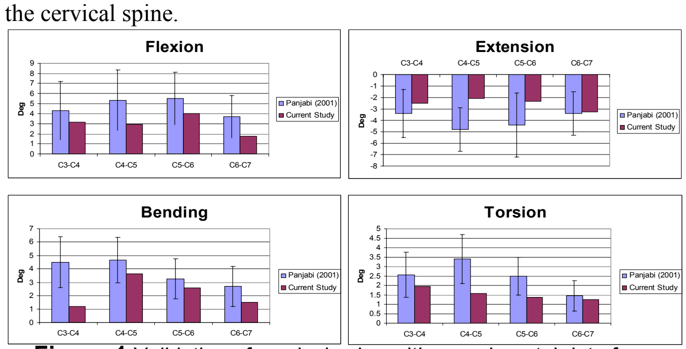


Figure 1 Validation of cervical spine with experimental data from Pamjabi [3]

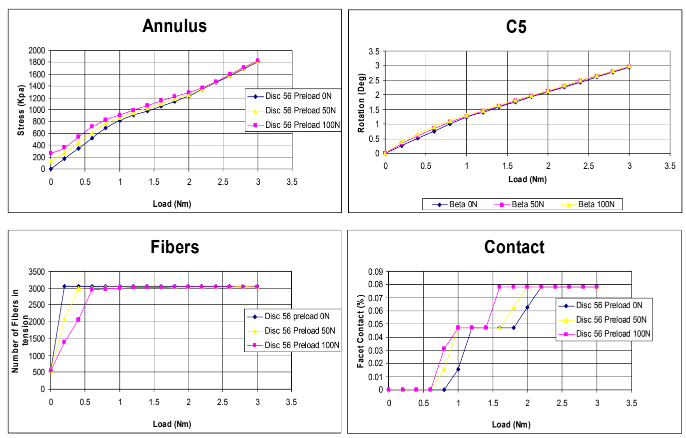


Figure 2 Results of a C5-C6 Functional Unit in torsion

REFERENCES

1. Carlos Lopez-Espina et al., ASME International Mechanical Engineering Congress – Bioengineering division – (November 2002)
2. Moroney SP et al. J Biomech 1988;21(9):769-79
3. Panjabi MM et al., Spine 2001 Dec 15;26(24):2692-700
4. Soderkvist I. and Wedin P. -A., Journal of Biomechanics, 26:1473-1477.
5. Goel VK, Clausen JD. Spine 1998 Mar 15;23(6):684-91
6. Maurel N, Lavaste F, Skalli *W.* J Biomech 1997 Sep;30(9):921-31
7. Voo LM, Kumaresan S, Yoganandan N, Pintar FA, Cusick JF. Spine 1997 May 1;22(9):964-9
8. Kumaresan S, Yoganandan N, Pintar FA. Clin Biomech (Bristol, Avon) 1999 Jan;14(1):41-53
9. Clark. *The cervical Spine.3rd Ed*. Lippincott – Raven, 1998.