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Short communication

Pure moment testing for spinal biomechanics applications: Fixed versus sliding ring cable-driven test designs

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ABSTRACT

In vitro multi-axial bending testing using pure moment loading conditions has become the standard in evaluating the effects of different types of surgical intervention on spinal kinematics. Simple, cable-driven experimental set-ups have been widely adopted because they require little infrastructure. Traditionally, "fixed ring" cable-driven experimental designs have been used; however, there have been concerns with the validity of this set-up in applying pure moment loading. This study involved directly comparing the loading state induced by a traditional "fixed ring" apparatus versus a novel "sliding ring" approach. Flexion-extension bending was performed on an artificial spine model and a single cadaveric test specimen, and the applied loading conditions to the specimen were measured with an in-line multiaxial load cell. The results showed that the fixed ring system applies flexion-extension moments that are 50–60% less than the intended values. This design also imposes non-trivial anterior–posterior shear forces, and non-uniform loading conditions were induced along the length of the specimen. The results of this study indicate that fixed ring systems have the potential to deviate from a pure moment loading state and that our novel sliding ring modification corrects this error in the original test design. This suggests that the proposed sliding ring design should be used for future in vitro spine biomechanics studies involving a cable-driven pure moment apparatus.

1. Introduction

In vitro experimental assessment of spinal implant devices and surgical techniques is important to the development and design optimization of new clinical technologies. These studies typically compare spinal flexibility between intact and treated cadaveric specimens in multiple anatomic directions. Among the methods of acquiring such data, pure moment biomechanical conditions are preferred because they ensure uniform loading along the column of the spine easing the comparison of techniques and technologies with previous literature (Wilke et al., 2001; Wilke et al., 1998). A variety of systems are used to apply pure bending moments to the cadaveric spine, including 6-axis testing machines (Beaubien et al., 2005; Kotani et al., 2005; Wilke et al., 1994; DiAngelo et al., 2004; Schwab et al., 2006; Kotani et al., 2006; Panjabi et al., 2007a, b; Panjabi, 2007; Crawford et al., 1995), suspended deadweights (Melcher et al., 2002; Puttlitz et al., 2004; Goel et al., 1988; Stanley et al., 2004), and cable-driven systems (Crawford et al., 1995; Acosta et al., 2008; Barnes et al., 2009). The cable-driven test set-up applies a couple via a continuous loop of cable that is wound around a "loading ring" attached to one end of the spinal segment. Since its development by Crawford in the mid-90s (Crawford et al., 1995), the cable-driven pure moment method has been widely adopted due to its simplicity and relatively low infrastructure requirements (it requires only a uni-axial testing frame). Recently, it has been called into question whether the cable driven pure moment test design is capable of generating a pure moment loading state (Panjabi, 2007).

The goal of this study was to determine whether the cable-driven pure moment apparatus generates accurate, consistent loading conditions for a range of possible multi-axial spinal testing scenarios. We considered differences in applied loads generated by the "fixed loading ring" design, which has been used historically for cable-driven pure moment testing(Crawford et al., 1995; Acosta et al., 2008; Barnes et al., 2009), and a modified "sliding loading ring" system that was developed by our group to address the limitations of the fixed ring design.

2. Methods and materials

2.1. Biomechanical testing set-up

The applied loading conditions for two different cable-driven pure moment set-ups were evaluated using an artificial lumbar spine model. The 5-segment spine model was constructed from common laboratory materials. Briefly, this model consisted of wooden cylinders simulating the lumbar vertebrae (50 mm

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length × 50 mm diameter) and neoprene rubber pads for the intervertebral discs (Shore durometer hardness 30, 15 mm height).

The artificial spine was tested quasi-statically in flexion and extension using an established pure moment testing protocol (Crawford et al., 1995). Flexion/extension moments were applied to maximum moment of 4.5 in 1.5 Nm increments with a 20 s hold per step. Three cycles were applied, and data were collected on the final cycle. This maximum load level was determined during preliminary testing to be well within the non-destructive testing range of the artificial spine. The testing platform was a standard servohydraulic testing frame (Mini Bionix 858, MTS, Eden Prairie, MN) with a uniaxial load cell mounted to the actuator head (Fig. 1). Rigid body markers were attached to each vertebra in the artificial spine construct, and a 3D camera system (Optotrak 3020, NDI, Waterloo, Canada) was used to measure the relative motion of each vertebra (0.1 degree accuracy). Multi-axial loads were monitored continuously throughout testing via a multi-axial load cell (AMTI MC3A-6-500, Advanced Mechanical Technology, Inc., Watertown, MA) connected to the caudal end of the specimen.

2.2. Cable-driven pure moment systems

The "fixed ring" cable driven testing system used in this experiment was identical to that described by Crawford (Crawford et al., 1995) (see Fig. 1). Briefly, it consisted of a loading ring (72 mm OD) affixed to the cranial specimen cup. A braided Spectra cable (200 lb capacity) was wound around the loading ring in the appropriate configuration to either induce flexion or extension. The cable was then routed to the actuator of the test machine via low friction pulleys attached to a static frame. The position of the pulleys was adjusted at the beginning of the test to achieve co-linearity of the cables as they extended from the loading ring.

The "sliding ring" test set-up was identical to the fixed ring set-up except in the design of the loading ring (see Fig. 1). A set of bearings—one in the axial direction and the other in the anterior–posterior direction—were rigidly attached to the cranial specimen cup, and the loading ring was allowed to float along these bearings. In the axial direction, the loading ring was counter balanced by light duty springs.

2.3. Specimen configurations

The fixed versus sliding loading ring test set-ups were compared for several different test specimen configurations. For the 5-segment spine model, tests were performed with and without standard bilateral posterior spinal screw-rod fusion. The 5-segment model was then sequentially "dissected" to 2 segments, with biomechanical tests performed at each different specimen length. Lastly, pure moment tests were conducted on fresh-frozen human cadaveric tissue ($N=1$, L3-S1, 48 y.o., female).

2.4. Outcome measures

Fixed vs. sliding ring designs were compared across several outcome measures throughout the flexion-extension bending cycle: (1) intended versus applied moment, where "intended moment" is the desired load level (4.5 Nm maximum in increments of 1.5 Nm), and "applied moment" was the moment actually experienced by the specimen (measured by the multi-axial load cell); (2) anterior–posterior shear loads; and (3) range-of-motion (ROM) at each intervertebral level. A true pure moment loading state would be indicated by a one-to-one relationship between intended and applied moment, zero anterior–posterior shear forces, and—for the artificial spinal section only—equivalent ROM at each intervertebral level (due to geometric and material heterogeneity).

3. Results

For the fixed ring set-up, FE moments were 52–59% less than intended values (Table 1), and anterior–posterior shear forces reached 10.3 N. FE ROM decreased cranially-to-caudally for the 5 level specimen from 2.6 degrees at the L1/L2 to 0.4 degrees at the L4/L5 motion segment. Shorter spinal sections (i.e. 4, 3, and 2 levels) demonstrated resultant loads that were opposite in direction from the intended loading, i.e., extension moments while flexing forward, and had relatively higher anterior–posterior shear forces of 12.6–15.7 N. Increasing specimen rigidity did not mitigate loading artifacts (see Table 1).

For the sliding ring set-up, specimen length and rigidity had little effect on resultant loads (5.8% maximum variation across all configurations). Differences between resultant and intended moments were 3.8–9.0% across all test configurations, and shear forces ranged from 3.6 to 6.6 N. Flexion-extension ROM were consistently in the range of 6.2–7.2 degrees for all testing configurations and intervertebral levels (Fig. 2a).

For the cadaveric specimen, FE ROM at L5/S1 was 9.8 degrees for the sliding ring set-up versus 2.4 degrees for the fixed ring. The sliding ring values are in-line with non-fixed ring, cable driven set-ups in the literature (9–14 degrees (Tzermiadianos et al., 2008; Panjabi et al., 1994)) (Fig. 3).

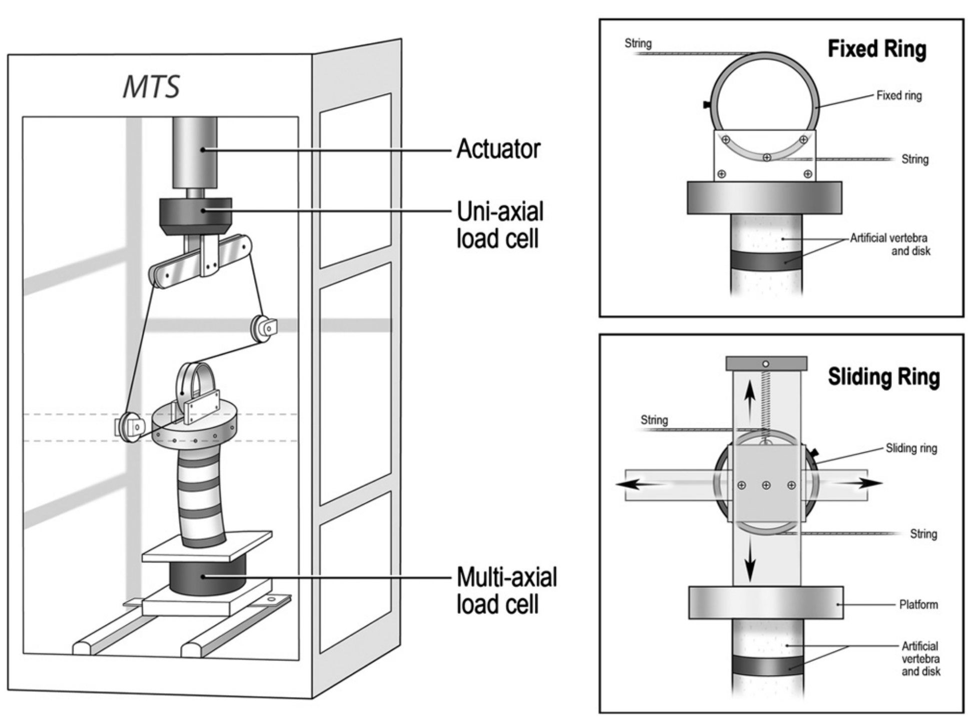

Fig. 1. Schematic of the cable-driven pure moment test set-up. Both fixed and sliding ring designs are shown.

Table 1
Results from biomechanical testing on artificial spine using the fixed ring test jig. Data are presented for maximum intended moment (±4.5 Nm) in flexion-extension. (+) Flexion, (+) Anterior.

No. spinal levels	Normal/Fusion	Range-of-motion (deg)		Applied moment (Nm)		Anterior–posterior shear (N)	
		Sliding ring	Rigid ring	Sliding ring	Rigid ring	Sliding ring	Rigid ring
5	Normal	20.6[a]	4.1[a]	4.3	1.9	6.6	10.3
5	Fusion	8.1[a]	2.9[a]	4.1	1.5	5.7	12.9
4	Normal	20.6	6.3	4.3	1.4	6.2	14.1
3	Normal	13.1	4.9	4.3	1.0	4.2	15.7
2	Normal	6.2	3.5	4.3	0.3	3.6	12.6

[a] ROM of the superior most motion segment of the 5 level pseudo segment was not recorded.

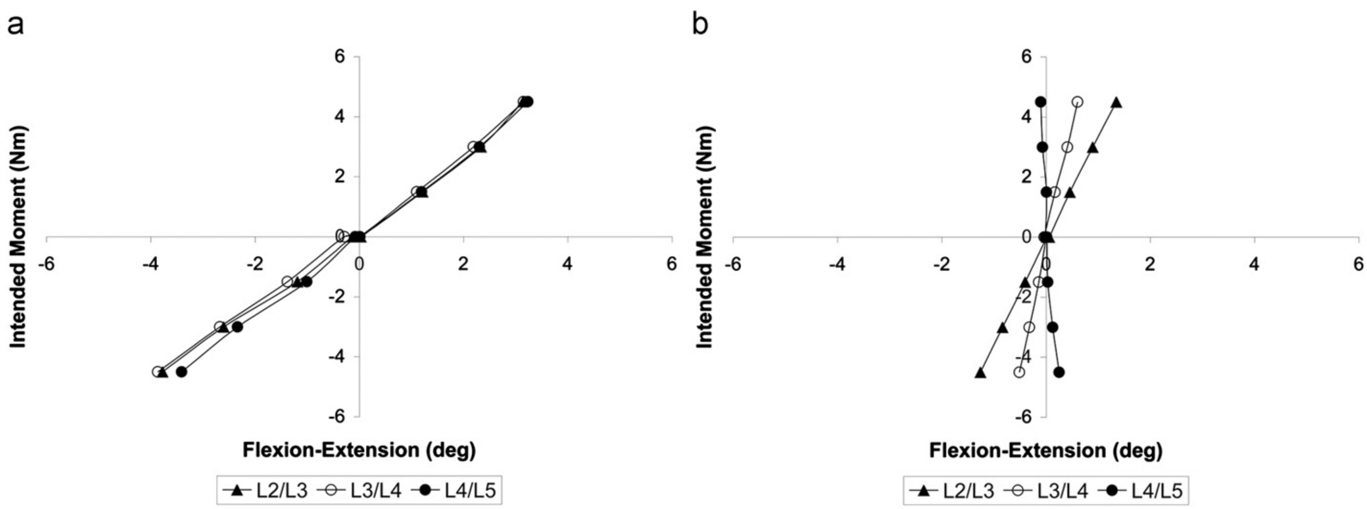

Fig. 2. Moment-angle curves for pure moment testing of 5-level artificial spine (unfused). (a) Sliding ring set-up; (b) fixed ring set-up.

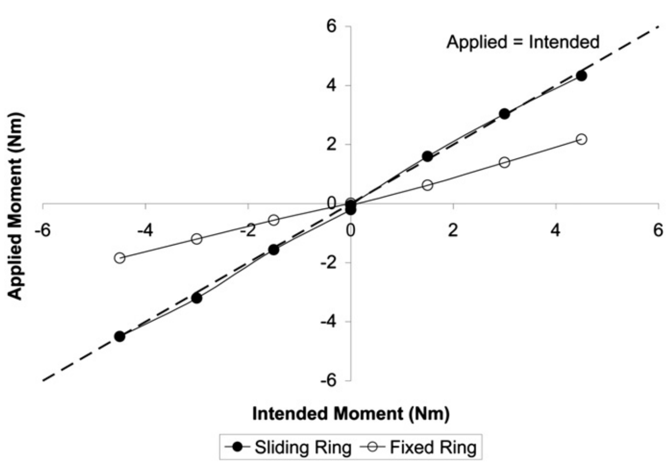

Fig. 3. Intended versus applied moment for the sliding and fixed ring set-ups. The applied moment was measured from the load cell at the base of the specimen. Results are presented for the unfused, 5-level artificial spine specimen.

4. Discussion

The results of this study indicate that the standard fixed ring cable-driven pure moment system (Crawford et al., 1995) has the potential to deviate from a pure moment loading state and that our novel sliding ring modification corrects this error in the original test design. In this study, fixed ring system moments were 50–60% less than the intended values. This design also induced non-trivial shear forces, and non-uniform loading conditions were induced along the length of the specimen, as was evident from variations in segmental ROM for the uniform artificial spine specimen.

These artifacts may be attributed to two inherent shortcomings of the fixed ring design. It has been previously asserted that (Panjabi, 2007) cable co-linearity cannot be maintained throughout testing without manual adjustment of the pulley position at each load interval. Manual adjustment does introduce a high level of user error into the protocol; however, preliminary studies by our group indicated that cable co-linearity was not the principal contributor to the loading artifact. The more likely culprit is a "tethering" effect present in the fixed ring design. The fixed length of the cable used to induce the moment inherently couples the anterior–posterior and axial translation of the loading ring, which, in the case of the fixed ring set-up, is rigidly attached to the cranial end of the specimen. This "tethering" can cause erroneous reverse loading at the base of the specimen. This artifact is not present in the new sliding ring design because the translation of the cranial end of the specimen is decoupled from the loading ring via the bearings.

Deviation from pure moment loading conditions alone would not invalidate previous studies that have used the fixed ring technique, as most make use of a repeat-measures test design. However, the fact that the magnitude of the error depends on the inherent rigidity of the specimen does call into question the results of previous studies in which treatment groups varied substantially in terms of rigidity. Furthermore, our results indicate that inter-study comparisons of similar treatments may be confounded by differences in test specimen length. The new sliding ring test system corrects these errors in the cable-driven pure moment test design by inducing the prescribed pure moment state regardless of specimen length or inherent rigidity. For this reason, we recommend that the sliding ring technique be used in place of fixed rings for all future studies involving the cable-driven pure moment approach.

Conflict of interest statement

All authors acknowledge that they have nothing to disclose in terms of personal relationships with other individuals or organizations that could inappropriately influence their work including those brought forth by employment, consultancies, stock ownership, honoraria, paid expert testimony, patent applications/registrations, and grants.

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References

Acosta Jr., F.L., Buckley, J.M., Xu, Z., Lotz, J.C., Ames, C.P., 2008. Biomechanical comparison of three fixation techniques for unstable thoracolumbar burst fractures. Laboratory investigation. Journal of Neurosurgery: Spine 8, 341–346.

Barnes, A.H., Eguizabal, J.A., Acosta Jr., F.L., Lotz, J.C., Buckley, J.M., Ames, C.P., 2009. Biomechanical pullout strength and stability of the cervical artificial pedicle screw. Spine (Phila Pa 1976) 34, E16–E20.

Beaubien, B.P., Derincek, A., Lew, W.D., Wood, K.B., 2005. In vitro, biomechanical comparison of an anterior lumbar interbody fusion with an anteriorly placed, low-profile lumbar plate and posteriorly placed pedicle screws or translaminar screws. Spine (Phila Pa 1976) 30, 1846–1851.

Crawford, N.R., Brantley, A.G., Dickman, C.A., Koeneman, E.J., 1995. An apparatus for applying pure nonconstraining moments to spine segments in vitro. Spine (Phila Pa 1976) 20, 2097–2100.

DiAngelo, D.J., Foley, K.T., Morrow, B.R., Schwab, J.S., Song, J., German, J.W., Blair, E., 2004. In vitro biomechanics of cervical disc arthroplasty with the ProDisc-C total disc implant. Neurosurgery Focus 17, E7.

Goel, V.K., Clark, C.R., Gallaes, K., Liu, Y.K., 1988. Moment-rotation relationships of the ligamentous occipito-atlanto-axial complex. Journal of Biomechanics 21, 673–680.

Kotani, Y., Cunningham, B.W., Abumi, K., Dmitriev, A.E., Ito, M., Hu, N., Shikinami, Y., McAfee, P.C., Minami, A., 2005. Multidirectional flexibility analysis of cervical artificial disc reconstruction: in vitro human cadaveric spine model. Journal of Neurosurgery Spine 2, 188–194.

Kotani, Y., Cunningham, B.W., Abumi, K., Dmitriev, A.E., Hu, N., Ito, M., Shikinami, Y., McAfee, P.C., Minami, A., 2006. Multidirectional flexibility analysis of anterior and posterior lumbar artificial disc reconstruction: in vitro human cadaveric spine model. European Spine Journal 15, 1511–1520.

Melcher, R.P., Puttlitz, C.M., Kleinstueck, F.S., Lotz, J.C., Harms, J., Bradford, D.S., 2002. Biomechanical testing of posterior atlantoaxial fixation techniques. Spine (Phila Pa 1976) 27, 2435–2440.

Panjabi, M.M., 2007. Hybrid multidirectional test method to evaluate spinal adjacent-level effects. Clinical Biomechanics (Bristol, Avon) 22, 257–265.

Panjabi, M., Henderson, G., Abjornson, C., Yue, J., 2007a. Multidirectional testing of one- and two-level ProDisc-L versus simulated fusions. Spine (Phila Pa 1976) 32, 1311–1319.

Panjabi, M., Malcolmson, G., Teng, E., Tominaga, Y., Henderson, G., Serhan, H., 2007b. Hybrid testing of lumbar CHARITE discs versus fusions. Spine (Phila Pa 1976) 32, 959–967.

Panjabi, M.M., Oxland, T.R., Yamamoto, I., Crisco, J.J., 1994. Mechanical behavior of the human lumbar and lumbosacral spine as shown by three-dimensional load-displacement curves. Journal of Bone and Joint Surgery Am 76, 413–424.

Puttlitz, C.M., Melcher, R.P., Kleinstueck, F.S., Harms, J., Bradford, D.S., Lotz, J.C., 2004. Stability analysis of craniovertebral junction fixation techniques. Journal of Bone and Joint Surgery Am 86, 561–568.

Schwab, J.S., Diangelo, D.J., Foley, K.T., 2006. Motion compensation associated with single-level cervical fusion: where does the lost motion go? Spine (Phila Pa 1976) 31, 2439–2448

Stanley, S.K., Ghanayem, A.J., Voronov, L.I., Havey, R.M., Paxinos, O., Carandang, G., Zindrick, M.R., Patwardhan, A.G., 2004. Flexion-extension response of the thoracolumbar spine under compressive follower preload. Spine (Phila Pa 1976) 29, E510–E514.

Wilke, H.J., Claes, L., Schmitt, H., Wolf, S., 1994. A universal spine tester for in vitro experiments with muscle force simulation. European Spine Journal 3, 91–97.

Wilke, H.J., Rohlmann, A., Neller, S., Schultheiss, M., Bergmann, G., Graichen, F., Claes, L.E., 2001. Is it possible to simulate physiologic loading conditions by applying pure moments? A comparison of in vivo and in vitro load components in an internal fixator. Spine (Phila Pa 1976) 26, 636–642.

Wilke, H.J., Wenger, K., Claes, L., 1998. Testing criteria for spinal implants: recommendations for the standardization of in vitro stability testing of spinal implants. European Spine Journal 7, 148–154.

Tzermiadianos, M.N., Mekhail, A., Voronov, L.I., Zook, J., Havey, R.M., Renner, S.M., Carandang, G., Abjornson, C., Patwardhan, A.G., 2008. Enhancing the stability of anterior lumbar interbody fusion: a biomechanical comparison of anterior plate versus posterior transpedicular instrumentation. Spine (Phila Pa 1976) 33, E38–E43.