STIFFNESS PROPERTIES OF HUMAN LUMBAR INTERVERTEBRAL DISCS IN COMPRESSION AND THE INFLUENCE OF STRAIN RATE

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ABSTRACT

There have been numerous researchers that have investigated the properties of human intervertebral discs. However, there has been no attempt to characterize the effects of dynamic loading on the compressive stiffness of human lumbar intervertebral discs. Therefore, the purpose of this study was to develop the compressive stiffness properties of lumbar intervertebral discs when subjected to various dynamic compressive loading rates. This was accomplished by performing a total of 33 axial compression tests on 11 human lumbar intervertebral discs dissected from 6 fresh frozen human cadavers, 5 male and 1 female. The adjacent vertebral bodies were fixed to a load cell with a custom aluminum pot and then subjected to three dynamic compressive loading rates using a servo-hydraulic Material Testing System: 6.8, 13.5, and 72.7 strain/ sec. The results show that the compressive stiffness of lumbar intervertebral discs is dependent on the loading rate. There was no significant correlation ($p > 0.05$) between functional spinal unit compressive stiffness and vertebral level at any of the three loading rates. Therefore, a linear relationship between loading rate and vertebral disc compressive stiffness was developed by curve fitting the stiffness data from the current study along with static compressive stiffness data reported by previous studies.

INTRODUCTION

It is estimated that the combined overall cost of vertebral fractures in North America is approximately $750 million dollars a year [19]. Vertebral fractures can occur as a result of moderate trauma, falls from standing height or less, as well as severe trauma, falls from greater than standing height or motor vehicle accidents [4]. A common fracture seen in motor vehicle accidents is anterior wedge fractures, caused by combined flexion and axial compression [13]. In addition, the increased risk of vertebral fractures with age is directly linked to increased incidence of osteoporosis in individuals over 45 [16].

In order to understand and reduce these injuries, various mathematical and mechanical models of the human spine have been developed. Given that the mechanical response of biological tissues demonstrates some degree of rate dependence, these models must be validated using mechanical properties obtained at the appropriate loading rates in order to accurately simulate spine kinematics and predict injury.

The literature on the biomechanics of the spine has primarily focused on the failure properties of isolated vertebral bodies or functional spinal units (FSU), defined as an intervertebral disc and all or part of the two adjacent vertebral bodies, or the compressive stiffness under static and quasi static different loading conditions [1, 2, 3, 7, 14, 23]. The compressive failure force and stiffness of isolated vertebral bodies have both been shown to increase with increasing of loading rate [9, 10]. Sundararajan et al. [2005] reported that the shear failure force of FSUs increases with increasing of loading rate. There have been a few studies that have investigated the viscoelastic response of functional spinal units in axial compression through static creep or stress relaxation testing [11, 12, 15]. Smeathers and Jones (1988) conducted cyclic axial compression tests on lumbar FSUs at 0.01 Hz, 0.1 Hz, 1.0 Hz, and 10 Hz, and found a moderate increase in compressive stiffness with increasing frequency. However, Smeathers and Jones (1988) used a large preload of 750 N and loaded the specimens to ± 250 N, which resulted in loading rates ranging from only $3.19e^{-6}$ m/s to $2.4e^{-1}$ m/s. Therefore, the purpose of this study was to develop the compressive stiffness properties of individual lumbar intervertebral discs when subjected to various dynamic compressive loading rates.

METHODS

A total of 33 axial compression tests were performed on 11 fresh frozen human lumbar spine intervertebral discs dissected from 6 fresh frozen human cadavers, 5 male and 1 female. The cadavers ranged in age from 18 to 56, with an average age of

42. Freezing was used as a means to preserve the specimens because previous studies have indicated that freezing does not significantly affect the response of FSUs [20].

Functional spinal units (FSU), defined an intervertebral disc and the two adjacent vertebral bodies, were dissected from the cadavers. Prior to specimen preparation, lateral view digital radiographs were taken of each spine in order to identify any pre-existing degenerative changes. The intervertebral discs for each spine were graded by a certified physician on a scale of 1 to 4 based on criteria presented by Gordon *et al.* (1991). Intervertebral levels with a degenerative grade of 3 or 4 were rejected.

For comparison with the standard population, the bone mineral density (BMD) of each cadaver was determined by the Osteogram technique. The left hand of the cadavers was x-rayed, scanned and processed by CompuMed incorporated (Los Angeles, CA). This type of BMD measurement, however, only provides an indication of overall bone strength and does not account for local changes in bone density or composition. Therefore, the BMD obtained through this method is referred to as the "Global BMD". The global BMD results are reported with respect to the normal population (Table 1). The T-score is used to compare the cadaver's global BMD with that of the general population, using 30 years of age as the comparison. The Z-score is used to compare the global BMD of the subjects with the average for their age. A T-score of -1 corresponds to one standard deviation below the mean for the general population, meaning the individual is at or above the 63rd percentile for global BMD, or close to normal. T-scores of 2 and 3 correspond to 97th and 99th percentiles, respectively.

A number of detailed steps were taken in order to ensure the FSUs were rigidly secured while maintaining the proper testing orientation. After the spine was sectioned into the desired FSU, all the soft tissue except the ligaments was removed from the FSU. It should be noted that the posterior elements were left intact because previous researchers found them to have a limited effect on axial compressive stiffness under small deflections [14, 17, 22]. Second, a custom potting cup was filled with a bonding compound (Bondo Corporation, Atlanta, GA), and one half of the proximal vertebral body of the FSU was placed into the bonding compound. Special care was taken to ensure that the mid-plane of the disc was parallel with the potting cup, and that the disc was centered in the potting cup (Figure 1). This potting orientation has been used by numerous previous authors [1, 2, 7, 14, 23]. The potted vertebra was then attached to the test apparatus, and the distal potting cup was filled with the bonding compound. Finally, one half of the distal vertebral body was lowered into the distal potting cup (Figure 1). This procedure prevented any induced flexion or extension moments. After the specimen was lowered into the bonding compound, the bonding compound was allowed to fully cure before testing. The specimen was kept hydrated during the entire potting process by spraying saline directly on the specimen.

Table 1: Test matrix and subject data.

Test ID	IVD Level	Gender	Age	Body Weight	Osteogram		
			(years)	(kg)	Global BMD	t-score	z-score
IVD_1	L2-L3	M	56	81.4	105.3	-0.5	0.3
IVD_2	L2-L3	M	45	73.9	81.4	-2.7	-2.0
IVD_3	L4-L5						
IVD_4	L1-L2	F	46	115.9	93.7	-1.6	-1.6
IVD_5	L3-L4						
IVD_6	L1-L2	M	45	53.0	120.1	0.9	0.9
IVD_7	L3-L4						
IVD_8	L1-L2	M	42	85.9	92.1	-1.7	-1.3
IVD_9	L3-L4						
IVD_10	L2-L3	M	18	100.0	138.3	3.2	3.2
IVD_11	L4-L5						

Mid-plane of disc parallel to potting cup

Disc Centered in potting cup

Distal end of FSU lowered into potting cup

Figure 1: Functional spinal unit potting procedure.

The primary component of the FSU compression test setup was a hydraulic Material Testing System (MTS 810, 22 kN, Eden Prairie, MN) (Figure 2). The MTS actuator deflection was measured using the internal LVDT of the MTS. A five axis load cell (Denton, 1968, 22 kN, Rochester Hills, MI) was used to obtain the reaction force and moment, and a single axis load cell (Denton, 1210AF-5K, 22 kN, Rochester Hills, MI) was used to obtain the impactor force. Additionally, accelerometers (Endevco, 7264B, 2000 g, San Juan Capistrano, CA) were placed on both the reaction and impactor load cell plates.

Each intervertebral disc was subjected to a four part test battery in which the loading rate was increased with each test (Figure 3). First, the intervertebral disc was preconditioned to a displacement of 0.5 mm (2.5 mm ± 2.5 mm) at a rate of 1 Hz, which is similar to the frequency of normal walking, for 10 cycles. Each intervertebral disc was then preloaded to 88.96 N and subjected to two dynamic displacement steps, 0.5 mm and 1.0 mm, at rates of 0.1 m/s and 0.2 m/s respectively. For 0.1 m/s and 0.2 m/s loading rates, the data was sampled at 20 KHz and then filtered to CFC 600. Finally, each intervertebral disc was preloaded to 88.96 N and subjected to a dynamic failure test at a rate of 1.0 m/s. For 1.0 m/s loading rate, the data was sampled at 50 KHz and then filtered to CFC 600. However, the failure results are not presented in this paper. After each test, the MTS actuator was returned to the original position of zero strain and the specimen was allowed to relax for 10 minutes. The specimen was kept hydrated during the entire preparation and testing process by spraying saline directly on the specimen. Points used to calculate compressive stiffness and strain rate values were taken at approximately 25% and 50% of the loading curves. Strain was calculated based on the lateral disc height obtained form the digital X-rays.

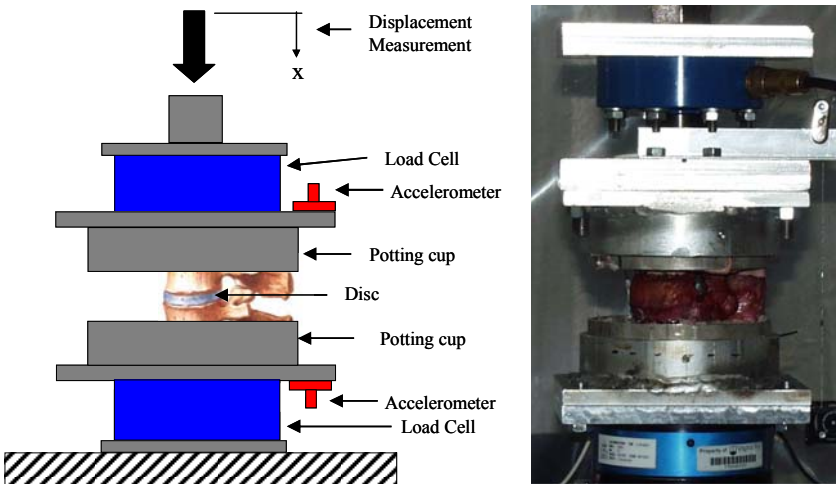

Figure 2: Individual intervertebral disc compression test setup.

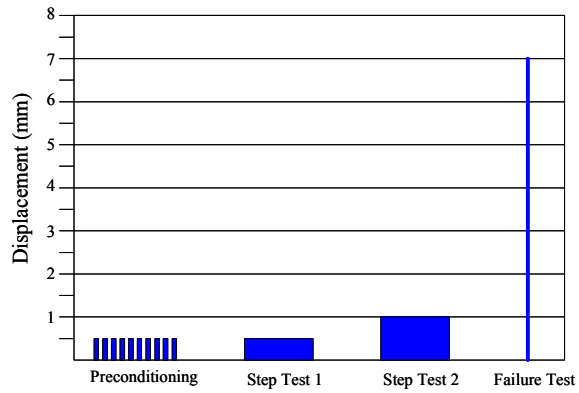

Figure 3. Individual intervertebral disc compression test battery.

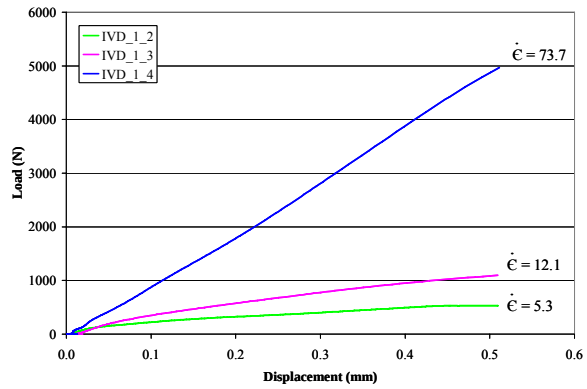

Figure 4. Compressive stiffness by loading rate. Note: Data cut at 0.51mm for tests shown.

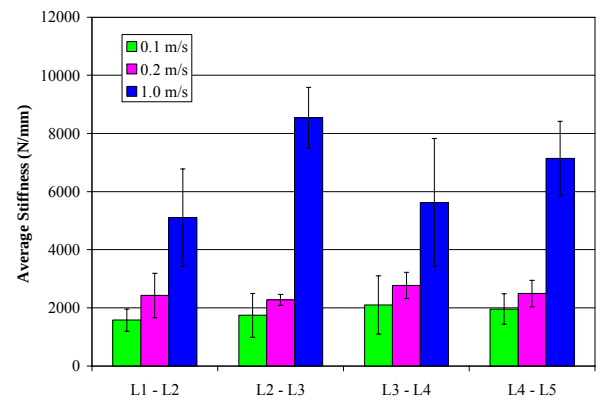

Figure 5. Functional spinal unit compressive stiffness by vertebral level.

RESULTS

The increased loading rate for each test in the three part test battery resulted in increasing compressive stiffness values for each specimen (Figure 4). In order to determine if the differences in vertebral disc compressive stiffness were significantly different with respect to loading rate, two statistical tests were performed. First, a two-tail t-test for the means, assuming unequal variances, was used to determine if the were any significant differences in compressive stiffness by vertebral level. There was no significant correlation ($p > 0.05$) between compressive stiffness and vertebral level at any loading rate (Figure 5). Therefore, all compressive stiffness data was grouped by loading rate. Then, a paired two-tail t-test for the means was used to determine if there were any significant differences in vertebral disc compressive stiffness with respect to loading rate. The statistical analysis showed that the average compressive stiffness at 0.2 m/s was significantly larger than at 0.1 m/s ($p=0.02$). In addition, the average compressive stiffness at 1.0 m/s was significantly larger than at 0.1 m/s and 0.2 m/s ($p< 0.01$).

The 0.1 m/s loading rate resulted in an average compressive stiffness and strain rate of 1835.1 ± 645.6 N/mm and 6.8 ± 1.5 s^{-1}, respectively. The 0.2 m/s loading rate resulted in an average compressive stiffness and strain rate of 2489.5 ± 474.1 N/mm and 13.5 ± 2.0 s^{-1}, respectively. The loading rate for the failure tests, 1.0 m/s, resulted in an average compressive stiffness and strain rate of 6551.1 ± 2017.0 N/mm and 72.7 ± 16.8 s^{-1}, respectively.

DISCUSSION

The results of the current study show that the compressive stiffness of lumbar intervertebral discs is dependent on the loading rate. However, the compressive stiffness at static loading rates was not determined in the current study. Therefore, the data from the current study was combined with static compressive stiffness data from previous studies. Gordon *et al.* (1991) reported an average compressive stiffness after 30 minutes of cyclic loading at 1.5 Hz (approximately 0.14 s^{-1}) to be 2453 ± 654 N/mm. Yoganandan *et al.* (1989) reported an average compressive stiffness for normal and degenerated discs compressed at 2.54 mm/s (approximately 0.22 s^{-1}) to be 2850 ± 293 N/mm and 1642 ± 447 N/mm respectively.

The initial disc heights from the current study were combined with the disc heights reported by Keller *et al.* (1987) to obtain an overall average initial disc height of 11.32 mm. The strain rate for previous studies was then calculated using the overall

average disc height and loading rate. It should be noted that Keller *et al*. (1987) did not report the loading rate. Therefore, the compressive stiffness data reported b Keller *et al*. (1987) could not be included in the curve fitting. Finally, a relationship between loading rate and vertebral disc compressive stiffness was developed by curve fitting the compressive stiffness data from the current study along with the compressive stiffness data reported by Gordon *et al*. (1991) and Yoganandan *et al*. (1989) with a linear relationship (Equation 1and Figure 6). The R^2 value for the data fit was 0.62.

$$k = 57.328\dot{\varepsilon} + 2019.1 \qquad (1)$$

This relationship is slightly lower than the linear relationship proposed by Smeathers and Jones (1988). However, Smeathers and Jones (1988) used a much larger preload, which has previously been found to affect the response of intervertebral disc [8, 18].

In order to predict the compressive stiffness of the entire lumbar spine, the compressive stiffness of each lumbar intervertebral disc was assigned the same predicted compressive stiffness value based on Equation 1 and added in series to obtain an effective compressive stiffness, k_{eff} (Equation 2). It should be noted that the vertebral bodies were assumed to be rigid.

$$k_{eff} = \frac{1}{\left[\sum_{1}^{N} \frac{1}{k_N}\right]}; \text{N=5} \qquad (2)$$

The predicted effective compressive stiffness for the lumbar spine was then compared to previously published quasi-static and dynamic compression tests performed on isolated cadaver lumbar spines, T12-L5, and the Hybrid II lumbar spine (Figure 7). The comparison shows that the predicted effective compressive stiffness at a loading rate of 0.1 m/s is slightly lower than the average compressive stiffness reported by Demetropoulos *et al.* (1998), but well within the standard deviation. Conversely, the predicted effective compressive stiffness at a loading rate of 1.0 m/s is slightly higher than the average compressive stiffness reported by Duma *et al*. (2006). However, the predicted effective compressive stiffness at a loading rate of 1.0 m/s was well within the standard deviation. Although this method does not take lumbar curvature into account, the relationship between lumbar intervertebral disc compressive stiffness and loading rate presented in the current paper provides reasonable effective compressive stiffness of the whole lumbar spine over a range of loading rates.

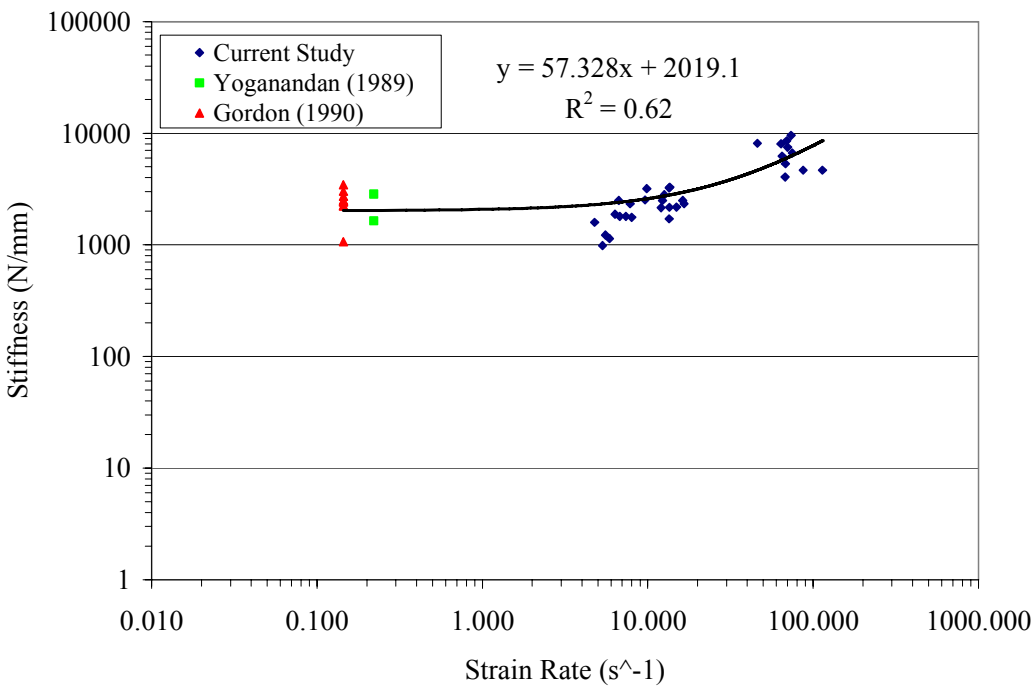

Figure 6. Relationship of intervertebral disc compressive stiffness to strain rate. (Note: log-log scale)

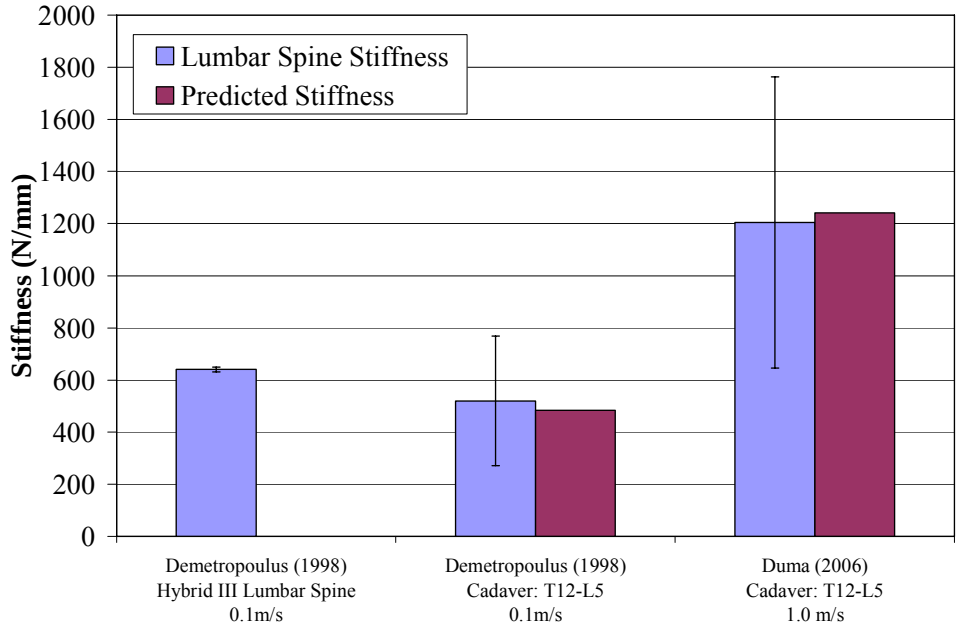

Figure 7. Comparison of predicted values to whole lumbar spine testing.

CONCLUSIONS

The compressive stiffness properties for the individual lumbar intervertebral discs were determined at three dynamic loading rates using a high rate servo-hydraulic material testing machine. The results showed that there was no significant correlation ($p > 0.05$) between compressive stiffness and vertebral level at any loading rate. In addition, the compressive stiffness of lumbar intervertebral discs in axial compression was found to dependent on the loading rate. Therefore, a relationship between loading rate and vertebral disc compressive stiffness was developed by curve fitting the stiffness data from the current study along with static compressive stiffness data reported by previous studies with a linear relationship.

The lumbar FSU research presented in this study will provide useful information for the development and validation of both mathematical and mechanical models of the human lumbar spine. However, in order to fully model the lumbar spine, additional testing will need to be conducted to quantify the effects of loading rate on stiffness in tension, shear, and bending.

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