

Determination of vertebral endplate deformation under load using micro-computed tomography

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Abstract

Endplate strength plays an important role in preventing vertebral failure of normal vertebrae and in cases where surgical intervention has replaced the disc with an implant or has altered the vertebral loading. We have developed a non-contact method based on the principles of image guided failure analysis, mechanical testing, and micro-computed tomography analysis, which allows for *in vitro* quantification of endplate deformation under axial load. The method allows for the implementation of a repeated measures experimental design, each specimen acting as its own control. Our methodology was validated using cadaveric functional spine units, loaded stepwise from 200 N to a maximum of 2000 N. The loading protocol was repeated over two days, allowing time for recovery of the disc mechanical properties. We found no meaningful difference in measured force, stiffness, and endplate deformation between day 1 and day 2. The mean fiducial registration error was less than 0.015 mm for all three axes. Endplate deformation could be reproducibly estimated. The root mean squared error was 0.03 mm, which is the effective precision of the method. Using this micro-CT based method, the effect of interbody implants, grafts, disc replacement strategies, and surgical procedures such as nucleotomy and vertebral cement augmentation on endplate mechanical behaviour can be ascertained.

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1. Introduction

Deformation of the endplates and the vertebral body are hypothesized to play an important role in minimizing peak impact loads and reducing strain on intervertebral disc (IVD) annular fibers. Endplate strength ensures mechanical stability and prevents implant subsidence for interbody fusion implants (Polikeit et al., 2003a, b; McAfee, 1999) and disc prosthesis (Goel et al., 2005), during vertebral body resection or total disc replacement. In addition, endplate mechanics are implicated in the development of vertebral fractures after surgical intervention (Baroud et al., 2003; Frei et al., 2001; Polikeit et al., 2003c; Keller et al., 2005; Kayanja et al., 2006) or a traumatic event (Hongo et al., 1999).

Spinal loads are transferred from one vertebral body to the next through the IVD, which acts on the vertebral endplate. Between the superior and inferior endplate, load is carried by two main pathways, the trabecular core and the dense trabecular shell (White and Panjabi, 1990). Adjacent to the superior and inferior endplate, almost the entire load fraction is carried by the supporting trabecular bone (Banse et al., 2001; Homminga et al., 2004; Ebbesen et al., 1999; Eswaran et al., 2006). Trabecular bone can only withstand small strains on the order of 1–7% (Keaveny et al., 1994), therefore, excessive bulging of the endplate will result in trabecular failure or microfractures (Brinckmann et al., 1983; Holmes et al., 1993). Changes to the vertebral loading (flexion/extension, axial compression, lateral bending), the magnitude of the applied load or changes to vertebral stiffness due to the addition of an implant or surgical procedure will be reflected in changes in endplate strain. With an ever increasing variety of surgical methods now available to correct disc and spinal

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deficiencies, there is a need to evaluate outcomes on the mechanical behaviour of not only the affected vertebra but on adjacent vertebrae as well. Understanding endplate mechanics will not only help determine short term prognosis following surgery but will help to predict long term local bone remodeling effects due to either stress shielding or increased trabecular loading.

While *in vitro* methods exist to study endplate mechanics by means of metal indenters (Grant et al., 2001; Ochia et al., 2003), pressure film located below the endplate (van Dieen et al., 2001) and linear variable differential transformers (LVDTs) or small stainless steel spheres implanted near the endplate surface (Holmes et al., 1993; Brinckmann et al., 1983), no method exists to determine endplate mechanics in an intact specimen under physiologically relevant loading conditions. Endplate mechanics derived from implanted markers yield only a point measure of endplate deformation and disrupt the 3D trabecular network, thus possibly overestimating endplate deformation (Bayraktar et al., 2004). Analysis of vertebral mechanics has also been performed using finite element models (Bayraktar et al., 2004; Van Rietbergen et al., 1995; Homminga et al., 2004; Baroud et al., 2003; Eswaran et al., 2006; Villarraga et al., 2005; Crawford et al., 2004; Polikeit et al., 2003b, c). However, finite element models must be validated using appropriate experimental methods. In addition, the fine micro-architecture of the vertebra, and the behaviour of the IVD, may not be adequately accounted for in continuum models. Advances in micro-CT have recently made it possible to determine 3D assessment of loading within bone specimens (Müller and Rügsegger, 1997; Müller et al., 2002; Nazarian and Müller, 2004; Müller et al., 1998). Micro-CT has the advantage of non-destructively providing a detailed assessment of vertebral bone architecture morphological parameters and structural characteristics which can be used to assess form and function.

The aim of this study was to develop and test a non-invasive technique to determine *in situ* endplate deformation of a functional spine unit subjected to axial compression. The technique must be suitable for a repeated measures experimental design to facilitate the investigation of altered loading due to surgical intervention, postural changes or degenerative effects on endplate deformation.

2. Methods

2.1. Sample preparation

Eight osteoporotic cadaveric human functional spine units (FSU) were obtained (median age 76 yrs, min 70 yrs, max 80 yrs) from the following levels T11–T12 ($N=1$), T12–L1 ($N=1$), L1–L2 ($N=1$), L2–L3 ($N=2$), L3–L4 ($N=3$). Previous investigators have found little difference in bone volume/total volume between levels T9 to L5 (1%, T9 to L5, 0.5% T11–L5), therefore we pooled all vertebrae together for analysis (Grothe et al., 1995; Stauber and Müller, 2006).

All surrounding soft tissue and posterior elements were removed from specimens and impressions were made of the cranial and caudal endplates in semi-cured bone cement (Sulfix, Sulzer Orthopaedics Ltd, Switzerland). A jig ensured that the two end caps were parallel to each other. The

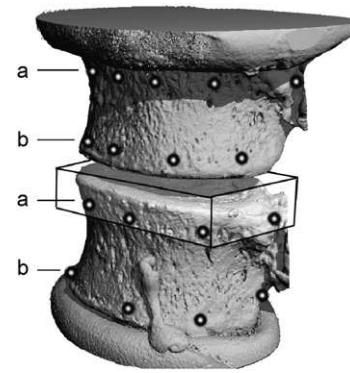


Fig. 1. A function spine unit (FSU) illustrating possible fiducial marker locations (glass beads): (a) denotes the superior marker set and (b) the inferior marker set. The box represents the region of interest (ROI) selected when determining endplate surface heights and ultimately endplate deformation.

moulded end caps extended only to the cortical rim, thereby not contributing any structural strength to the FSU. A minimum of five small glass beads (~ 1 mm) were affixed with cyanoacrylate glue directly to the cortical shell just below each superior and inferior vertebral endplate to provide fiducial reference points (Fig. 1).

2.2. Loading device

Functional spine units were loaded in a custom micro-compression device (Fig. 2). The device allowed for the axial compressive loading of a FSU within the bore of the micro-CT scanner (XtremeCT, Scanco Medical AG, Bassersdorf, Switzerland). Dimensionally, the length and width of the specimen cylinder were sufficient to accommodate any lumbar spinal segment. The cylinder (Ertacetal C, POM, Angst+Pfister AG, Zürich) was designed to be radiolucent and stiff, allowing X-ray penetration and ensuring dimensional stability under load. The FSU was compressed in a controlled stepwise manner using a loading screw (0.5 mm pitch). The rotational motion of the screw was decoupled using a bearing to apply only compressive load to the specimen. Load applied to the specimen and displacement of the loading platen were monitored using a miniature load cell (0–1000 lb, Honeywell, Model 31, Columbus, US) and a LVDT (Burster model 8740-5005, Gernsbach, Germany), respectively. The entire cylinder was sealed ensuring that the specimen would be immersed in physiological saline solution throughout the experimental protocol.

2.3. Specimen loading

The loading protocol is shown in Fig. 3. Loads were chosen to represent rest (200 N), *in vivo* standing load (1000 N) (Villarraga et al., 2005), and light manual work (1500–2000 N depending upon the size of the vertebra, to keep the applied compressive stress similar between specimen) (Nachemson, 1981). Two scanning protocols were used: “static” and “quick”. The first protocol (“static”) incorporated a 45 min pre-load phase at the target load, which was followed by a scan of the entire FSU. The 45 min static loading phase was performed to minimize stress relaxation during scanning (Adams et al., 1990, 1996; van Dieen et al., 2001). The second protocol (“quick”) did not contain a pre-load phase. Instead, scans were immediately preformed after reaching the target load. The region of interest (ROI) for the “quick” scans was limited to only the endplates, resulting in reduced scan times, limiting the stress relaxation of the specimen. Each endplate was a separate scan allowing the load to be adjusted upto the target load prior to scanning. The entire loading protocol was repeated the next day after 14 h of rehydration within a saline bath with no applied compressive load, allowing the disc to

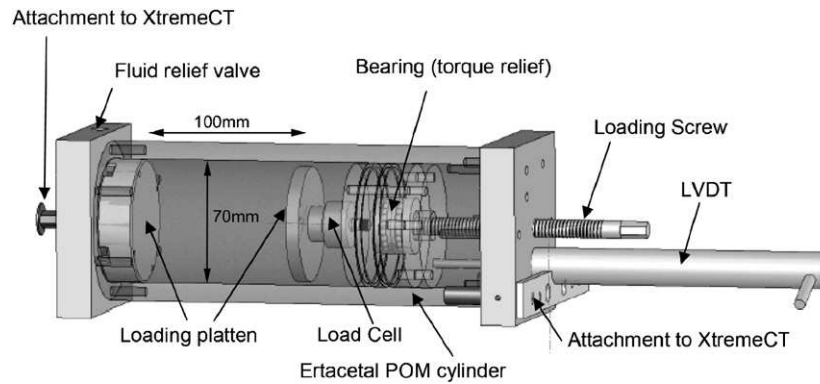


Fig. 2. Loading device to apply axial compressive loads to the FSU within the micro-CT bore. The device was composed of a radiolucent cylinder within which the specimen was located. The load was applied using a screw, and the applied force and displacement were monitored by a load cell and LVDT.

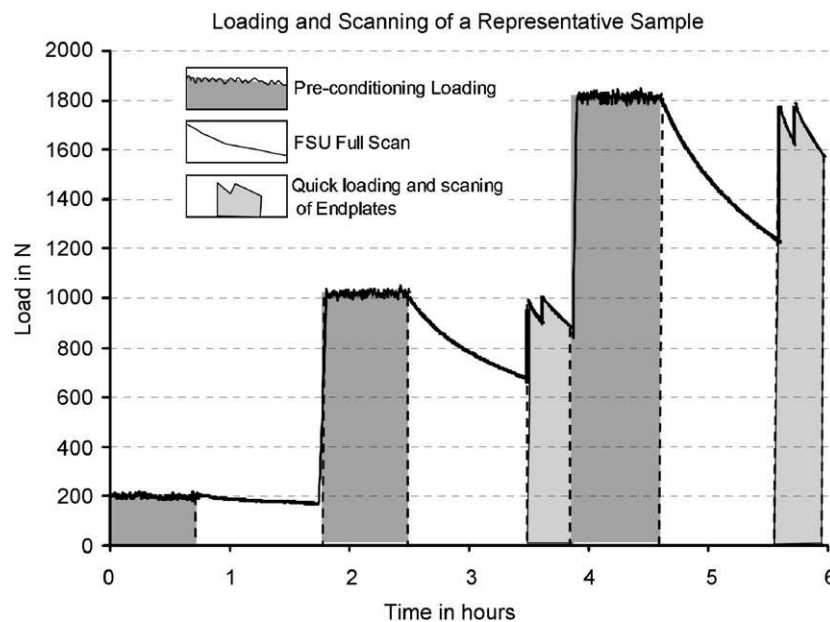


Fig. 3. Loading protocol for the FSU. Loads were chosen to reflect rest (200 N), standing (1000 N) and light manual work (1500–2000 N depending on vertebra size). Two scans were performed; a full FSU scan (approx. 60 min, “static”) and a quick scan of the endplates alone (approx. 8 min, “quick”). Prior to the full FSU scan the specimen was pre-conditioned for 45 min to reduce stress relaxation during scanning.

recover height and mechanical properties (Johannessen et al., 2004). Differences between FSU mechanical behaviour on day1 and day2 were determined by comparing the stress relaxation curves during the micro-CT scans and the stiffness of the FSU, determined from a linear regression of the linear loading portion of the load versus displacement curve.

2.4. Micro-CT scanning

All measurements were performed on a micro-computed tomography system (XtremeCT, Scanco Medical AG, Bassersdorf, Switzerland). The number of slices ranged between 770 and 990 with a nominal isotropic resolution of 82 μm (field of view 125 mm, 1536 $\times$ 1536 pixels, integration time 399 ms). Total scan time per specimen approached 60 min, for scans of an entire FSU, and 8–16 min for the endplates alone.

2.5. Image registration and transformation

Registration and subsequent transformation were performed to ensure that for each loading step, rigid body rotations and translations of the

vertebral bodies, and deformation of the vertebral bodies between the inferior and superior fiducial marker sets were not included when determining the location of the endplate surface from the 3D image data (Fig. 1). ROI from the 3D image data were identified to analyse endplate deformation. The ROI encompassed the endplate including its associated fiducial markers (glass beads). Within each ROI, fiducial markers were segmented (threshold 20% maximum gray value) and uniquely identified using component labelling and subsequent bead size analysis (Image Processing Language, v4.29d, Scanco Medical AG, Bassersdorf, Switzerland). Subsequently, the centroid of each bead was determined (custom C program) and a rigid point-based registration was applied using the Orthogonal Procrustes method with those centroids as input (Matlab, MathWorks Inc, Natick, MA) (Fitzpatrick et al., 2000; Arun et al., 1987). All images were registered to the baseline loading case representing rest (200 N). For each load step the resulting rigid body transformation (three rotations and three translations) was applied to the original (unsegmented) image data and re-sampled using a linear interpolator (Boyd et al., 2006). Fiducial registration error was estimated for the resulting image by comparing the root mean squared (RMS) error of the fiducial marker locations.

The deformation of the vertebral bodies could be determined by monitoring fiducial markers attached near the inferior and superior endplates. Thus the total endplate deformation was decoupled into vertebral body deformation and endplate deflection. Endplate deformation reported herein combines both vertebral and endplate deflection.

2.6. Estimation of the endplate deflection

The transformed images of the endplate ROI were segmented by Gaussian filtering (sigma 1.2, support 1) and thresholded (4.75% of maximum gray value) to extract the mineralized phase of the endplate and underlying trabecular bone. Endplate deflection was assessed by measuring the relative difference in position of the surfaces of two aligned images (i.e., 200 and 1000 N loaded specimens) and results are expressed relative to the baseline image (i.e., 200 N loaded specimen). A knowledge of the organization of the tissue phases at the endplate surface allowed for the boundary to be detected with subvoxel precision (Hwang and Wehrli, 2002). Intermediate attenuation values located along the boundary represent partial volume effects (voxel values represent the combined contribution of the bone and the IVD material). The attenuation value of bone, IVD material and the structural organization were known allowing for the subvoxel determination of the boundary using 1D linear interpolation (custom code written in C, Visual Studio 6.0, Microsoft; and Matlab, MathWorks Inc. Natick, MA). The approximation of endplate surface height, and hence deflection relative to the baseline image, resulted a spatial map of deflections that could be represented as scalar values superimposed on the endplate surface (Fig. 4) at each load.

The point of maximum deflection was located on each endplate surface for each load step by searching for a local maximum. The search space was confined to a user-defined region on the endplate to avoid any errors caused by inclusion of image artifacts that can occur at the margins of the endplate (semi-automatic Matlab routine, MathWorks Inc, Natick, MA). The same confined search region was applied consistently to all load steps for each specimen, and the position of the local maximum was tracked to monitor if the region of maximum deflection shifted during the experimental protocol.

Endplate deformation is reported for four loading cases, “min static” and “max static” (1000 N and maximum load for cases in which 45 min preconditioning and full FSU scans were performed) and “min quick” and “max quick” (1000 N and maximum load for cases in which there was no preconditioning and scans only encompassed the endplate).

2.7. Statistics

Differences between endplate deformation on day1 and day2, loading force and FSU stiffness were evaluated using repeated measures ANOVA. Increases in endplate deflection with a change in load were investigated using a one-way ANOVA and a Neuman–Keuls post hoc test (Statistica 7, StatSoft, Tulsa). A significance value of $p = 0.05$ was defined.

3. Results

The mechanical behaviour of the FSU was similar on day1 and day2. The amount of stress relaxation was consistent between the two days (Fig. 5). The force on day2 was always greater than the force on day1 (repeated measures ANOVA, $p < 0.05$). However, the magnitude of the percentage difference between day1 and day2 was minimal ($3.6\% \pm 2.3\%$, $n = 48$). The stiffness of the IVD remained consistent between testing on day1 and day2 (1000 N: day1 $1184 \text{ N/mm} \pm 220 \text{ N/mm}$, day2 $1277 \text{ N/mm} \pm 117 \text{ N/mm}$, $p = 0.52$; max load: day1 $2800 \text{ N/mm} \pm 438 \text{ N/mm}$, day2 $2648 \text{ N/mm} \pm 275 \text{ N/mm}$, $p = 0.3$).

Mean fiducial registration error was $0.013 \text{ mm} \pm 0.006 \text{ mm}$, $0.012 \text{ mm} \pm 0.007 \text{ mm}$, $0.010 \text{ mm} \pm 0.006 \text{ mm}$ for the x , y and z -axes, respectively. Deformation of the cortical shell can be calculated as a change in position of the cranial markers relative to the caudal markers or vice versa on each vertebra. Small deformation of the cortical shell was noted at maximum load (mean deformation $0.018 \text{ mm} \pm 0.011 \text{ mm}$, max 0.04 mm , $n = 5$).

Endplate deformation response to compressive load varied considerably between specimens. In general, an increase in endplate deformation was noted with an increase in applied load (Figs. 6, 8). The difference between day1 and day2 endplate deformation defined by the RMS error (Gluer et al., 1995) was 0.030 mm . There was little

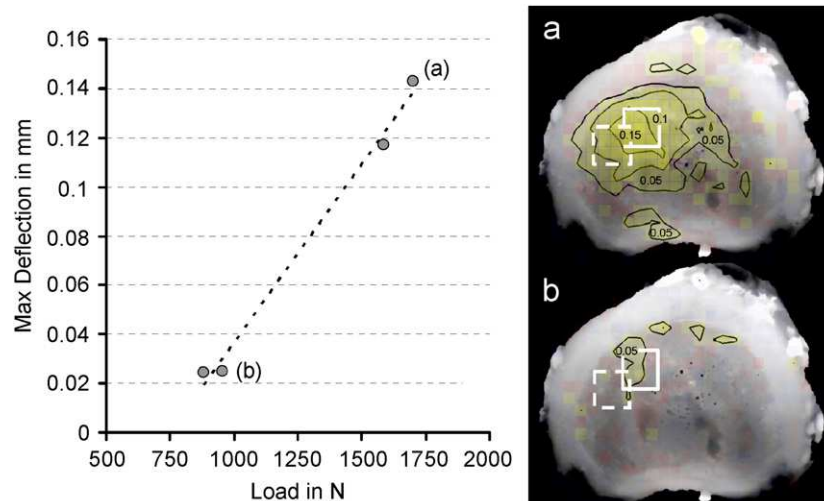


Fig. 4. Day1 endplate deformation of a cranial inferior endplate. Images on the right hand side show the endplate deformation mapped over the endplate surface. Initially there is very little endplate deformation (b). However, upon application of more compressive load the endplate deformation is evident in the central left region (a). Shaded, yellow squares indicate greater endplate deformation. Endplate deformation contours are shown for 0.05, 0.1 and 0.15 mm. Solid and dashed squares show the regions of current maximum deformation for day2 and the location of the previous day1 maximum endplate deformations, respectively.

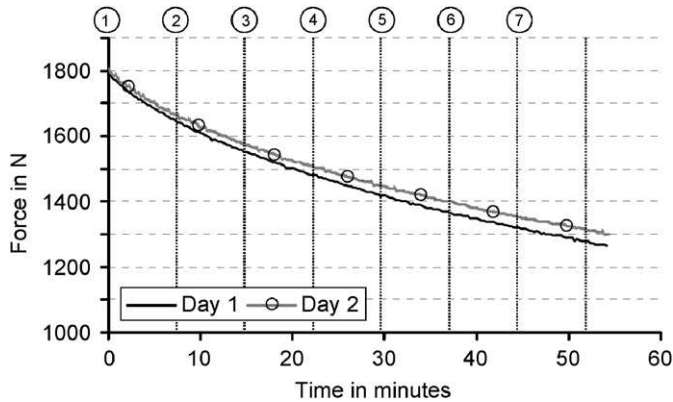


Fig. 5. Representative example of differences in load relaxation profile between day1 and day2. There is a small, but significant, difference in force profiles between the two days, indicating a change in disc mechanical behaviour. Numerals on the top of the graph indicate the start of each 360° rotation in which data for 110 slices, once reconstructed, are collected. The reported force at a location within the FSU is the average force recorded during the start and end of a rotation which contains the required data.

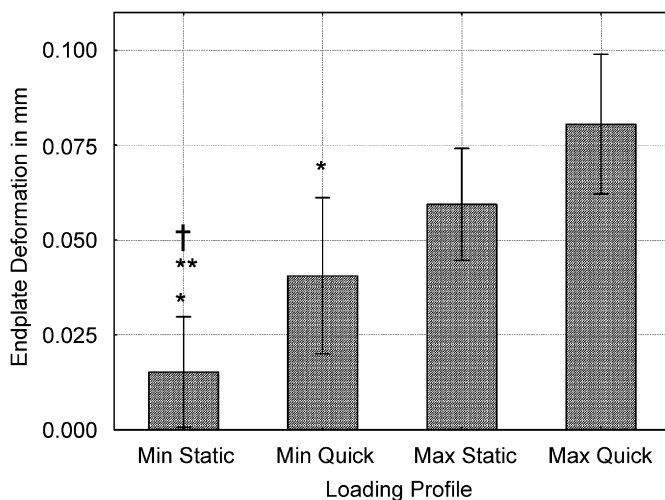


Fig. 6. Increase in endplate deformation for the four loading stages. Day1 and day2, superior and inferior endplate deformation has been combined. *, **, † indicates statistically significant changes in endplate deformation ($p < 0.05$) relative to Max Quick, Max Static, and Min Quick, respectively. Min and Max refer to the 1000 N and 1500–2000 N loading step. The mean stress applied at each loading step: (1) 0.52 ± 0.08 MPa ($n = 20$); (2) 0.6 ± 0.1 MPa ($n = 31$); (3) 0.96 ± 0.18 MPa ($n = 16$); (4) 1.03 ± 0.17 MPa ($n = 32$).

variation in RMS error when considering each loading situation (RMS error for max quick 0.031 mm, max static 0.032 mm, min quick 0.028 mm, min static 0.027 mm). Only one loading condition, max static, had a small but significant difference between its loading on day1 and day2 (Table 1, Fig. 7).

4. Discussion

The use of micro-CT is a unique approach to investigate the 3D progression of endplate deformation under

increasing load. Deformation and failure of local bone has been determined previously using micro-CT based image guided assessment (Müller et al., 1998; Nazarian and Müller, 2004). We extended this technique to endplate deformation. Unlike previously reported methods to measure endplate deformation, it does not compromise the structural integrity of the FSU by invasive procedures, thus, preserving the structure of the vertebrae (Brinckmann et al., 1983; van Dieen et al., 2001; Holmes et al., 1993). The inclusion of the IVD into the test construct allows the investigation of endplate mechanics in a manner which more closely simulates the *in vivo* conditions (Ochia et al., 2003; Grant et al., 2001). Although experimental testing using a FSU is complicated by the inclusion of the IVD, due to the time sensitive viscoelastic behaviour of the IVD and its change in response due to changes in disc health (van Dieen et al., 2001; McMillan et al., 1996a, b; Pollintine et al., 2004a, b), the methods proposed here can control for these confounding factors (e.g., consistent loading rates, identification of degree of disc degeneration and bone quality). Consistent with the findings of others, endplate properties and behaviour vary significantly between specimens, requiring large numbers of matched samples to perform meaningful investigations (Grant et al., 2001). The major advantage of our method to assess endplate deformation using micro-CT is the ability to perform a repeated-measures experimental design. Furthermore, the effect of interbody implants or grafts, surgical procedures such as nucleotomy, vertebroplasty, kyphoplasty or disc replacement on endplate deformation can be ascertained by this method. The ability to perform repeated-measures experiments increases statistical power while decreasing the number of specimens required.

A limitation of micro-CT based mechanical studies is that they preclude dynamic analysis due to the length of the involved scan times. Two loading/scanning protocols were performed, “static” and “quick”. Sustained loading, as experienced during the “static” protocol, changes the stress patterns within the disc, resulting in an expression of water from the IVD, a drop in nucleus pulposus pressure and an increase in stress in the annulus fibrosus, culminating in a more uniform stress distribution over the endplate (van Dieen et al., 2001; McMillan et al., 1996a, b). The “quick” protocol minimized stress relaxation and load redistribution within the disc by minimizing scan times (8 min) and through performing the scans directly after reaching the target load. Large changes in endplate behaviour were noted between these two loading/scanning protocols, although it was not noted for all specimens (8 of 10 endplates with both loading/scanning protocols). It was observed in some instances that the endplate deflection under “static” loading was lower than a corresponding “quick” loading scheme at lower loads, possibly due to the afore mentioned redistribution of load to the periphery of the endplate and the cortical shell as a result of the sustained pre-conditioning load (for “static” loading only).

Table 1
Average deformation of the endplates at the four loading steps

Series	Loading	Endplates (No.)	Applied stress (MPa)	Endplate deformation in (mm)			p
				Combined	Cranial–inferior	Caudal–superior	
Day 1	Max Quick	10	1.02±0.18	0.084±0.045	0.074±0.057	0.093±0.034	
	Max Static	15	0.94±0.18	0.069±0.055	0.061±0.067	0.079±0.042	
	Min Quick	8	0.61±0.10	0.048±0.032	0.044±0.038	0.053±0.031	
	Min Static	16	0.51±0.08	0.022±0.031	0.015±0.030	0.028±0.028	
Day 2	Max Quick	10	1.05±0.17	0.077±0.051	0.058±0.048	0.097±0.051	0.41
	Max Static	15	0.98±0.18	0.051±0.053	0.04±0.058	0.062±0.048	0.02*
	Min Quick	8	0.62±0.10	0.033±0.026	0.02±0.028	0.046±0.020	0.12
	Min Static	16	0.53±0.09	0.009±0.022	−2E−04±0.020	0.020±0.018	0.22
Local Max	Max Quick	10	1.05±0.17	0.092±0.050	0.072±0.053	0.113±0.042	0.26
	Max Static	15	0.98±0.18	0.071±0.071	0.05±0.066	0.095±0.075	0.83
	Min Quick	8	0.62±0.10	0.041±0.023	0.03±0.020	0.052±0.023	0.37
	Min Static	16	0.53±0.09	0.017±0.025	0.004±0.030	0.029±0.016	0.53

Significant differences between the endplate deformation on day1 and day2 are noted with “*” ($p < 0.05$). Min refers to 1000 N loading step and Max refers to 1500–2000 N loading step.

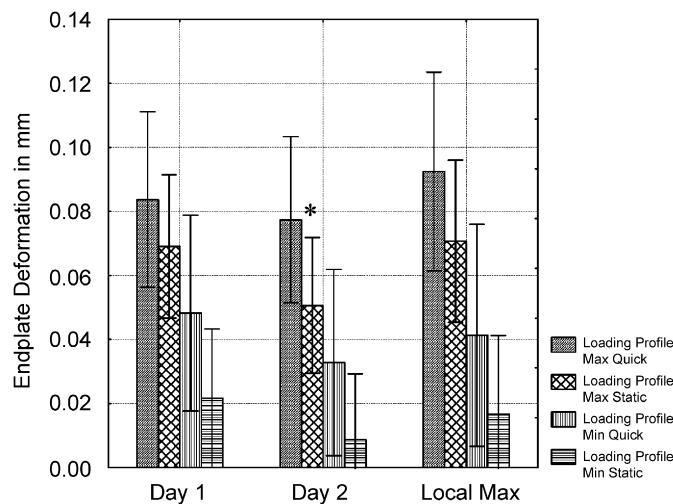


Fig. 7. Combined superior and inferior endplate deformation at the four loading states. Only “max static” loading significantly differed between the deformation on day2 and the other loading states. Also a change in magnitude of the local maximum endplate deflection was not detected from day1 to day2.

The “quick” loading protocol is probably more illustrative of endplate response to changes in vertebral loading *in vivo*. Future studies on endplate deformation could benefit by minimizing scan times by measuring the endplate alone to limit stress relaxation, redistribution of load towards the periphery of the endplate and to expedite the experimental procedure.

Loading, in the current experimental series, was limited to that which may occur during normal daily activities *in vivo*. The importance of endplate deformation is not necessarily realized at low loads. At low loads, the compliance of the FSU is mainly achieved through the medial or sagittal bulging of the IVD. However, at higher loads there is less possibility of bulging of the annulus and

compliance is realized through the deformation of the endplates (Biggemann et al., 1988; Brinckmann et al., 1983).

Increased loads will result in increased stress relaxation and creep rates. However, we have demonstrated that a strict adherence to experimental protocol will ensure similar loading conditions are experienced by the endplate on both days in which it is loaded. Similar test conditions are required to ensure that changes in endplate behaviour can be attributed to the treatment and not the experimental protocol. Stress relaxation was always reduced on day2, even after 14 h of recovery. Changes in IVD mechanical behaviour may be due to non-recoverable deformations and a loss of IVD swelling pressure (Maclean et al., 2006; Urban and Maroudas, 1981). However, the overall stiffness of the FSU did not change between testing days and the increase in endplate loading, although statistically significant, was minor ($3.6\% \pm 2.3\%$).

Registration and transformation of micro-CT image arrays is required to ensure that the observed endplate deformation is a result of loading and not a result of rigid body rotations or translation of the specimen (Boyd et al., 2006). We reported the changes in endplate height between day1 and day2 which are primarily due to the experimental protocol which altered the mechanical behaviour of the FSU, although registration inaccuracies could be a confounder (Fig. 8). The importance of locating the fiducial markers near the investigated endplate was noted during the experimental procedure. Three specimens had markers located near the endplate furthest from the endplate abutting the disc (i.e., for the superior endplate which abuts the disc, the markers were located near the inferior endplate on the same vertebra). Although we assumed that cortical shell deformation would be considerably less than endplate deformation, under compressive loading there may be involvement of the cortical shell

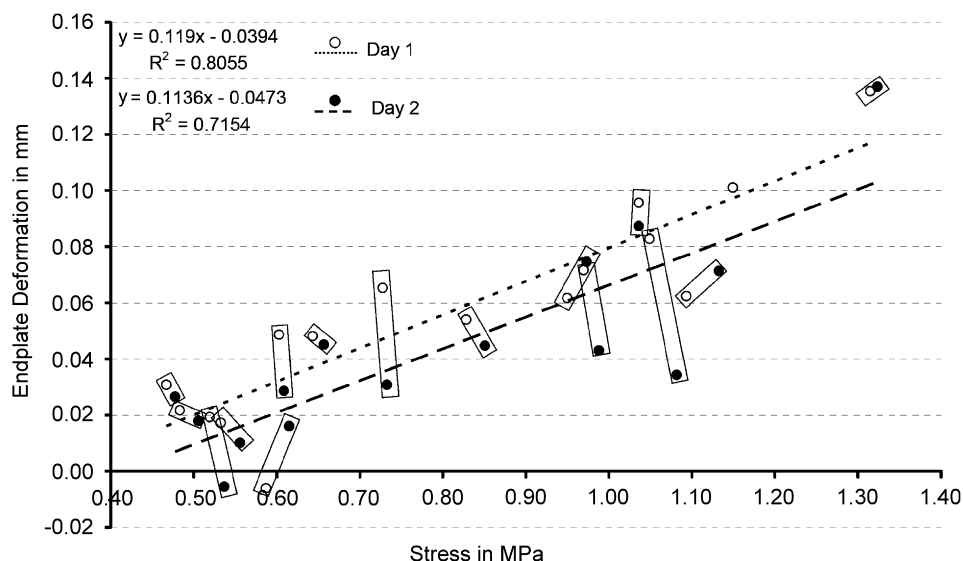


Fig. 8. Average superior and inferior endplate deformation on day1 and day2. Data shown is for the “quick” loading protocol. Boxes group specimen pairs. Pearson’s correlation coefficient shows a linear trend between endplate deformation and applied compressive stress.

(Eswaran et al., 2006; Kayanja et al., 2006). Decoupling cortical shell and endplate deformation is not possible with only one marker set (>3 beads) present on a vertebra.

Measurement resolution of endplate deflection is clearly dependent on the image resolution of the micro-CT (82 μm nominal isotropic resolution). Partial volume effects were minimized during image transformation and during endplate surface identification through voxel interpolation of the gray-scale attenuation data. When performing analysis on objects which require a large field of view and scan length, there is a trade off between detector resolution, acquisition speed, and the ability to work effectively with the generated data set. Obviously one would expect increased precision with increased resolution. However, the goal of this study was not to resolve trabecular micro-architecture, but rather to determine the boundary between the dense endplate surface and the radiolucent IVD, for such a situation using an interpolative technique is an effective means of achieving subvoxel precision. Caution must be used interpreting endplate deformations below the resolution of the scanner, since while an approximation of endplate deformation can be determined reproducibly the absolute accuracy is limited.

We have demonstrated the feasibility of using a novel micro-CT method to directly determine endplate mechanics. This combined approach using the principles of image guided failure analysis (Müller et al., 2002; Nazarian and Müller, 2004), mechanical testing and micro-CT, allows not only the determination of endplate deformation, but the non-destructive nature of micro-CT lends itself to more comprehensive study designs when combined with biochemical assays (e.g., proteoglycan or collagen content of the disc) to provide a complete analysis of vertebral loading mechanics.

Conflict of interest

The authors have no conflict of interest.

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