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BIOMECHANICS OF OSTEOPOROTIC CRUSH FRACTURES USING SYNTHETIC VERTEBRAE

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Abstract: An estimated 30-50% of women and 20-30% of men will develop vertebral fractures in their lifetime, 85% of which are associated with osteoporosis. Bone loss in osteoporosis leads to reduced stiffness and strength of vertebral cancellous bone, and eventual loss of cortical wall thickness. This study developed synthetic vertebra models for investigation of vertebrae mechanics in osteoporosis - specifically the effects of bone density and micro-architecture on vertebral body strength and stiffness. Rapid prototyping techniques were used to create models with dimensions based on human lumbar vertebrae. A three dimensional matrix of longitudinal and transverse struts was used to simulate trabecular bone. Matrix dimensions were varied to represent differing degrees of osteoporosis. Cortical wall thickness was also reduced to simulate serverely osteoporotic vertebrae. Foam and silicone mouldings were used as intervertebral disk phantoms. The synthetic vertebrae were tested in uniaxial compression at constant strain rate (5mm/min). Force and displacement were logged until specimen failure and video was used to record gross vertebral fracture patterns. Reduction in trabecular bone volume fraction from 19% to 10% resulted in 65% decrease in strength and 70% decrease in stiffness, while a 25% reduction in cortical wall thickness reduced the strength of the vertebrae by 58%. The large decreases in vertebral stiffness and strength with bone loss are comparable to those found in previous studies. Reducing cortical wall thickness did not affect vertebral stiffness greatly, but this was most likely due to limitations in the dimensional accuracy of the rapid prototyping technology. Overall, the synthetic vertebra models provide a repeatable and biomechanically useful tool for the study of vertebral compression fracture mechanics.

Introduction

The most common musculoskeletal disorders in the elderly are osteoporosis and osteoarthritis [1]. Osteoporosis is a condition characterized by low bone mass and micro-architectural deterioration of bone tissue, with a consequent increase in bone fragility and susceptibility to fracture [2]. In osteoporosis, bone resorption outpaces bone deposit resulting in normal bone quality, but reduced bone density. Micro-architectural deterioration in osteoporotic trabecular bone includes a reduction in both the thickness of longitudinal trabecular struts and in the number of transverse struts [3]. While osteoporosis affects the entire skeleton, most osteoporotic fractures occur in the vertebrae [4, 5]. Vertebral compression fractures generally involve only the anterior column of the spine [6].

A number of previous studies have investigated the compressive strength of both healthy and osteoporotic vertebrae [5, 7-9]. It is known that the load-bearing capacity of the vertebral body decreases with age and depends on vertebral body geometry, cortical shell integrity and thickness, and cancellous bone apparent density [10]. However the loss of strength due to reduced cortical wall thickness has not been thoroughly investigated. Several studies have also quantified the effects of trabecular bone density on vertebrae strength [4, 7, 8] but much less attention has been given to the effects of trabecular micro-architecture on vertebral strength.

Rapid prototyping is a technique used for producing geometrically accurate, repeatable synthetic models from three-dimensional computational models. This technology allows precise specification of model dimensions and avoids large variations between specimens which often hinder biological studies.

The aim of this research was to create synthetic, biomechanically analogous human vertebrae with differing degrees of osteoporosis and to use the synthetic vertebrae to investigate the effects of cancellous density and micro architecture as well as cortical wall thickness on vertebral stiffness, strength and fracture mechanisms.

Materials and Methods

L1 Vertebra Model

A three dimensional solid model of an L1 human lumbar vertebra was created using SolidWorks 2005 (SolidWorks Corporation, USA). Dimensions were based on data from Panjabi et al. [11]. To simplify the model, approximations were made where there were minor differences in dimensions (e.g endplate depths for upper and lower vertebral endplates were given as 41.2mm and 43.3 mm respectively. This was approximated to 42.3mm.) The posterior elements of the vertebrae were not included in the model as they play a minor role in axial compressive loading [9]. The pedicles also served as exit holes for removal of excess material in the rapid prototyping process. The model was shelled to a thickness of 0.64mm [12] to produce the cortical shell (Figure 1).

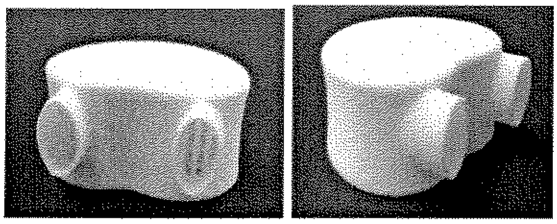

Figure 1: Vertebra model produced using rapid prototyping.

To create a biomechanically representative cancellous core, a three dimensional matrix was created inside the vertebral cortex. The matrix consisted of an array of longitudinal and transverse struts, and the architecture was varied to simulate differing degrees of osteoporosis. For the healthy vertebra, a transverse trabecular strut diameter of 0.7mm was used, with a displacement of 2.33mm between adjacent struts [13]. Longitudinal trabecular struts had a diameter of 0.9mm with a displacement of 3.5mm between adjacent struts [13]. This gave an overall bone volume to total volume ratio (BV/TV) of 0.19. As osteoporosis results in a reduction of longitudinal trabeculae thickness and a loss of transverse ties [3], the diameter of the vertical struts were varied and transverse struts were removed to achieve a lower BV/TV. For mild osteoporosis, the diameter of the longitudinal struts was reduced to 0.8mm and one third of the transverse struts were removed to give a BV/TV of 0.10. For severe osteoporosis, longitudinal strut diameter was further reduced to 0.7mm and two thirds of the transverse struts were removed to produce a BV/TV of 0.05. The end result was three models of varying trabecular density and architecture.

In the later stages of osteoporosis, the cortical shell also begins to thin. To simulate this, the stage 2 osteoporotic matrix was used and the cortical shell was thinned from 0.64 to values of 0.50mm and then to 0.36mm. This resulted in three different models of varying wall thickness (all using the severe osteoporotic trabecular density.) Table 1 gives details of the various models used.

Rapid Prototyping

A number of rapid prototyping techniques were considered for the manufacture of the synthetic vertebrae. Suitable mechanical properties for the available materials and the dimensional accuracy of the technique were both important requirements. Stereolithography was investigated as it utilised materials with considerably stronger and more brittle material properties than any other technique. However this technique was unable to produce such a small, complex model. The rapid prototyping technique eventually employed was selective laser sintering (SLS). The material used was a glass filled fine polyamide (PA 3200 GF EOSINT P) with tensile modulus of 3200 ± 200 MPa and elongation at break of 6 ± 3%. While these properties are less stiff than bone tissue, they were the strongest and most brittle material available for the SLS technique. Three copies of each vertebra model were produced. Once the models were made, they were each weighed and wall thickness measured to determine the accuracy of the manufacture.

Table 1: Details of models used to represent progressively more severe osteoporosis

Model Name	Trabecular Density (BV/TV)	Cortical Thickness (mm)
T0C0	Healthy (0.19)	Healthy (0.64 mm)
T1C0	Mild loss (0.10)	Healthy (0.64 mm)
T2C0	Severe (0.05)	Healthy (0.64 mm)
T2C1	Severe (0.05)	Mild (0.50 mm)
T2C2	Severe (0.05)	Severe (0.36 mm)

Foam and Silicone Moulding

Before testing the models, silicone mouldings were created to act as intervertebral disc phantoms. These were made by applying a small quantity of liquid silicone to a 2mm thick, flat, foam base. The vertebra model was firmly pressed on the silicone to leave an impression of the endplate. The foam and silicone provided a barrier between the metal platens and evenly distributed the load over the endplate during testing.

Compression Testing

A foam and silicone moulding was placed either side of the vertebra model. The vertebra was then placed on a 5kN Hounsfield testing machine platen (Hounsfield Test Equip. Ltd, United Kingdom). A video recorder was set up to view through the holes in the pedicles to record the deformation of the internal trabecular beams during testing. A 5000 N load cell was used to record compressive force during testing at a constant strain rate of 5 mm/min. The model was loaded until ultimate failure and the force and displacement data recorded. Following testing the specimens were examined and photographed to identify any fracture patterns. In order to quantify the stiffness of the intervertebral disc phantoms, two discs were placed either side of a metal cylinder of approximately the same cross sectional area as the vertebrae and compressed as before to obtain a characteristic force-displacement curve for the silicon discs.

Results

Table 2 gives the weight and wall thickness of each vertebra model following manufacture. Figures 2 and 3 give the load-displacement graphs for the varying trabecular densities and cortical wall thicknesses. The ultimate strength of the vertebra was defined as the first force data point before three consecutive force data points which decrease in force. Table 3 gives the mean and standard deviation for each model (n = 3).

Table 2: Dimensional accuracy of cortical wall thickness in SLS models

Vertebra Model	Weight (g)	Wall Thickness (mm)	
		Nominal	Actual
T0C0a	13.02	0.64	0.86
T0C0b	14.01	0.64	0.88
T0C0c	13.18	0.64	0.91
T1C0a	10.42	0.64	0.84
T1C0b	10.19	0.64	0.78
T1C0c	10.64	0.64	0.85
T2C0a	6.74	0.64	0.8
T2C0b	6.87	0.64	0.81
T2C0c	6.68	0.64	0.82
T2C1a	5.88	0.5	0.67
T2C1b	4.55	0.5	0.63
T2C1c	6.46	0.5	0.62
T2C2a	4.81	0.36	0.58
T2C2b	4.9	0.36	0.61
T2C2c	4.58	0.36	0.6

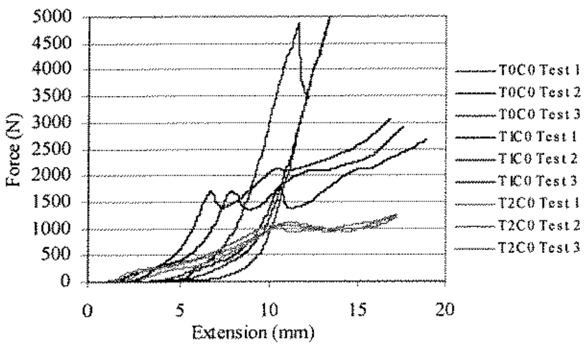

Figure 2: Force-displacement graphs for varying trabecular densities.

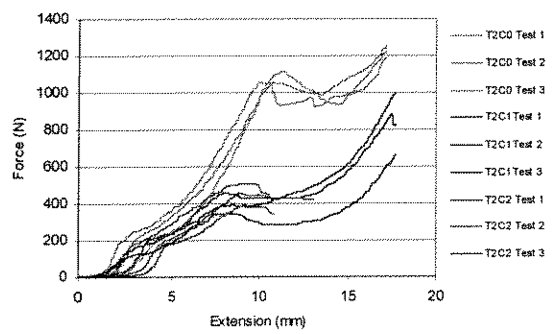

Figure 3: Force-displacement graphs for varying cortical wall thicknesses

The stiffness of the whole system (i.e. vertebra and discs) was determined for each test using intervals between 50-95% of the ultimate strength. Taking the first point at 50% ensured the initial non-linear behaviour of the disc was excluded, while the 95% upper point guaranteed any decrease in stiffness just before failure was excluded. The mean stiffness and standard deviation for each model are shown in Table 4.

Table 3: Ultimate strength of vertebra models

Model	Mean Strength (N)	SD (N)
T0C0	4962	66
T1C0	1715	22
T2C0	1076	34
T2C1	400	58
T2C2	451	63

Table 4: Stiffness of overall system between 50-95% of ultimate strength.

Model	Mean Stiffness (N/mm)	SD (N/mm)
T0C0	1121	6.9
T1C0	225	13.9
T2C0	169	22.7
T2C1	60	15.2
T2C2	60	5.4

To determine the stiffness of the vertebra alone, the system was analysed as two springs in series. One spring of stiffness K_{disc} represented the combined effect of the two discs while the other spring of stiffness $K_{vertebra}$ represented the vertebra. The overall stiffness of the system was K_{system}. The stiffness of the whole system and the stiffness of the disc were calculated using a range of force from 245N to 345N. Vertebrae stiffness was then calculated using equation (1). The mean vertebral stiffness and standard deviations are given in Table 5.

$$K_{System} = \frac{1}{\frac{1}{K_1} + \frac{1}{K_2}} = \frac{1}{\frac{1}{K_{Disc}} + \frac{1}{K_{Vertebra}}} \quad (1)$$

Therefore

$$K_{Vertebra} = \frac{-K_{Disc} K_{System}}{K_{System} - K_{Disc}}$$

Table 5: Mean stiffness and standard deviation of vertebra.

Model	Mean Stiffness (N/mm)	SD (N/mm)
T0C0	1714	404
T1C0	508	347
T2C0	91	17.4
T2C1	99	41.6
T2C2	100	11.7

Discussion

The results of this study show that moderate reductions in trabecular bone density greatly affect vertebral strength and stiffness. This finding compares favourably with previous studies (e.g. [3]). With a reduction in trabecular strut diameter and number of transverse ties to reduce bone volume to total volume ratio (BV/TV) from 0.19 to 0.10, mean strength dropped from 4962N to 1715N. Further reduction in BV/TV to 0.05 resulted in an 85% decrease in strength.

Cortical shell thicknesses suffered from considerable dimensional inaccuracy due to the manufacturing process, however based on measured (rather than nominal) shell thickness, the thickest cortical shell had the highest strength. However the T2C1 (nominally 0.50mm) and T2C2 (nominally 0.36mm) models had apparently random wall thicknesses ranging between 0.58 and 0.67mm, and Figure 3 shows similar failure behaviour and ultimate strength between the two.

In future work the models will be scaled up to reduce dimensional inaccuracies due to manufacturing tolerance limits, allowing a more detailed investigation of the effects of cortical wall thickness and micro architecture on vertebral body mechanics.

While the synthetic models were successful in allowing investigation of vertebrae stiffness and strength, they did not usually exhibit realistic fracture patterns due to the high ductility of the material. Future investigations will examine the use of other (more brittle) rapid prototyping materials. Figure 4 shows an example of a synthetic vertebrae fracture using a more brittle material which is closer to observed compression fracture patterns.

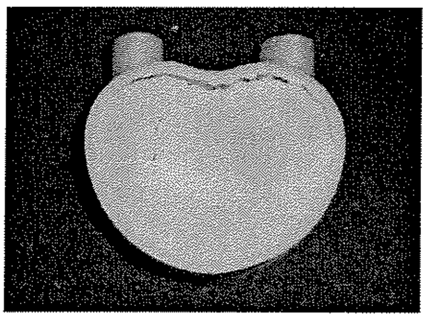

Figure 4: Fracture patterns produced during compression of synthetic vertebra made from a brittle rapid prototyping material.

Conclusions

This study developed a technique for producing synthetic vertebrae where the level of osteoporosis was able to be specified through variation of trabecular micro-architecture and cortical wall thickness. Measured reductions in strength compared favourably with previous literature. Trabecular bone micro-architecture (and therefore density) plays a pivotal role in determining both strength and stiffness. With further development, the synthetic vertebrae will provide a useful tool for investigating the biomechanics of vertebral compression fractures and treatment methods such as vertebroplasty and kyphoplasty.

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