The effect of axial load on the sagittal plane curvature of the upright human spine in vivo.

Judith R. Meakin, PhD[1], Francis W. Smith, MD[2], Fiona J. Gilbert, FRCR[2] & Richard M. Aspden, DSc[1]

[1]Department of Orthopaedics, [2]Department of Radiology, University of Aberdeen, Foresterhill, Aberdeen, AB25 2ZD

Corresponding author

Dr Judith R. Meakin

Room 2.26 IMS Building, University of Aberdeen, Foresterhill, Aberdeen, AB25 2ZD, UK

Tel. +44 (0)1224 553555   Fax. +44 (0)1224 559533   Email j.meakin@abdn.ac.uk

Keywords

Spine; active shape model; load bearing; positional MRI

Type of manuscript

Original Article.

Abstract

Determining the effect of load carriage on the human spine *in vivo* is important for determining spinal forces and establishing potential mechanisms of back injury. Previous studies have suggested that the natural curvature of the spine straightens under load but are based on modelling and external measurements from the surface of the back. In the current study, an upright positional MRI scanner was used to acquire sagittal images of the lumbar and lower thoracic spine of 24 subjects. The subjects were imaged in standing whilst supporting 0 kg, 8 kg and 16 kg of load which was applied axially across the shoulders using an apron. An active shape model of the vertebral bodies from T10 to S1 was created and used to characterise the effect of load. The results from the shape model showed that the behaviour of the average shaped spine was to straighten slightly. However, the shape model also showed that the effect of load exhibited systematic variation between individuals. Those who had a smaller than average curvature before loading straightened under load, whereas those who had a greater than average curvature before loading showed an increase in curvature under load. The variation in behaviour of differently shaped spines may have further implications for the effects of load in lifting manoeuvres and in understanding the aetiology of back pain.

Introduction

Understanding the effects of load carriage on the human spine *in vivo* is important for determining the forces generated within the spinal tissues. This in turn is important for establishing potential mechanisms of injury and elucidating possible relationships to back pain. The effect of load *in vivo*, however, is not well defined.

Laboratory experiments on isolated spines have limited ability to reproduce physiological loading conditions and, in the worst case (Patwardhan et al., 1999), show the spine buckling under unreasonably small loads. When more realistic loading patterns are used, the spine is able to support more reasonable loads but shows either an increase or decrease in lordosis depending on the path of the loading (Patwardhan et al., 1999). In comparison to this, it has been predicted using a model that, *in vivo*, the lumbar spine should straighten under load to maintain its structural stability (Aspden, 1989). Several *in vivo* experimental studies suggest that this prediction is valid. Placing a 15 kg load, via a frame, on the head of a subject, for example, was found to produce an increase in height and a posterior movement of markers placed against the lumbar region (Perrin, 1966). It was deduced from this that, under axial load, the spine elongates and reduces its natural curvature. More recently, experiments were carried out on a group of 10 male subjects with skin markers placed at various points along the thoracic and lumbar spine. The subjects were given loads up to 445 N, distributed evenly between the two hands, with the arms extended downwards at the sides of the body. The distance between L1 and S1 was found to increase significantly with load, as was the posterior tilting of the pelvis. This was taken to indicate that that the lumbar spine had flattened. Similar experiments were also performed on 15 male subjects holding loads up to 380 N (El-Rich et al., 2004). The effect of load was found not to be significant, but, as in previous experiments, was reported to produce a flattening of the lumbar lordosis and a posterior tilting of the pelvis. The main drawback of these experiments is that they all measured changes in spinal shape from the external surface of the back. As acknowledged by the authors themselves (El-Rich et al., 2004), this may not truly reflect what is happening to the vertebral bodies, the major weight bearing component of the spine.

To observe the spine within the body requires an imaging method, such as MRI (Wisleder et al., 2001a) or biplanar radiography (Reuban et al., 1979), to be undertaken. MRI, which has the advantage of using non-ionising radiation, has been used to observe the spine *in vivo* under axial loading (Wisleder et al., 2001a). The horizontal orientation of a conventional MRI scanner, however, meant that the loading was applied using a harness (Wisleder et al., 2001a; Wisleder et al., 2001b). This may not exactly represent the condition of a freely supported load under gravity since there are additional constraints imposed by the action of the harness and the contact between the subject's back and the bed of the scanner.

Using a positional MRI scanner (pMRI), where subjects can be scanned in upright postures, provides an alternative method for observing the behaviour of the spine under axial load. In the study reported here, we used pMRI to investigate the effects of loading on a group of healthy volunteers. To help assess the effects of load on the shape of the spine, an active shape model (ASM) was used. ASM and, more recently, Active Appearance Modelling (AAM), are image processing methods used to locate and characterise objects in a set of images (Cootes & Taylor, 2004). The method has recently been shown to be a useful way of describing the shape of the lumbar spine in a reliable and precise manner (Meakin et al., 2008).

Methods

Subjects

The data from twenty four volunteer subjects were used in this study. The subjects comprised 11 females and 13 males with an average age of 36 years (21 – 59 years); height, mass and body mass index are given in Table 1. None of the subjects reported any current back pain or any serious previous episodes of back pain (requiring time off work or a visit to their doctor). Subjects with any known spinal pathology were excluded. The study was approved by Grampian Local Research Ethics Committee and all subjects gave their informed consent to participate.

Imaging

Imaging was performed using a Fonar 0.6 T Upright TM positional MRI scanner (Fonar Corporation, Melville, New York). A single slice image of 4.5 mm thickness was acquired as close to the mid-sagittal plane of the spine as possible using a T2 weighted imaging protocol (TR = 1156 ms, TE = 140 ms). This protocol was chosen because of its relatively short acquisition time of 28 seconds; thus minimising artefacts from subject movement. The image data were acquired as a 256 x 160 matrix using a 40 cm field of view, reformatted as a 256 x 256 image and saved in bmp format. The images were subsequently scaled by a factor of 2 with cubic interpolation using GIMP software (http://www.gimp.org/).

Experiments

The subjects were imaged in the upright standing posture. The subjects were asked to stand as naturally as possible, without leaning on the scanner walls and with their arms hanging downwards. They were given a specially made tabard (cobbler) style apron to wear which had pockets along the lower edge of both the front and back of the apron.

Load was applied by placing lead weights into the apron pockets. The weight was distributed evenly at the front and back, left and right sides, to minimise bending moments. Three images were acquired with the subject supporting a total of 0 kg, 8 kg and 16 kg. The maximum load of 16 kg was chosen because it is the suggested upper limit, for lifting and lowering by female subjects, in the UK manual handling guidance (HSE, 2004).

Shape modelling

The ASM was created using the Active Appearance Modelling software tools from the University of Manchester, UK (http://www.isbe.man.ac.uk/~bim/software/am_tools_doc/index.html). The model comprised 28 landmark points placed around the edges of each of the vertebral bodies from T10 to S1 (Figure 1). The same number of landmark points (252 in total) was used for each image and always referred to the same feature (e.g. the mid-point of the superior end-plate of a given vertebral body). The points were then used by the software to create the model for the shape of the spine. As part of this process the points were aligned into a common co-ordinate frame by scaling, translating and rotating; this means that size differences and rigid body movements were removed from the model. Principle component analysis (PCA) was then used to analyse the variation in the position of the points across the 72 images. PCA is a statistical analysis method that allows multi-dimensional data sets (in this case there were 504 dimensions; one for the x and y co-ordinate of each landmark point) to be summarised using far fewer dimensions. This is achieved by calculating the eigenvectors and eigenvalues of the covariance matrix of the data. The eigenvectors describe patterns of variation in the data; these are orthogonal and therefore statistically independent. The relative importance of each eigenvector is denoted by the eigenvalue and by discarding the less important eigenvectors the data can be efficiently described whilst maintaining its essential characteristics. In the ASM, the eigenvectors are called ‘modes of variation’ and each image was assigned values to describe the spinal shape in terms of these modes. The mode values were expressed as the number of standard deviations from the mean shape calculated from all 72 images.

A previous analysis of the data (Meakin et al., 2007a) only considered the vertebral bodies from L1 to S1. Although these six vertebrae describe the anatomical lumbar region, the lordotic region of the spine has been shown to extend up to T10 (Roussouly et al., 2005). An additional three thoracic vertebrae were thus added to the model to ensure that whole of the lordosis was captured for all subjects.

Additional measures for model validation

The points used for the shape model were also used to determine two other parameters; these were used to validate the model. Firstly, the nine vertebral body centroids were calculated using Matlab software (The MathWorks Inc., Natick, Massachusetts). Matlab code to do this was downloaded from the MathWorks file exchange; it was contributed by Damien Garcia and uses a method described by Russ (1995). The centroids were then used to determine two measures of straightness: length ratio and deviation. The length ratio was defined as the ratio of the cumulative distance between consecutive vertebral body centroids from T10 to S1 to the linear distance between these two end-points. This ratio has a minimum value of 1 for a line of zero curvature; higher values indicate increasing curviness. The deviation was defined as the root mean square distance of the vertebral body centroids to the line connecting T10 and S1 (measured along lines normal to this straight line). Large values of deviation indicate greater curvature with a value of zero for a perfectly straight line.

Statistics

Repeated-measures analysis of variance (within subject effects and contrasts) was used to evaluate the consistent effects of load, and measure of agreement correlation (Bland & Altman, 1986) was used to evaluate the systematic differences in the effects of load. Post-hoc tests were performed using a Sidak adjustment for multiple comparisons. Correlations between parameters were evaluated using Pearson's R correlation coefficient (normally distributed data) or Spearman's ρ rank correlation coefficient (non-normally distributed data). The Kolmogorov-Smirnov test was used to asses whether data followed the normal distribution. For the RM-ANOVA, data were also tested using Mauchly's test of sphericity; for data that failed this test the Greenhouse-Geisser correction was used. For all tests, a probability of 5% or less was taken to indicate statistical significance.

Results

Shape model

The first mode of variation (Figure 2) identified by the shape model accounted for 70 % of the total variance in shape. The second mode accounted for 13 % and subsequent modes for less than 5 % each; these were ignored as they described small differences in shape compared to the first mode. It can be seen from Figure 2 that a mode 1 value 2 standard deviations below the mean (to the left in Figure 2) describes a spine that has a large lordotic curvature, which changes to kyphotic at the level of T11. At 2 standard deviations above the mean, mode 1 describes a spine with much smaller lordotic curvature that changes to kyphotic at L2.

The mode 1 values were found to differ significantly with load ($F = 5.9$, $P = 0.01$). A significant linear contrast ($F = 7.1$, $P = 0.01$) was also found, demonstrating the linear relationship between mode 1 and load as shown in Figure 3. This corresponded to the average mode 1 value for the 24 subjects increasing by approximately 0.1 standard deviations for each increment of load. Post-hoc tests showed that the increase was significant between 8 kg and 16 kg (95% CI = 0.01 – 0.22; $P = 0.03$) and between 0 kg and 16 kg (95% CI = 0.01 – 0.36; $P = 0.04$), but not between 0 kg and 8 kg (95% CI = -0.06 – 0.19; $P = 0.4$). These results indicate that the average spine straightens under load.

In addition to this, a positive linear relationship was found between the mode 1 values and the change in value with load (Figure 4). This corresponds to spines with low mode 1 values becoming more negative under load (i.e. moving to the left in Figure 2) and spines with high mode 1 values becoming more positive (i.e. moving to the right in Figure 2). The correlation was not statistically significant for 0 kg and 8 kg ($\rho = 0.40$, $P = 0.06$), but was significant for 8 kg and 16 kg ($R = 0.44$, $P = 0.03$) and for 0 kg and 16 kg ($R = 0.59$, $P = 0.002$). This result indicates that spines of different unloaded shape behave differently under load.

Model validation

A significant correlation was found between mode 1 and both the length ratio ($R = -0.97$, $P < 0.001$) and the deviation ($R = -0.85$, $P < 0.001$). This confirmed that, as could be deduced visually from Figure 2, a spine with a high mode 1 value was straighter (i.e. has a lower length ratio and deviation) than one with a low mode 1 value mode.

Discussion

A shape model was used to describe the shape of the spine, in the sagittal plane, from T10 to S1 in 24 subjects standing with 0 kg, 8 kg and 16 kg of load applied via the shoulders. The model showed that, for the average shaped spine, the spine straightened slightly under load. The model also indicated that the effect of load varied between individuals in a systematic way. Those who had unloaded spines described by a large, mainly lordotic curvature became more curved under load whereas those with a smaller lordotic region became straighter under load.

Straightening of the spine under load has previously been predicted by the arch model of the spine (Aspden, 1989), and found experimentally by measurements on the external surface of the back (Shirazi-Adl & Parnianpour, 1999; El-Rich et al., 2004). The relationship between unloaded shape and change in shape under load is a new result but may explain the observation by Wisleder et al. (2001a) that some of the measurements they made on axially loaded spines were dichotomous. It may also account for the contradictory conclusions that have been reached about whether the lumbar lordosis increases or decreases in pregnancy (Franklin & Conner-Kerr, 1998).

The shape of the unloaded spine, and the change in shape with load, will influence the shear and compressive stresses that are generated in the spinal tissues. With the exception of one analysis (Keller et al., 2005), biomechanical models have predicted that straightening the spine leads to a greater proportion of the resultant force becoming compressive (Aspden, 1989; Shirazi-Adl & Parnianpour, 1999; Shirazi-Adl et al., 2002). The variation in spinal shape, and the differing behaviour under load, indicates that there could be quite considerable inter-subject variability in the forces that the spine experiences. For some individuals, load bearing and lifting may result in an increase in shear stress, placing them at a greater risk of injury than individuals for whom shear stresses are reduced. We are currently performing a study to determine how spinal shape effects the way people prefer to lift loads and will use the data in a model to predict how this affects the spinal forces. Many biomechanical studies only consider the average spinal shape; our results suggest that it may be beneficial to incorporate variability to help elucidate the complicated relationships between biomechanics, injury and back pain. We hypothesise that this may lead to manual handling advice being tailored to each individual, perhaps advocating free-style techniques (Arjmand & Shirazi-Adl, 2005), rather than issuing uniformly prescriptive advice.

The consistency between the results of our study and those that made measurements on the surface of the back (Perrin, 1966; Shirazi-Adl & Parnianpour, 1999; El-Rich et al., 2004) suggests that external measurements are adequate for determining changes in spinal shape. However, they are unable to determine the absolute shape unless the spinous process length is known. Determining the relationship between absolute shape and change in shape requires a more sophisticated method such as that used here. Shape modelling is advantageous as it is able to describe the shape of the spine comprehensively and quantifiably and allows easy visualisation (Meakin et al., 2008). Simpler methods may fail to achieve this; the length ratio and deviation used as validation in this study, for example, can determine straightness but can not differentiate between a large single curvature and two smaller curvatures. Furthermore, the data can provide not only the accurate position and dimensions of the vertebrae, but also their relative translations and rotations; supplying valuable kinematic data for biomechanical models.

It is not possible, from the current study, to determine the mechanisms behind the differences in the behaviour of the spine under load, although several possibilities can be hypothesised. Firstly, it may be a passive response of the spine to the additional load. The spine can be thought of as a pre-buckled structure (Meakin et al., 1996) which, on the application of axial load, will buckle further. The new buckled shape will depend on the initial shape and also on the position of the applied load (Patwardhan et al., 1999). Although we aimed to distribute the loads evenly about the subject's body to minimise bending moments, the position of the load with respect to the sacrum may have differed due to variation in the subject's upper body shape. The second possibility is that it was an active response where the recruitment strategy of the spinal muscles, used to maintain spinal stability, differs in spines of dissimilar shape causing their behaviour under load to differ accordingly. We have recently performed pilot work to enable us to look at recruitment strategies of spinal muscles (Meakin et al., 2007b) and propose to investigate the relationship to lumbar spine shape.

The portion of the spine incorporated in the model was based on a study that found lumbar lordosis could extend as far as T10 (Roussouly et al., 2005). We did not find lordosis to extend this far in any of our subjects, but this is not unreasonable given that our cohort was smaller (24 compared to 160). However, we did find that the transition point was frequently above L1, justifying our use of more vertebral bodies than in our previous analysis (Meakin et al., 2007a). Ideally, we would also have acquired data on the remainder of the spine, the position of the head, the morphology of the pelvis and its orientation about the hips. This would have, for example, allowed eccentric loading induced by variation in the subject's body shape to be estimated. However, the maximum field of view of the MRI scanner limited how much of the body could be imaged. We intend to address this in a future study by combining information on spinal shape acquired using MRI with externally measured data.

Acknowledgements

The authors thank Mrs B MacLennan (research radiographer) for acquiring the MR images and NHS Grampian Endowments Trust for funding the research.

References

Arjmand, N., Shirazi-Adl, A., 2005. Biomechanics of changes in lumbar posture in static lifting. Spine 30, 2637-2648.

Aspden, R.M., 1989. The spine as an arch. A new mathematical model. Spine 14, 266-274.

Bland, J.M., Altman, D.G., 1986. Statistical methods for assessing agreement between two methods of clinical measurement. Lancet I, 307-310.

Cootes, T.F., Taylor, C.J., 2004. Anatomical statistical models and their role in feature extraction. British Journal of Radiology 77, S133-S139.

El-Rich, M., Shirazi-Adl, A., Arjmand, N., 2004. Muscle activity, internal loads, and stability of the human spine in standing postures: combined model and in vivo studies. Spine 29, 2633-2642.

Franklin, M., Conner-Kerr, T., 1998. An analysis of posture and back pain in the first and third trimesters of pregnancy. The Journal of Orthopaedic & Sports Physical Therapy 28, 133-138.

HSE, 2004. Getting to grips with manual handling. A short guide. Health and Safety Executive INDG143(rev2).

Keller, T.S., Colloca, C.J., Harrison, D.E., Harrison, D.D., Janik, T.J., 2005. Influence of spine morphology on intervertebral disc loads and stresses in asymptomatic adults: implications for the ideal spine. The Spine Journal 5, 297–309.

Meakin, J.R., Aspden, R.M., Smith, F.W., Gilbert, F.J., 2007a. The effect of load on the shape of the lumbar spine. Journal of Biomechanics 40, S270.

Meakin, J.R., Semple, S.I., Redpath, T.W., Aspden, R.M., Gilbert, F.J., 2007b. Recruitment of spinal muscles in static postures. In Proceedings of the Joint Annual Meeting of the International Society for Magnetic Resonance in Medicine and the European Society for Magnetic Resonance in Medicine and Biology. Berlin, Germany.

Meakin, J.R., Gregory, J.S., Smith, F.W., Gilbert, F.J., Aspden, R.M., 2008. Characterising the shape of the lumbar spine using an active shape model: reliability and precision of the method. Spine 33, 807-813.

Patwardhan, A.G., Havey, R.M., Meade, K.P., Lee, B., Dunlap, B., 1999. A follower load increases the load-carrying capacity of the lumbar spine in compression. Spine 24, 1003-1009.

Perrin, R., 1966. Rééducation vertébrale: principes, techniques. Librairie le François, Paris, pp. 13-32.

Reuban, J.D., Brown, R.H., Nash, C.L., Brower, E.M., 1979. In vivo effects of axial loading on healthy, adolescent spines. Clinical Orthopaedics and Related Research 139, 17-27.

Roussouly, P., Gollogly, S., Berthonnaud, E., Dimnet, J., 2005. Classification of the normal variation in the sagittal alignment of the human lumbar spine and pelvis in the standing position. Spine 30, 346-53.

Russ, J.C., 1995. The image processing handbook, second edition. CRC Press, Boca Raton, pp. 487-490.

Shirazi-Adl, A., Parnianpour, M., 1999. Pelvic tilt and lordosis control spinal postural response in compression. In Transactions of the Orthopaedic Research Society. Anaheim, CA, p 1012.

Shirazi-Adl, A., Sadouk, S., Parnianpour, M., Pop, D., El-Rich, M., 2002. Muscle force evaluation and role of posture in human lumbar spine under compression. European Spine Journal 11, 519-26.

Wisleder, D., Smith, M.B., Mosher, T.J., Zatsiorsky, V., 2001a. Lumbar spine mechanical response to axial compression load in vivo. Spine 26, E403-E409.

Wisleder, D., Werner, S.L., Kraemer, W.J., Fleck, S.J., Zatsiorsky, V.M., 2001b. A method to study lumbar spine response to axial compression during magnetic resonance imaging. Spine 26, E416–E420.

Table 1. Height, mass and body mass index (BMI) of the subjects (n= 24).

	Mean (sd)	Range
Height (m)	1.74 (0.09)	1.53 – 1.91
Mass (kg)	79 (16)	49 - 124
BMI (kg m^{-2})	26 (4)	21 - 39

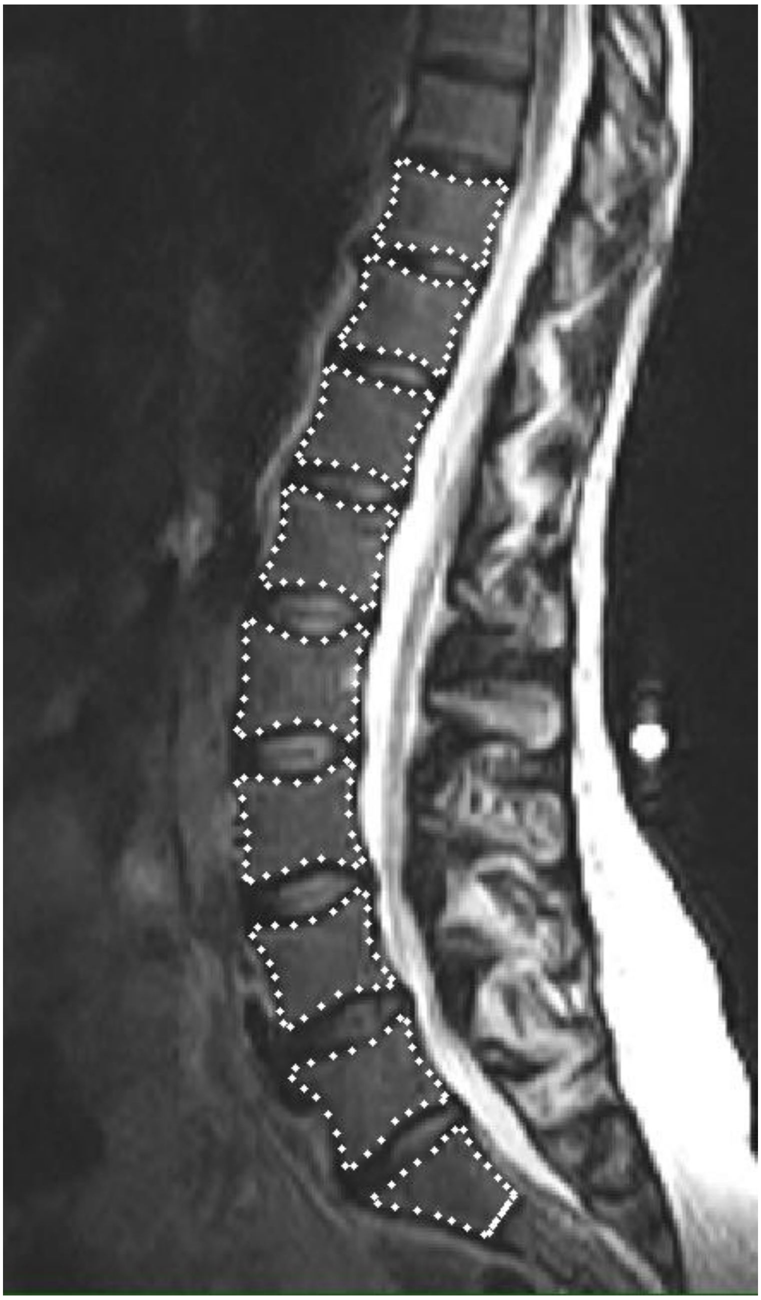

Figure 1. Shape model of the spine from T10 to S1.

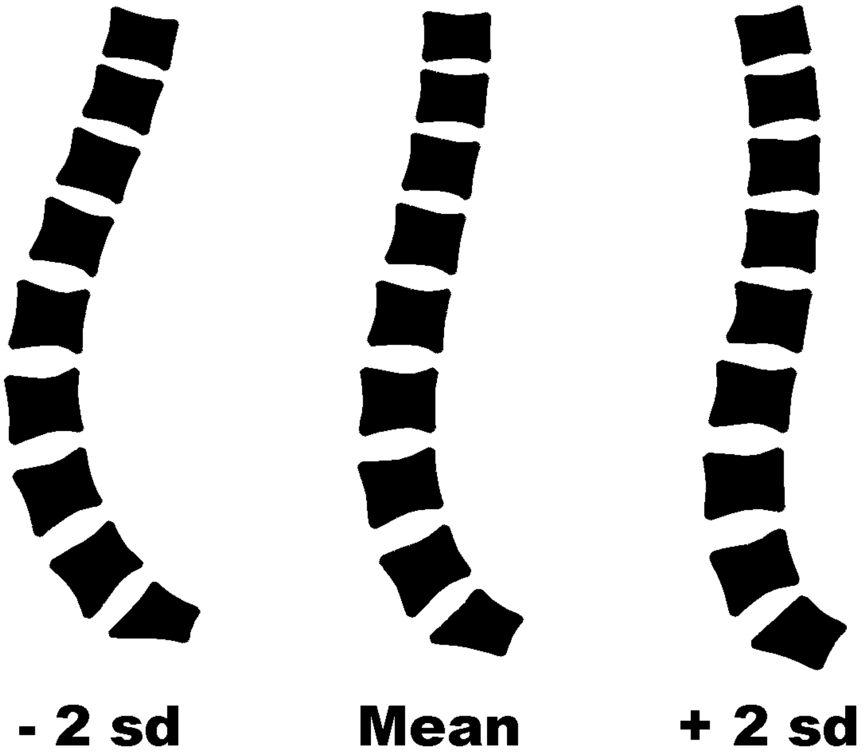

Figure 2. The mean shape of the spine from the 72 images is shown in the centre with –2 and +2 standard deviations in the first mode of variation.

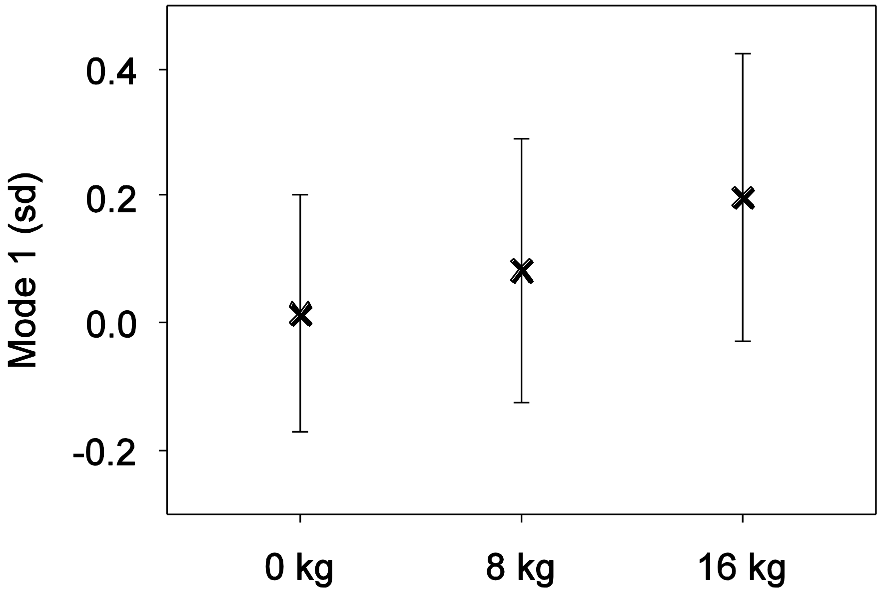

Figure 3. Mode 1 value for each load case. Error bars denote the standard error (n = 24).

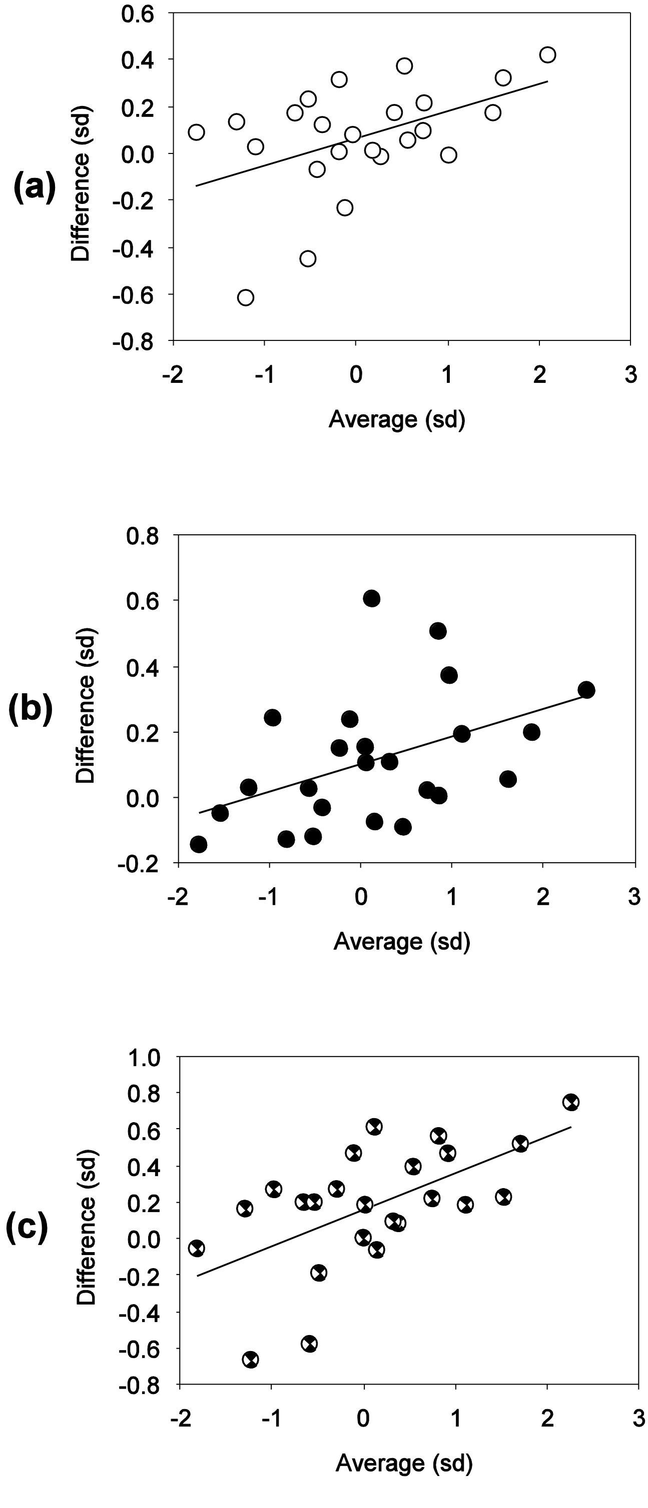

Figure 4. Measures of agreement plots for (a) 0 & 8 kg, (b) 8 & 16 kg, and (c) 0 & 16 kg of load.