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MATHEMATICAL MODELLING OF HUMAN SPINE AND DESIGN

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Abstract: Many individuals suffer from back trouble and a large number of sufferers provide a hidden cost to industry, from the increasingly high level of absenteeism. Back pain and injury may result from inadequately designed artefacts and workplaces. In order to achieve better designs which prevent such injuries one has to have a greater understanding of the source of the problems. The mechanics of human spine can be studied by conducting experiments directly on humans in a laboratory. Alternatively mathematical models which represent subtleties and geometric complexities may be studied. Such models of human spine could look at how the spine behaves in specific situations. This paper is about generating a general purpose spine model that is suited a wide range of design applications. The geometric model and the mathematical modelling aspects will be explained. The result of the research infeasibility of range of models representing the spine will also be discussed. The paper will conclude with suggestions on the potential use of human spine models in design.

1. Introduction

Despite advances in modem working conditions, the physical demands of human work inevitably lead to loading of the spine that potentially can cause difficulties in many respects such as reduced performance, physical inability to carry out the task, acute damage to the spine or long term spinal problems leading to back pain. In Western societies most people suffer at least one severe attack of back trouble during their working life. In the UK, 60% of adults were estimated to suffer from back pain at some stage in their lives. according to National Back Pain Association (NBPA, 1995). In addition to the pain and distress this may cause to the individual, financial costs to industry is huge. The same source also reveals that this bears a hidden cost to the employers who have increasingly high level of staff absenteeism. The number of days of certified incapacity of musculoskeletal disorder rose from 34.2 million days in 1971/72 to 81.3 million days in 1992/93.

Recognition of the value of human factor discipline in Western societies has helped the improvement of design in many areas however it is still not unusual to observe the artefacts being designed without ensuring that an optimal posture is attainable by people who will use them. Rarely any problems discomfort in the back is taken seriously until medical intervention is required. There is clearly a need for a design tool which can predict both the range of postures and loads that will be imposed on the spine for people of different age, gender, professions and lifestyles.

One approach to alleviate the consequences of poor design practice might be the laboratory testing of products with human subjects, but this may not always be possible. Sometimes it is impossible to imitate the environment in the laboratory, sometimes the duration of experiments is unrealistic for research projects but most importantly under certain conditions direct human experiment may pose risks for human welfare.

In order to identify the possible causes of spinal disorders, experimental investigations can also be performed on the spine using cadaver specimens. Tests on cadaver specimens have been used to investigate the behaviour of the inter vertebral joint under various forms of loading. However, while this allows a more detailed investigation, the absence of a neuromuscular response, and the altered geometric and material properties of dead tissue, may reflect behaviour different to that of a spine of living subjects.

Alternatively, the problem can be analysed within a computer mathematical model to provide a humanly safe method of evaluating a wide range of possibilities. In this case, the natural variation in geometric and material properties of spinal specimens can be controlled. Mathematical modelling techniques will provide opportunity to repeat the experiments as many times as one likes without distress to the subjects or deformity of the material.

Within the constraints of other techniques, mathematical modelling techniques have become an invaluable means of investigating the behaviour of the human spine.

2. Spine Modelling

Mathematical models of the human spine have been studied by many distinguished scientists. Much of the early spine modelling efforts have dealt with the responses of the spine to impact loads. These studies go back more than 100 years (e.g. Messerer, 1880). Later the focus on spinal injuries originated from studies of pilots involved in assisted ejection from high speed aircraft (Perry, 1974). For example, Belytschko *et al* used a three-dimensional model of the head and neck, to perform parameter studies regarding the influence of various helmet and helmet mounted devices, the rate of ejection and the angle of seat inclination. By predicting combinations which would reduce the likelihood of vertebral fracture, the design of safer equipment could be pursued.

Similar models have been developed to investigate spinal behaviour in a horizontal impact situation such as automobile collision. In particular the effects of various restraint systems such as lap and chest harnesses have been investigated in Williams and Belytschko, 1983. Modelling simulations of the spine in a situation which is difficult to investigate experimentally have contributed towards the design of automobile and aircraft.

Several models have also contributed towards the techniques used in the correction of the lateral deformation of scoliosis. Clinical orthotic devices used to correct this disorder have been simulated using models, in order to predict the effectiveness of their corrective forces (Wynarsky and Schultz, 1991). Modelling has therefore contributed significantly in this area.

Microscopic disorders affecting the individual components of the spine have been investigated using finite element techniques. The clinical importance of the finite element models, in addition to those investigating the effectiveness of spinal fixation devices, has been fully discussed by Goel and Gilbertson, 1995. For example, in the vertebrae, alteration of the material properties has allowed the effect of osteoporosis common in post menopausal women.

Increasing attention has been paid to spine modelling mainly due to difficulties in achieving precise diagnosis of back pain problems. An extensive critical review of existing mathematical models of human spine can be found in Grilli and Acar, 1997(a).

3. Methodology

Advances in computer modelling, especially in computer graphics and surface modelling provide opportunities to generate the images of irregular structures such as human spine. These images are realistic and bear the true likeness to the human spine however most of these are either not functional or use simple modelling concepts. For example some medical packages have a realistic geometric representation of the spine. Although it can be used as an electronic atlas and provides service to the medical students, it has no analytic or design application functionality.

Human modelling has for some time been used in the design context, but even the most sophisticated commercially available human modelling packages do not have an effective spine model.

The work reported in this paper therefore has adopted an integrated approach to the geometric modelling of the spine, mathematical modelling of the spine and integration of these into a human model. The progress in two areas, namely the geometric and mathematical modelling of the spine, will be explained in the next two sections. Their eventual integration with the SAMMIE Computer Aided Ergonomics Design System will improve the kinematic and analytic capabilities of the human model.

4. Geometric Modelling

Functional geometric models have also been studied by other scientist, mostly for use in finite element analysis. The information for these models has been retrieved from CT (computed tomography) scans, radiographs or dried spinal specimens and the models constructed in a finite element mesh. Even though these mesh models provide a realistic representation of the vertebrae, they tend to be very individualised and new digitised data would be required to develop a new model.

Our geometric model needed to be developed as a solid so that material properties can be assigned and mechanical analysis can be carried out at later stages. The geometric model, also uses standards such as IGES or STEP, and can be imported into other packages for further analysis to be carried out.

The detailed description of the generation and use of this geometric model can be found in Stepney *et al* (1996, 1997) however, the description below summarises the main points.

As a first step, the form and structure of the individual vertebrae and disks have been studied to identify the 'features' and to analyse their structure in order to determine areas of possible parameterisation. The parameters which have been defined and measured either tend to be related to the main structure of the spine, such as the width, height and depth of the vertebral body and pedicles, or they tend to concentrate on a specific part of the vertebra, such as the articulating facets.

In order to find the parameters and information for this new computerised model, a physical spine model was analysed to understand its structure and shape. From this study the ten main structures of the vertebra were viewed to see how they fit together and 60 parameters were measured. The measurements were taken by using a calliper and compass, providing a linear accuracy to 0.1mm and an angular accuracy to 1°. The structures within the vertebra are the vertebral body, spinous process, left and right pedicles, left and right transverse processes, left and right inferior facets, and left and right superior facets. The intervertebral disks were also studied for form and contain only 6 parameters, having a far simpler geometric structure. The reason for the large number of parameters is the complex non-linear nature of each vertebra.

The vertebrae are constructed from complex shapes which cannot be easily modelled. Furthermore, each vertebra is individual to each person. Due to both individuality and its complicated mathematical nature it is necessary to simplify the features into regular shaped sections in order to create the model. The parameters, developed above are used to control the geometry of these simplified features.

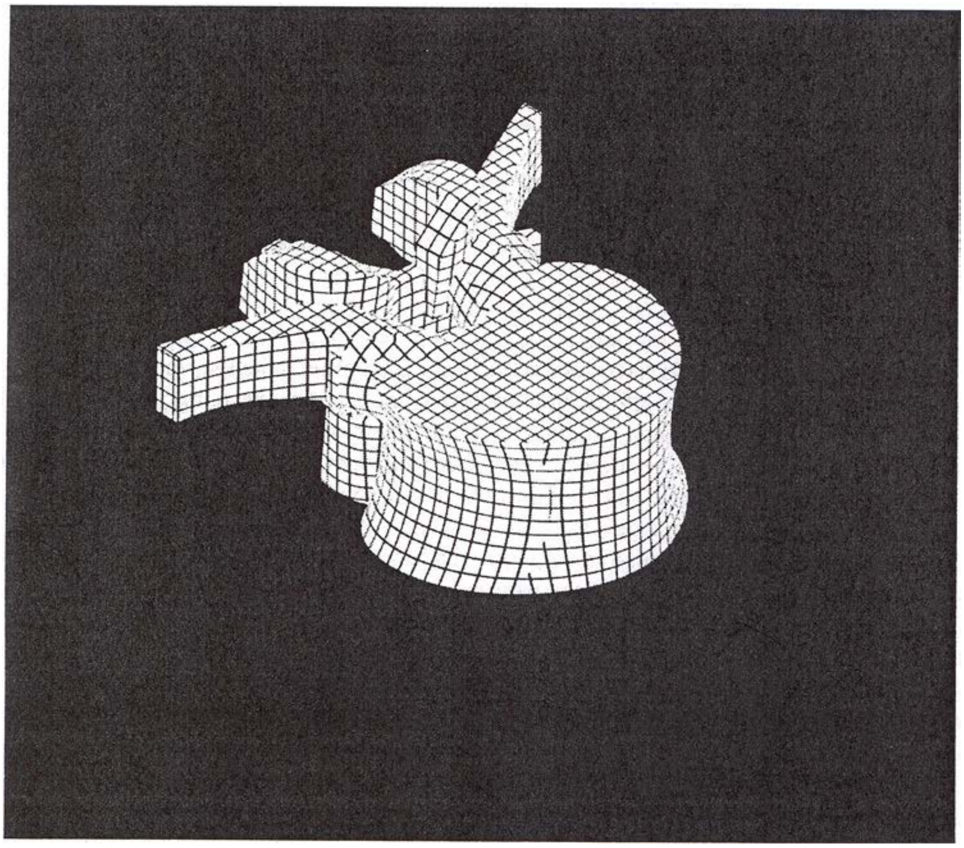

Figure 1: The L3 Lumbar Vertebra

For the vertebral body, the closed sections were created by fitting a cubic B-spline curve through ten points in 2-D space incorporating the parameters of length and width. The lower section was created by the same method, but with different dimensions for length and width, and using the Z co-ordinate to define the height. The end result is two parallel kidney shaped curves. The guide curves were created by utilising two existing points and defining a third point half way between the curves. This third point incorporates the two parameters describing the central cross-section of the vertebral body. These parameters give us the concavity of the vertebral body. The height of the vertebral body is assumed to be constant throughout each individual vertebra. The difference between the front and the back heights of the vertebral bodies have been added onto the height of the intervertebral disk. This allows us to maintain the correct vertebral inclination in the transverse plane.

The pedicles are simplified as two elliptical cylinders with the height, width and length parameters providing all the necessary information. The position of the pedicles is related to the width of the spinal canal. The pedicles were then rotated in both the transverse and sagittal planes to provide their inclinations.

Three parallel rectangles make up the section curves for each transverse process. The distance between the nearest and furthest rectangles define the length of the transverse process. The second rectangle is positioned mid way between the other two. The height and width of each rectangle relate to the dimensions of these parameters. The guide curves are created by fitting a quadratic B-spline curve to the corresponding corner points. The two outer rectangles have translations in both the X and Z axes to provide inclinations in both the sagittal and longitudinal planes.

The spinous process is also defined by three parallel rectangles created in the same way as those for the transverse processes. These two furthest rectangles have a translation in the Z axis to provide the transverse inclination.

For the lumbar vertebrae, the superior facets are created from four rectangles, two of which are positioned at right angles to the other two. These rectangles are then swept to provide an internal curve. This geometry was then rotated about the Y-axis to provide an inclination.

The inferior facets are constructed by sweeping from one rectangle to another with an internal angle of 45°. This creates an external curve which is compatible with the internal curve of the superior facet.

Once all these features have been created they were all merged to create a single solid body.

The model, as seen in Figure 1, is representative of a typical vertebra (excluding the atlas and axis) created using the program.

The models created have a functional rather than aesthetic appearance. The vertebra are created as individual solids and values for their density can be assigned to them. When the models are placed within an assembly the articulating facets meet each other to provide the points of contact required for rotation.

The purpose of generating a solid model of the human spine has been to create a representation of one generic spine rather than one which is representative for (most of) the population. The model which is created here can be used as a template for many different spinal segments, and spines by editing the parameters.

Furthermore, if the nature of spinal degeneration is known then this information could be used in the development of techniques for preventative medicine. The model created in this study is based on a healthy spine, but it can be eroded manually for use in the analysis and study of spinal disorders.

5. Mathematical Modelling and Discussions

The most commonly used model in the literature, the lever model, typically describes the spine as a rigid lever with no proper consideration of spinal curvature. Loading is limited to external loads applied directly or indirectly to the spine with balancing reaction forces at the sacrum. This model can predict unacceptably large reaction forces at the sacrum. On the other hand modelling the spine as an arch as Aspden, 1989 suggested might provide a more realistic representation since it considers spinal curvature and internal forces.

The resultant forces acting at each point of loading on the arch collectively follow a path known as the thrustline. Stability of the arch is achieved when these forces can be transmitted through its entire length. For this to occur, the thrustline must lie within the cross section of the arch.

Posterior extrusion of fibrocartilage from the disc, caused by hydraulic wedging pressure and the stretching of ligaments may cause back pain. A criterion for the risk of spinal disorders states that if the thrust line for a specific spinal posture is not located within the safety core of spine then the risk of back pain might increase. The spine is a statically indeterminate structure therefore an infinite number of thrust lines can be obtained. To find a thrust line which is as close as possible to the reference line of the spine might suggest a good posture. This can be obtained by using optimisation techniques.

For direct comparison purposes an example which was studied by Morris *et al* 1961 and by Aspden, 1989 has been considered. A 77 kg (755 N) male subject carrying a 900 N load in both hands is in a stooped position. This load is assumed to act on T6. The weight of the head is estimated as 130 N, and the trunk weight as 230 N. These weights are assumed to act on T2 and TI2 respectively. Intra-abdominal pressure of 70 N acts on each of the vertebrae LI-L5 ..

Morris *et al* made an analysis by calculating the moments of the forces about the sacrum and found a lumbo-sacral reaction of 9200 N which was agreed that 'will overstress the spine'. Introducing the intra-abdominal pressure could act as a load relieving mechanism reduced the reaction to 6600 N.

Re-analysing the same case Aspden found no trust line to satisfy the safe theorem. Even the one which fitted most closely lied outside the cross section of the arch formed by the vertebral bodies which implies an unstable spine. However he claims that if the posture of the spine was adjusted to introduce a lumbar lordosis the thrust line would lie safely within the arch. He concludes that this may explain the posture of the weight lifters who actively maintain a lumbar lordosis. He finds lumbo-sacral load as 1400 N.

However, infinite number of thrust lines can be obtained by selecting different pole points which also contributes to the calculation of reaction forces at the sacrum. For example when we use one of the following optimisation techniques

$f_1 = \text{Min} \{ \max |d_i| \}$ $\quad i=1, .. , n$

$f_2 = \text{Min} \{ \sum d_j^2 \}$ $\quad i=1, .. , n$

$f_3 = \text{Min} \{\sum w_i * |d_i|\}$ $\quad i=1, .. , n$

$f_4 = \text{Min} \{ w_i * |d_n| + w_2 * \max |d_i| \}$ $\quad i=1, .. , n-1$

(where |d| is the distance between the current calculated thrust line and the spine reference or central line, w is the weighting coefficient, and n is the number of forces considered) we found thrust lines which are much closer to the centre line that produce forces of approximately 1400-1600 N at the sacrum (Xiao *et al*, 1997)

In another set of calculations, when the intra-abdominal forces are reduced by 80%, (i.e. taken as 14 N at each intervertebral level) by using the same optimisation techniques, we found thrust lines which are much closer to the centre line of the spine and they even safely lied within the cross section of the arch formed by the vertebral bodies. On the other hand as Aspden explains if the weight is lifted with the spine in a stoop posture then tensile forces would be developed in the lumbar spine that are greatest along the posterior margin of the spine and could lead to damage of the intervertebral discs.

This example demonstrates that even by using the same model one can obtain contradictory results unless the spine is taken as a system. The simplified models would lead to more sophisticated ones however one should not ignore the constraints imposed by the assumptions which are not necessarily satisfied by complex structures like spine. For example, in the case above calculations are limited by the collective forces and partial representation of spinal curve. The collective forces are not representative and the stability of part of the spine does not necessarily mean the stability of the whole spine.

The spine should be considered as a system of vertebrae, discs, muscles ligaments and their interacting functional relations. The absence or malfunction of one subsystem at one level may upset the function of the whole system, not just at the level of the particular subsystem. The investigations reported in Grilli and Acar, 1997(b) reveals that when the greater anatomic details were considered in terms of distributed body weight, detailed muscle forces for the whole spine together with the spine curvature joint reaction predictions on the sacrum for the arch model lie between the values reported for the lever model and the arch model with collective loading.

It is our intention to integrate the geometric and mathematical models to produce a model of the spine which is capable of providing guidance to the designers. The design tool will be particularly useful for the design of 'non-standard' artefacts. A perfectly comfortable and safe to use equipment may be uncomfortable and even dangerous to use if the changing conditions of the spine are considered.

For example pregnant women are notorious sufferers of back pain. Changes in a pregnant woman's body such as change in the curvature of the spine, increase in body weight, change in body weight distribution and variation in muscle strength due to hormones would probably suggest specialised artefact designs for pregnant women to prevent back pain.

Elder people would probably need specifically designed artefacts due to changes in the whole body including the spine parameters as in the pregnant women's case.

The model can also be used in the design of car seats. This is another important example of application area since the long term drivers are other well-known sufferers of back pain. Limited space and muscle activities and the change in properties of spinal elements such as dimensions due to vibration and hence change in curvature of the spine have to be considered during the design.

Effective use of the spine model in practical workplace design situations requires a method of application. This will be achieved by the eventual integration of the combined mathematical model within the SAMMIE Computer Aided Ergonomics Design System.

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