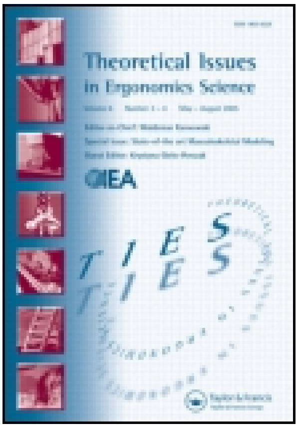

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A nonlinear contact algorithm predicting facet joint contribution in the lumbar spine of a specific person

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The objective of this study was to realistically include the effect of facet loading in an EMG-assisted lumbar biomechanical model. Most biomechanical models lack detailed facet geometry, the inclusion of cartilage, and fail to model the full lumbar spine. Several new facet-specific components were added to an EMG-assisted biomechanical model, including realistic geometry and facet-specific contact algorithms. These algorithms defined nonlinear contact between each lumbar spine facet. Subject-specific data were applied to the model for assessment. As expected, resultant disc loads were generally lower in the model with facets. This information improves our understanding of how loads are distributed in the spine, and it can lead to a better understanding of causal pathways. If we understand those pathways, we then realise how to design better ergonomic interventions.

Keywords: facets; low back pain; spine modelling; contact forces; low back disorder

1. Introduction

Facet joint degeneration has been hypothesised to be a potential low back pain (LBP) pathway (Cavanaugh *et al.* 2006). As the intervertebral discs deteriorate, the zygapophyseal (facet) joints are consequently called upon to bear a greater load. The facet surfaces also degenerate, via deterioration of the surrounding cartilage (connective tissue surrounding the joint) which can then lead to progressive joint degeneration (Yue *et al.* 2008). Unfortunately, our ability to quantify this degeneration is poor, and most lumbar biomechanical models simplify or ignore the facet geometry (Calisse *et al.* 1999, Ezquerro *et al.* 2004).

Modelling facets with detailed geometry is a clear way to better understand load paths in the lumbar spine. The facet joints have been reported to transmit anywhere between 2% and 59% of the load through the spine in previous models and experiments (Yang and King 1984, Kim *et al.* 1991, Teo *et al.* 2003, Guo *et al.* 2007). This large range indicates that load paths are not yet fully understood (Moore 1992).

Geometry is an important consideration in modelling the facets. The articular cartilage surrounding the bony portion of the facets is covered with a film of synovial fluid which facilitates the gliding movement between the articular processes and acts to consequently reduce friction (Bogduk 2005). Curvature of the articular facet surfaces varies both by the

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level of the spine as well as between subjects. In the lower levels of the lumbar spine, the facets are more likely to have a flat, planar surface (Horwitz and Smith 1940). The fact that the shape of the facet surfaces can change both between subjects and between levels of the spine becomes very important when modelling the facet contact interaction at all spinal levels. Thus, the magnitude and direction of force transmission can be complex and it is important to quantify load transmission in specific portions of the spine. One potential source of pain occurs when a superior facet bottoms out on (comes into contact with) the lamina of the inferior vertebral body (Yang and King 1984). Second, when facets are injured or deteriorating, they can cause irritation to the surrounding spinal nerves and this is hypothesised to be a significant cause of LBP (Cavanaugh *et al.* 1996, 1997, Kalichman and Hunter 2007). Hence, it is important to understand how loading occurs within the facet joint.

Biomechanical modelling is one way to understand the nature of facet contact. Many linear elastic finite element method (FEM) solutions of the spine include posterior elements (Kim *et al.* 1991, Goel *et al.* 1993, 1994, Yoganandan *et al.* 1996, Maurel *et al.* 1997, Natarajan *et al.* 1999, Teo *et al.* 2003, Rohlmann *et al.* 2006, Guo *et al.* 2007, Schmidt *et al.* 2007, 2008). The facet contact force has been chiefly represented by gap (contact) elements, which provide a restoring force when two deformable surfaces approach each other, and removes the contact force when the two surfaces separate.

One of the first steps in understanding facet contact is to understand how the synovium, cartilage and bone act as a unit and the role they play on contact forces. There are few previous models that consider the system behaviour of the synovium and the cartilage on the facets (Eberhardt *et al.* 1990). Other models are static and usually do not include muscles or use a single equivalent muscle (Kim *et al.* 1991, Kumaresan *et al.* 1998, Calisse *et al.* 1999). Also, many previous models do not include subject-specific motion or load data (Lorenz *et al.* 1983, Yang and King 1984, Kim *et al.* 1991, Teo *et al.* 2003, Guo *et al.* 2007).

The foundation for the modified facet model described here is an established electromyography (EMG)-assisted rigid body dynamic model developed and validated by Marras and coworkers (Marras and Sommerich 1991a, b, Granata and Marras 1993, 1995, Marras and Granata 1997). This biologically assisted model was developed in order to represent spine loading under dynamic, three-dimensional motion conditions (Marras and Sommerich 1991a). This model has evolved to include inputs that together predict the loading on the spine at *each* vertebral level of the lumbar spine (Knapik and Marras 2009). This model has previously used generic facets instead of personalised descriptions of the facets.

The objective of this study was to develop a contact model which more realistically assessed facet load in magnitude and direction. The model includes a more detailed representation of the forces occurring between facet joints in the spine during engaged facet contact. In order to assess the feasibility, the model was exercised with data from a subject collected under laboratory conditions where large facet contact loads would be expected (Knapik and Marras 2009).

2. Model structure

This model uses a detailed description of the anatomy of the facet. The model includes the material properties of the bone and cartilage, the radius of curvature of each individual facet contact region, and accounts for articular cartilage while considering the synovial fluid's properties that make the joint frictionless. Though a physical representation of the capsular ligament was still under development, an attempt at mimicking the support provided by all of the spinal ligaments is embedded in the stiffness parameters chosen for the discs in the model. The modelling software used for this model is the MSC ADAMS (Automatic Dynamic Analysis of Mechanical Systems, ADAMS) rigid body modelling software package. The facets were modelled in this software environment. The layer representing the cartilage was created in the Rapidform XOR (INUS Technology) solid modelling program.

The facet joints in the model comprise the homogenous properties of cortical bone covered with a cartilage layer. The material properties of cortical bone were adapted from Goel *et al.* (1994). The material properties of the cartilage were determined from Kumaresan *et al.* (1998), and were used for cartilage layers on both facets at each lumbar vertebral level.

The intervertebral discs were represented by multidimensional spring–dampers with translational and torsional stiffness and damping parameters found in the literature. Since the values derived by Gardner-Morse and Stokes (2004) were determined from motion segments with the ligaments and the facets intact, the values used in this model were reduced slightly to account for the stiffness from all of the spinal ligaments but to ignore the stiffness due to facet contact, which is treated directly as shown below.

2.1. *Lumbar spine geometry*

The geometry for the lumbar spinal segment used in this model was obtained via computed tomography (CT) *in vivo* with 0.625 mm slices. The CT image data were segmented into a solid model of the vertebral bodies. The sample subject was a 24-year-old male with no previous history of LBP. The CT scans were compiled to create a three-dimensional model of the lumbar spine from T12/L1 to L5/S1. The detailed geometry of the CT scans allows for extraction of radii of curvature at any point on the individual facets. According to the literature, cartilage thickness ranges from 0.2 (Sharma *et al.* 1995, Teo *et al.* 2003, Holzapfel and Stadler 2006) to 3.02 mm (Sokoloff 1966). Cartilage degenerates with time (E. Mendel, personal communication, 2008), so an estimate of 0.5 mm (on the lower end of the aforementioned range) was made for the thickness of the cartilage surrounding the facet joints (Natarajan *et al.* 1999). This value was chosen as a realistic approximation for the amount of cartilage in an adult, given that there is definite deterioration of cartilage over time.

During model building, the spine was rotated from the vertebral orientation with which it was built (from CT scans) to vertebral orientations derived from an upright MRI. Scans from the Fonar 0.6 T Upright^{TM} pMRI scanner (Fonar Corporation, Melville, New York) gave the benefit of more realistic intervertebral angles since the subject was in a standing rather than a supine posture.

2.2. *Articular facets*

In order to determine the components of the force equation describing facet contact, the following assumptions were made: (1) the cartilage layer has a uniform thickness of

0.5 mm over the entire facet surface; (2) the material properties of the bone and the cartilage are linear and isotropic; (3) the synovial fluid, which serves the purpose of lubricating the joint, enforces the condition of zero friction during the contact. Further it is assumed that one of the two articulating facets in each contact location has a much larger radius of curvature than the other one. This larger radius of curvature is considered infinite in calculating stiffness constants.

The contact force between two articulating facets is defined by the nonlinear function:

$$F = k \times (\text{distance})^n \tag{1}$$

where F is the contact force, k the contact stiffness, n the force exponent, and 'distance' the distance between the centroid of the intersection volume and the closest point on each of the intersecting bodies (Figure 1). The larger this distance, or *approach* as it is called in the contact literature, the closer the two bodies are being pushed together and the greater the force in the contact defined by Equation (1).

An important parameter in determining the contact stiffness and force exponent of the contact of the facet is the local radius of curvature of the contacting surfaces. During a typical simulation, two articulating facets make contact at different places at different times. Using an initial estimate for the contact stiffness and force exponent, the contact areas are noted during an initial simulation. Then the radius of curvature for each local contact surface area was determined by fitting spherical patches of surface to the geometric surface model of the facet in each local contact area. The fitted values of the curvature at all of the contact points on a single facet were averaged and used in the calculation of single contact stiffness (k) and force exponent (n) values for that particular facet. Note that the individual radii of curvature of the different patches on a single facet did not vary very much from these calculated averages.

Several studies using mathematical descriptions of contact force between deformable bodies were reviewed in order to determine how to obtain the stiffness, k, and exponent, n, in the ADAMS function. The most common model for contact of elastic bodies is the

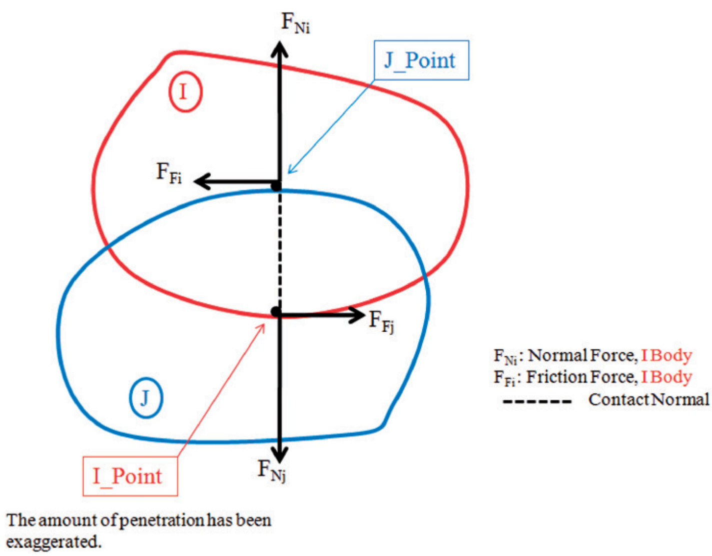

Figure 1. Contact description relating contact stiffness and contact approach to contact force.

Hertz model (Johnson 1985). It estimates that locally near the contact, each contacting body may be modelled by an elastic half-space loaded over a small region on its surface, and the contact area between the two surfaces is assumed to be elliptical in a three-dimensional setting (sphere on sphere or sphere on a flat plane). The theory assumes that the two contacting surfaces contact initially at only one point and takes into account the actual relative curvatures of the two contacting bodies. The model results in, amongst other things, a closed form formula for the approach as a function of the applied load pushing the two bodies together and the elastic properties and curvatures of the two contacting bodies. An extension of the Hertzian contact theory to coated bodies (Liu *et al.* 2005), in which the Hertzian formulas are modified by adjusting certain constants to account for the coatings, was selected for use in this study. In the facet contact problem, the bone of the articulating facets was considered to be the substrate of the coated body and the cartilage was the coating. Figure 2 shows the current contact problem description, where one facet is modelled as a sphere and the other as a plate and both bodies are coated. Given a substrate of cortical bone and a coating of articular cartilage, several equations were adapted (Liu *et al.* 2005). The contact approach (Equation 1) can be solved for via these equations given the coating thickness, material properties, radius of curvature and applied load.

The next steps in model building were to use the information about the specific facets (material properties, cartilage thickness and radii of curvature) in the Liu model to determine force constants to be used in the contact force definition. An iterative process was used to determine the force–displacement relationship. This procedure started with loading data from the literature (Yang and King 1984, Kim *et al.* 1991, Teo *et al.* 2003, Guo *et al.* 2007) which were used as inputs into the modified Hertz formulas (Equation 1). Results show individual stiffness values and exponents for each of the facet contact regions in the lumbar spine.

2.3. *Experimental conditions*

In order to explore whether the model would produce detailed and potentially realistic results, data from one subject were chosen from a combined laboratory/modelling study assessing biomechanical loads during pushing and pulling tasks (Knapik and Marras 2009). Previous model results indicated prolonged contact between facets during pushing tasks and intermittent contact (disengaged facets) during pulling tasks. This previous study also found significant shear loading at the upper levels of the lumbar spine that are

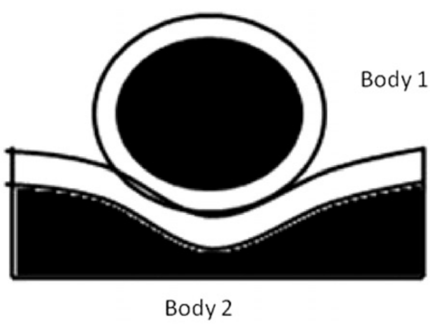

Figure 2. Representation of problem description, contact between two coated bodies. In this study, the bodies represent two facets covered in cartilage.

expected to load the facets. A number of these tasks which were recorded and studied were chosen for use in this study in order to demonstrate a range of lumbar motions. Several tasks were chosen for which the spinal loading was at the lower end of all of these results, and several were chosen for which the spinal loading was at the upper end of all of these results. Data from the selected tasks were used to derive the inputs to the present model.

The test conditions were a heavy load (40% of the subject's weight, 29.9 kg), a light load (20% of the subject's weight, 15.0 kg), at the highest handle height for pulling (80% of subject's height, 143.2 cm) and at a low handle height for pushing (50% of subject's height, 89.5 cm). The model was used to: (1) compare facet contact forces between a pushing and pulling task and (2) compare spinal loads at each intervertebral level in the lumbar spine model with modified facets to spinal loads in a model with the facet contact forces turned off.

Each of the trials from the push–pull data was run through the model twice. First, the facets were active and the model was controlled by a torque designed to minimise the error in matching the spinal motion to the push–pull data from which it was obtained. The intervertebral angular velocities were then recorded at each disc and used to drive the motion in the second simulation with the facets inactive. In this way, the motion of each disc and segment was exactly the same between the two simulations and all changes could be contributed solely to the addition of the facets to the model.

3. Results

Overall, the model performed well and matched the intuitive expectations. Allowing the facets to bear load took a large portion of the load off of the discs at most of the vertebral levels of the spine. Figure 3 shows the percentage of difference between the maximum resultant disc loads in the model *with* facets compared to the model *without* facets for trials pulling the heaviest load. Negative values indicate that the resultant load in the model *with* facets was lower than the resultant load in the model *without* facets, thereby signifying that the facets reduced substantially the loading in the discs. In most of the trials, there was a

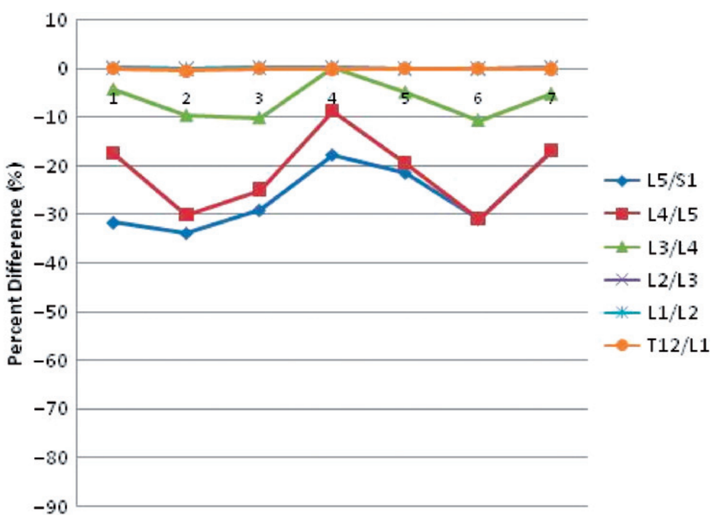

Figure 3. Percentage of difference in disc loads between facet and nonfacet model prediction, by trial (study conditions were pulling, the heavy load).

minimal difference between having the facets included or excluded in lumbar levels T12/L1 to L2/L3 (where the loading was not reduced by the facets).

The model predicted that load-sharing between the discs and the facets was complex for this specific subject. The facets were engaged most of the time during torsion and lateral bending, however this did not always correspond to a decrease in lateral and AP (anterior–posterior) shear loading of the discs. The peak resultant disc force and the peak facet contact force were calculated for each trial, and then averaged to obtain an overall value for both pushing and pulling. Figure 4(a) and (b) compares the effects of pushing

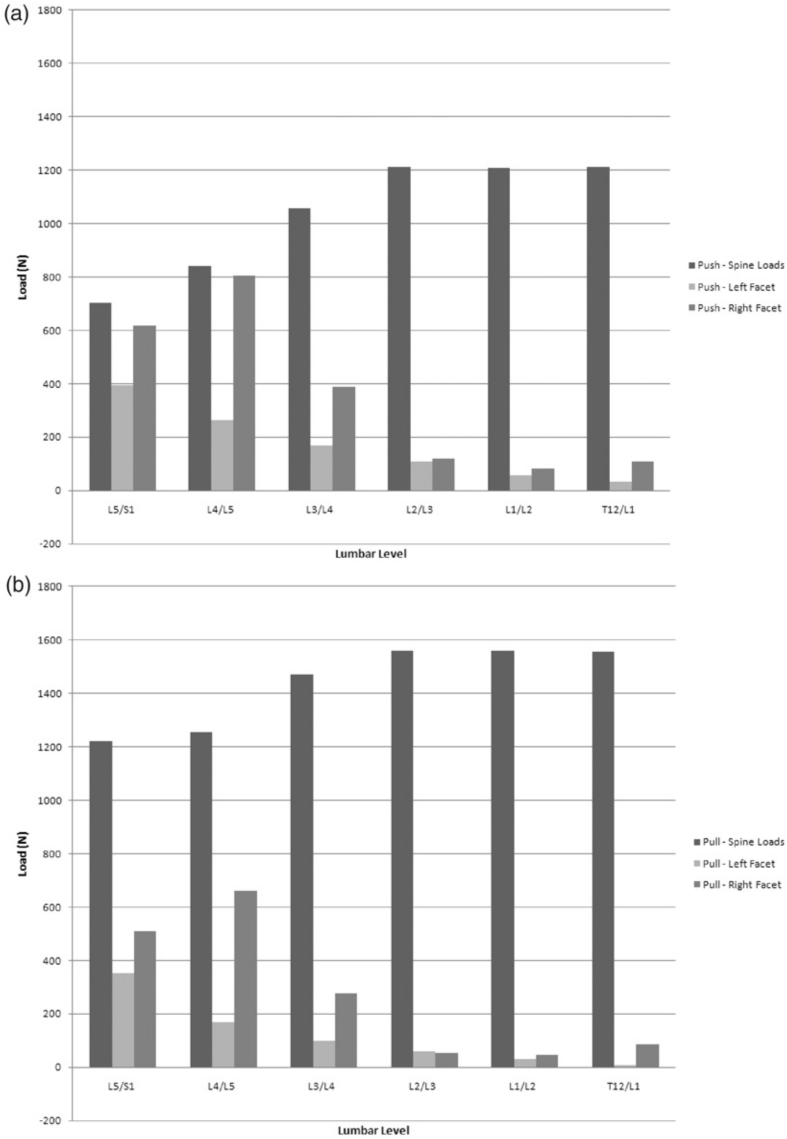

Figure 4. (a) Comparison of magnitude disc load to magnitude facet load, pushing. Loading of the discs was lower than in pulling and (b) Comparison of magnitude disc load to magnitude facet load, pulling.

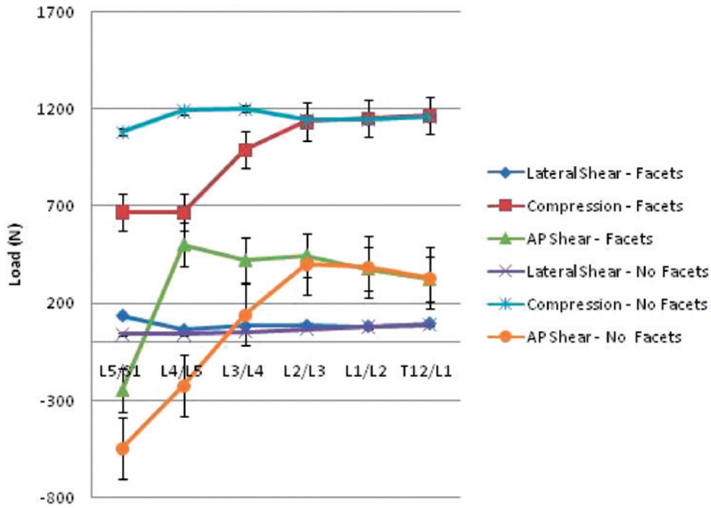

Figure 5. Superior disc loads with and without facets, for a pushing task (lateral shears shown as absolute values).

and pulling on the disc loading during the trials with facets. As expected, the overall loading of the disc was lower during pushing where the spine was more lordotic and the facets were engaged, bearing some of the load for the disc (Figure 4a). The disc loads were higher (~1600 N pulling and ~1200 N pushing) at the upper levels of the lumbar spine, which is reasonable because the facets bore a greater load at the lower levels (~1200 N pulling and ~700 N pushing). Without the facets, the loading on the discs stayed fairly consistent at 1600 N pulling and 1200 N pushing through all the levels of the spine.

Figure 4(a) and (b) also indicates the left and right resultant loading at the facets. As expected, when the disc loads decreased from T12 to S1, the facet contact loads increased from T12 to S1. At their peak, the left facet contact was near 800 N. The only exception was for pushing at L4/L5, where the contact force was higher at L4/L5 than at L5/S1. The contact load was greater in pushing than in pulling for all levels of the spine.

Figure 5 shows the peak disc loads averaged over each condition. In general, disc loads were higher in all directions at the upper levels of the spine and decreased going downward towards L5/S1. From levels L3/L4 to L5/S1, where the highest facet contact was occurring, the compression on the disc was greater when the model had facets turned off. The AP shear, however, was lower when the facets were off at these same levels. This was one indication that the direction of the offloading of the disc may not be clear, and so it was beneficial to present a resultant disc load as shown in Figure 4(a) and (b). The lateral shear did not change much when the facets were included or not included, but the largest difference was found at L5/S1 where the lateral shear was slightly higher with the facets turned on.

Figure 6 shows the results for one of the conditions with the lowest loading of the trials run in the facet model. This trial demonstrates a check that the model behaves as expected. It displays time-dependent, processed data at L5/S1 (including AP and lateral shears, and compression with and without facets as well as left and right resultant facet contacts) from a pulling trial at the lowest handle height and with the lightest load. This figure shows the behaviour of the model as the spine moved into a posture where the facets were in contact.

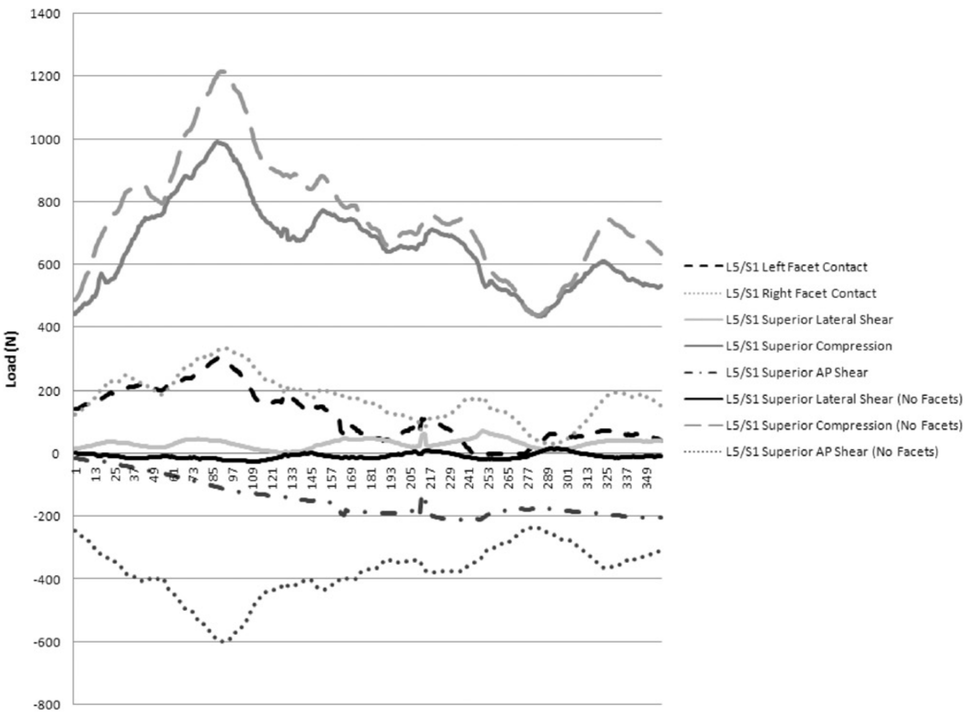

Figure 6. Disc and facet loads at L5/S1 for a pulling task, study conditions were 50% handle height, light load.

As expected, the largest difference between the disc loading in the trial with facets compared to the trial without facets occurred at the point in time where the facet contact load resultant was the highest. At 277 s, the compression value with facets was almost identical to that without facets. Correspondingly, it can be seen that the left and right facet contact forces at that point in time were each less than 40 N.

As an example of this personalised detailed model, in a trial with the highest spinal loading, the facets bore the least amount of load at L1/L2, accounting for only 0.54% of the total load through that motion segment. The lower levels, especially L4/L5 and L5/S1, accounted for around 40% of the load.

4. Discussion

This particular model is specific to the geometry of only one subject, so the model results should be treated as preliminary and as having the purpose of demonstrating the benefits of a detailed, geometry-specific facet model. Modelling with this degree of specificity at present requires a detailed spinal geometry only available via CT scans. While this procedure is infeasible for a population of subjects, CT data from one subject were available to employ in this modelling exercise. Hence, the results presented here are not intended to generalise spine loads representative of a population of subjects during pushing or pulling activities.

This spine model including facets and their contacts represents a very complex interactive system, and it is appropriate to treat it as such. Each element included in the model must work together with the others to embody the complex movements of a human spine *in vivo*. The dependent variables in testing the model are facet loads and intervertebral disc loads, whose relationship depend strongly on the type of loading and the spine geometry. This modelling study has revealed some interesting insights relative to loading. In particular, the facet loads on the left and right sides of the vertebra suggest very unbalanced loading of the spine for this subject. At L4/L5 especially, the majority of the loads transferred through the facets followed the path on the right side of the joint. The magnitude of this contact force was also higher than at any other levels. The right L4 inferior facet articulating surface began in contact with the right L5 superior facet. At the greatest point of extension in the trial, the L4 inferior facet slipped outside of the L5 articulating surface to come into contact with the L5 lamina. For this subject, this contact continued throughout most of the remaining length of the trial and was the primary source of the highest contact loads at this level. This may suggest some new avenues of exploration in understanding LBP causality.

The circumstance of a facet bottoming out on the lamina below is not a new discovery (Yang and King 1984, el-Bohy *et al*. 1989). Yang and King (1984) cited this as a serious possible site for origination of back pain, given that this area is richly innervated and the contact might be irritating adjacent nerves or wearing away at the capsular ligament. This model was not designed to account for cartilage-to-bone or bone-to-bone contact, but solely the cartilage-to-cartilage contact that occurs between the two facet articulating surfaces. The material properties of the contacting bodies are integral in defining the stiffness coefficients of the contact force; therefore, they should be tailored to account for the cartilage-to-bone contact that is happening during facet-to-lamina contact. Though changing the stiffness values might affect the *magnitude* of the force at this location, it is still expected that the model will show similar facet-to-lamina contact behaviour, which would therefore serve as an excellent indicator of what may be a very real source for LBP. Another subject might have lumbar geometry such that the facets only contact each other in extension, rather than the adjacent lamina. This could also be determined in the same manner by viewing recorded trials and observing contact load magnitudes at each level.

One of the other factors that appeared to have a great effect on loading patterns was the initial orientation of the vertebrae. In this subject, lower measures of facet contact force were observed at the upper levels of the spine. Recall that the spine was rotated to MRI vertebral orientations during model building. This analysis indicated that T12/L1 and L1/L2 were extended slightly and therefore come into contact less throughout most of the trials than at lower spinal levels. This could be the trend with further subjects, depending on whether or not the subject's spine was pushed past normal lordosis with the use of the rolled towel while the CT scan was performed.

The results from the test of the model provide an indication of the role that the facets can play in load bearing under these types of conditions. The highest concentration of contact force was found at L4/L5 and L5/S1. This is logical since these levels begin in contact and remain in contact throughout a trial. Since a contact force was defined for both the left and right sides of each vertebra, this might indicate which portion of the disc is more likely to degenerate first. In the instance of L4/L5, the right facet contact force was always much higher than the left. This could indicate that the right portion of the disc is offloading some of its force to the facets, causing tension on the left side of that disc and creating a torque in the disc.

In the push–pull study from which the data run in this model was obtained (Knapik and Marras 2009), the researchers found values for AP shears at the upper levels of the lumbar spine that approached or exceeded a proposed safe limit of 1000 N (McGill 1997).

However, the results for AP shears at these levels in this model indicate that the offloading of the discs is more complex than was believed previously. In many of the trials with the highest facet contact loads, the AP shears actually increased. In pushing, the average AP shear at L5/S1 of the peak values from all the trials increased from about 300 to 600 N and in pulling the difference was slightly greater, increasing from about 350 to 750 N. It has been shown that the contact loading at L5/S1 was also very high, at about 40% of the total load going through that level. Therefore, it is possible that rather than the facets bearing some of the shear load on the disc, they are actually bearing more of the compressive force on the disc. In turn, the facets at L5/S1 were stiff enough that they actually generated shear forces by pushing the disc anteriorly or posteriorly. It should also be noted that using the current facet model based precisely on subject geometry, the shear loading never reached the same dangerous magnitude as in the previous study (whether the facets were on or off). The average AP shear values approached 800 N but never exceeded 1000 N. These findings suggest the potential utility of such a subject-specific model in assessing spine loading and risk during various activities.

The resultant loading on the disc was an important measure because of the complex direction of load transmission through the facets. Though the average peak AP shear in both pushing and pulling was higher with the facets included, the average peak resultant disc loads were higher without including the facets. This indicates that the facets indeed played an integral role in the loading of the spine for this subject.

Previous experiments have measured the facet *normal* contact force via a pressure sensitive film or a load cell. For example, Lorenz *et al.* (1983) found in their static loading measurements that the facets at L2/L3 bore 47% of an applied compressive load of 392.4 N and 57% of an applied load of 196.2 N. At the point of maximum loading in the current model trial, the facets bore 39% of the loading through L4/L5 and 43% of the loading through L5/S1. The values from the current model are seen to be quite close to those found in the Lorenz study. A more fitting comparison might be to the results of Sharma *et al.* (1998) which are based on the FEM analysis of an L3/L4 motion segment. The applied load in that study was a torque rather than an axial compression. An applied axial torque of 10 Nm yielded facet loads of 59% of the total applied load in this model. The data reported in these two studies serve as indication that the current model results fall into a plausible range within the experimental and modelling literature.

Other studies (Lorenz *et al.* 1983, Yang and King 1984, Kim *et al.* 1991, Teo *et al.* 2003, Guo *et al.* 2007) have yielded a wide range of percentage of load transmission values in the literature. Results from this study are generally on the higher end of what has been found in the previous studies. Differences between the current results and any of these studies can be accounted for primarily by the different geometries of the tested specimens and the model geometry. Segment stiffnesses can vary significantly between subjects as well as between a spine *in vivo* and a cadaveric spine. Facet curvature is very specific to an individual, and therefore could account for both the magnitude and direction of the load going through the facet joints. Cartilage and bone thicknesses and material properties will also change amongst different spines, especially with age. Most studies have tested or modelled only one or two motion segments. On the other hand, the current full lumbar spine model behaves as a system, each level having some influence on the next; therefore it is difficult to extrapolate comparable results from a study with a single motion segment. The current model is dynamic, EMG-driven and very complex (Knapik and Marras 2009). The complexities that have validated it in previous studies might also account for differences between these model results and those from other facet models. Because each model is geometry dependent, results should be taken to represent the loading for this specific individual. Once again, the potential value of such a subject-specific model is demonstrated through this example.

Given the objective of this study, we should acknowledge the potential limitations. First, and most importantly, the data evaluated in the modified facet model came from only one subject so we cannot generalise the population. Next, there was also no physical representation of the capsular ligament in this model. The geometry would be more complete with all of the spinal ligaments included. The stability provided by these ligaments would surely affect both disc and facet loadings. These ligaments are currently under development and will be added to the model.

Nonetheless, this modified biomechanical model which includes facet contact shows promise in several respects. First, the model accounts for specific, individual geometry of the spine because of the high-resolution CT scans used to build the model. The realistic asymmetries of an individual's spine lead to the uneven loading patterns on the facets, and helps indicate where degeneration is occurring both in the facets as well as in the adjacent intervertebral discs. Another advantage of this model compared to many models in the literature is that it encompasses the whole lumbar spine. Finally, facet contact forces calculated in this model are more realistic than previous models because the model is biologically driven and the data used to drive the model came from dynamic tasks *in vivo*. Most facet loading studies to date yield contact forces from a single load applied to a cadaveric spine, and it is clear that this may not be an appropriate measure of the actual facet loading *in vivo*. It has been shown that including facet loads at each level in a model which evaluates dynamic tasks leads to a more detailed understanding of facet and disc degeneration and facet pain.

5. Conclusions

(1) There is value in precise modelling of the facet joints. Preliminary results suggest that the facets play a major and complex role in load support.
(2) For this subject, the facets bore a large portion of the load though the lumbar spine, reaching more than 40% at L4/L5 and L5/S1.
(3) The direction of the offloading of the intervertebral discs may not be as straightforward as previously hypothesised. Including facets in the model did not necessarily indicate that the AP shear loads borne by the disc decrease. However, from L3/L4 to L5/S1, the resultant load was always lower through the disc when the facets were turned on to provide an alternate load path.
(4) In this young, healthy, 24-year-old male, the percentage of load transmission through the facet joints is comparable with values reported in the literature for both static compressive load testing as well as modelling techniques with applied moments.
(5) The facets in this single subject-specific model bore a greater amount of load in pushing than in pulling tasks, resulting from a flexed posture during pushing that disengages the facets.

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Relevance to ergonomics theory

It is hypothesised that facet joints play an important role in the causality of low back pain. Most models used in ergonomics simply look at spine load and ignore the facets. However, we know that facets are pain generators so it is important to determine how much loading they might bear.