Developing a 3D Multi-Body Model of the Scoliotic Spine with Lateral Bending Motion for Comparison of Ribcage Flexibility

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Abstract: In this paper, a fully discretized bio-fidelity simulation model for bio-mechanical (kinematic) analysis of scoliosis for various patients was developed to analyze forces on vertebrae, loads acting on the intervertebral disc, corresponding angles between vertebrae, and tension in the spine muscles during lateral bending. It was further developed to study the movement limitation and muscle activation of a scoliotic subject. This system was built by using the commercial software LifeMOD. The whole spine (pre-defined by the software) was discretized into individual vertebra segments with rotational joints representing intervertebral discs. In this study, two female subjects (with 40±1.0 kg weight and 154±3 cm height), one with normal spine and the other one with scoliotic spine, were asked to do lateral bending and bend as far as they can. Motion capture data of these two subjects was obtained. Next, motion capture data was assigned to the model using a motion agent set and the inverse dynamics simulation was performed to simulate the complicated multi-body motion of lateral bending. The mobility of the ribcage, activity of muscles which are important in the lateral motion of body, as well as joint angles were analyzed using the developed simulation model. According to the obtained results, the mobility of the ribcage in scoliosis model was less than that of the normal model with the same anthropometric data. This finding is in direct agreement with the qualitative experimental results done by other researchers.

Keywords: Lateral Bending, Musculo-Skeletal, Multi-Body Model, Scoliotic Spines, Spinal Instrumentation and Fusion

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Biographical notes: Kh. Hajizadeh is pursuing her PhD in the department of Mechanical Engineering, in NUS. **I. Gibson** is an Associate Professor at the NUS, and he has been a pioneer in the development of Additive Manufacturing Technology. **G. Liu** is an Associate Professor at the department of Orthopaedic Surgery at NUS. His research interests are in degenerative spinal disease.

1 INTRODUCTION

The spinal column plays very important roles during human daily activities. It not only protects the spinal cord, but also supports the body frame (weight of the body and skull) and acts as the structural foundation for ribcage, shoulders, and pelvic. In order to function properly, the spine should maintain its normal shape. However, there are several conditions in which the spine is abnormally curved or aligned. Scoliosis is defined as a lateral curvature of the spine with rotation of the vertebrae about its axis. In clinical spinal problems, scoliosis is a less common but more complicated disease compared to low back pain, whiplash injury, etc.

Severe cases of scoliosis can be treated by spinal instrumentation and fusion to stabilize and straighten the curve in 3D space. Currently, decision on the number, type, and shape of the implants mainly depends on surgeons' experience. This treatment technique suffers from lack of accurate biomechanical understanding of the scoliotic spine and also predictive information about the general outcomes of the operation. This necessitates understanding the complicated biomechanics of the scoliosis spine and developing a realistic biomechanical model of the scoliotic spine capable of simulating trajectories and forces for various postures and patients. This model can assist physicians in pre-operative planning and post-operative treatments.

In order to gain better understanding of the biomechanics of the spine, a large number of experiments (both in-vivo and in-vitro) have been conducted over the past decades [1-4]. In most of these works, because of the complexity, the whole spine with its individual parts (e.g., vertebrae, disks, ligaments) was rarely modeled. Instead, researchers tried to consider the spine as major separate rigid bodies (e.g. cervical spine or lumbar spine) and investigated the motion of each section [5], [6]. However, since all the functional spinal regions are connected and the spinal column normally works together with the lower limbs and upper extremities, such assumptions failed to present a realistic model of the spine.

In order to accurately investigate the effect of spine deformities on its motion performance (basic body motions and muscle activities), in the current work a virtual bio-fidelity musculo-skeletal multi-body scoliotic spine model is being developed for integrative study of the whole spinal column together with the limb kinematics would be very valuable.

This scoliotic spine simulation system was built and developed in a number of stages within the recent years. The first stage was completed by Tay et al. in 2008 [7]. This work was basically focused on developing a realistic and detailed normal spine model. This model was developed by discretizing the whole spine and adding necessary ligaments using the commercial software LifeMOD (by LifeModeler Inc., a front-end to MSC Adams). The model was able to simulate kinematic behavior of musculoskeletal forms. Although more realistic, it didn't stabilize when lateral or backward forces were applied on certain vertebrae. To address this problems, Huynh et al. developed a more detailed normal spine [8]. To stabilize the model, interspinous, flaval, anterior/posterior longitudinal and capsule ligaments were created.

These guide segmental motion and contribute to intrinsic stability of the spine by limiting excessive motion. Huang et al. further developed this model to investigate the effect of different sitting postures on spinal angles and joint forces in the pain-free human body by the application of motion data capturing combined with virtual musculo-skeletal modeling [9]. Hajizadeh et al. developed a multi-body biomechanical model for examining the scoliotic spine [10]. The obtained model was tested and validated by investigating three hypothetical scoliosis models with three different Cobb angles of 38°±2, 52°±2 and 62°±2 and comparing their behaviour with a model of a healthy spine as a control. For validation, a horizontal force from posterior to anterior in the sagittal plane was applied onto vertebra T7.

The resultant axial and shear forces as well as the momentum about the lumbo-sacral (L5/S1) joint were calculated. The results were compared with other models [11] and also with those of the normal model. Effect of asymmetry on the scoliotic spine was evident in the resultant forces and momentum and the results were consistent with those of the other models.

In the current work, this model is used to study the role of severity of scoliosis (Cobb's angle) on the motion limitations of the body. In this paper, details of developing a detailed spine model using LifeMOD software and conducting motion capture experiments are explained. This model was used to study the side-bending motion of a scoliosis patient and the results have been compared with those of a normal subject with similar anthropometric specifications.

2 MATERIALS AND METHODS

2.1. Human subjects

In this study, two female models were used, one with normal spine and the other with right scoliotic curvature in thoracic region of the spine. The anthropometric data of two subjects were depicted in Table 1. Subjects gave informed consent to participate in this experiment, which was approved by The University NHG Domain Specific Review Board

(DSRB). The subjects were instructed to do a lateral bending motion on the ground.

Table 1 The anthropometric data of two subjects used in the experiments

Subject	Age(years)	Height(cm)	Weight(kg)
Normal	20	157	40
Scoliosis	15	151	41

2.2. Motion capture data

A VICON motion analysis system was used in this experiment. The motion agents have the effect of guiding the model to track the segment motion contained in the motion input file. Plug-in gait marker protocol was applied during the motion capturing, in which 35 markers were attached at specific places on the subject's body. These small markers can reflect the strobed light back into the VICON system cameras. Based on this information from all the cameras, the location of each marker is calculated and highly accurate 3D trajectories are established.

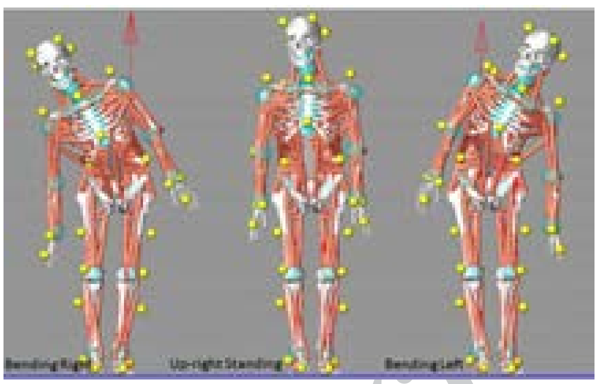

Fig. 1 Details of the position of the motion agent markers and lateral motion in sagittal plane

The subject was asked to perform lateral bending motion in a serial manner, including left bending, up-right standing and right bending (Figure 1). Three separate trials of each action were conducted to obtain the dynamic motion capture data. Plug-in Gait modeling in the VICON NEXUS system generated the human body modeled segments according to the anthropometric input and the virtual marker trajectories which indicate kinematic and kinetic quantities, such as angles, moments etc.

The output angles for hip and thorax in the lateral plane were measured and calculated by comparing the relative orientations of two related segments in the plug-in gait model.

2.3. Musculo-skeletal Human-Body Modeling

In this phase, the SLF file which is exported from the VICON NEXUS system is used to create the human body model from measurements; joints from joint data, posture from posture data and motion from recorded motion data. This file contains information on the subject name, gender, age, height and weight. LifeMOD uses this information to extract body segment measurements and mass properties from its internal anthropometric database.

The motion trajectory data is included in the SLF file as well. The motion data (MOCAP) for the lateral bending motion is imported into the model and used to drive the motion agents created on the models.

For this model, passive joints will be created for the inverse-dynamics simulation. The passive joint consists of a tri-axis hinge joint (3 DOF) which includes angulation stops, stiffness and damping torques. These types of joints are used primarily to stabilize the body during the inverse-dynamics simulation. They are later removed and replaced with Servo-type torque generators for the "trained" phase.

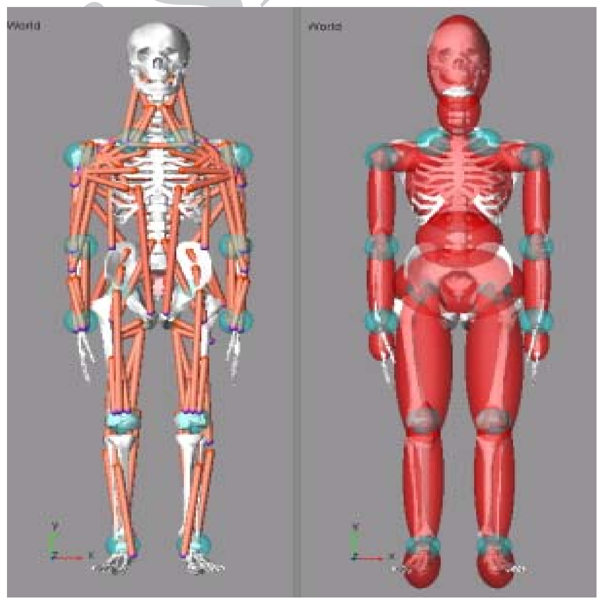

Fig. 2 The base musculo-skeletal model

The basic model in LifeMOD consists 19 body segments, represented by ellipsoids which include characteristics like mass, principal moments of inertia, location of center of gravity, and orientation of the principal axes. In addition, the model consists of 118 muscles attached to the body at anatomical landmarks. Figure 2 shows the base model. For the discretized spine model, the spine is refined into individual vertebra segments, cervical (C1-C7), thoracic (T1-T12) and lumbar (L1-L5). The rotational joints representing the intervertebral discs were added to the model later. Fig. 3 shows all ellipsoidal segments of 24 vertebrae in the cervical, thoracic and lumbar regions after discretizing in a lateral bending posture.

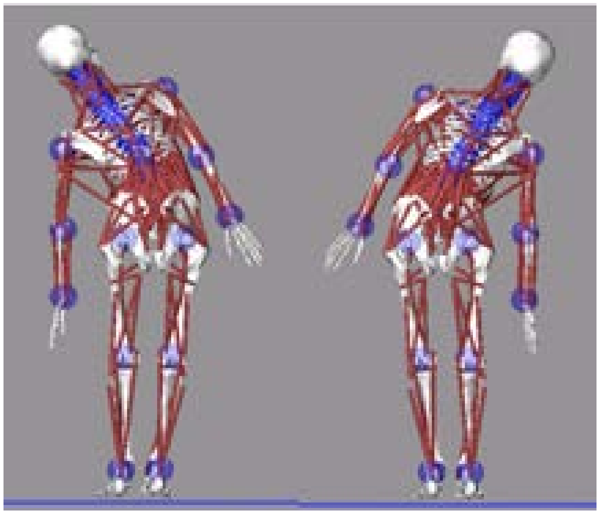

Fig. 3 Back view of the complete discretized spine model in lateral bending motion

With the newly created vertebra segments, the muscle attachments to the original segment must be also reassigned. The physical attachment locations will however remain the same. The individual joints representing intervertebral discs between vertebrae were then created. These joints were used in an inverse dynamics analysis to record the joint angulations during simulation. Although they may change after applying motion agents to the model, the properties of the joints in the literature [12-17] are used as initial values.

To stabilize the spine model, 5 types of ligaments and 6 types (sets) of lumber and abdominal muscles were implemented to the model. The name of the ligaments and muscles were shown respectively in Tables 2 and 3. These ligaments and muscles surrounding the spine guide segmental motion and contribute to intrinsic stability of the spine by limiting excessive motion. The initial stiffness of these ligaments can be referenced from [18, 19]. They may also change after applying motion agents to the model. These guide segmental motion and contribute to intrinsic stability of the spine by limiting excessive motion.

Table 2 Ligaments

Ligaments			
interspinous	flaval, anterior/posterior	longitudinal	capsule

Table 3 Abdominal and Lumbar muscles

Abdominal muscles		Lumber muscles			
Obliquus Internus	Obliquus Externus	Psoas Major	Erector Spinae	Multifidus	Quadratus Lumborum

After building the model, the Motion Agents of the subjects were added to drive the muscles. The experiment was conducted using the following steps:

1- Being in a up-right standing posture for 1 second

2- Bending left side and stay 1 second in bending posture

3- Going back to up-right standing posture

4- Bending right side and stay 1 second in bending posture

5- Going back to the up-right standing posture.

It is noteworthy that in this experiment, the subjects are asked to bend as much as they can. To produce smooth simulations for both the inverse-dynamics and forward-dynamics simulations, an equilibrium simulation was performed to equalize the forces in the model. These forces occur due to misplacement of the contact ellipsoids, balancing the preloaded soft tissues, etc. This is a dynamics analysis which holds the positions of the data-driven motion agents fixed, while finding the minimum energy configuration in the springs of the motion agents.

After equilibrium analysis, the contact constraint was added to the simulation and inverse dynamics was performed for the each simulation model. With the joint angle history recorded from the inverse-dynamics simulation, it may now be used in a proportional-derivative controller to produce a torque to recreate the motion history. During this analysis the muscle contraction histories will be recorded as well. The process entails removing the Motion Agents and updating the Joints to include the proportional-differential (PD) controllers or "trained" joints and the muscles to include the proportional-integral-differential (PID) controllers or “Trained” muscles. With the trained joints and muscles based on motion recorded from the inverse-dynamics analysis, the model is now ready for forward dynamics simulation.

Therefore, the lateral motion in normal and scoliosis models were simulated and the force and torque distribution of scoliosis model with normal one at lumber joints were compared in this study.

3 RESULTS AND DISCUSSION

3.1. Investigating the mobility of the scoliosis spine

Based on our modeling method, the final detailed scoliotic spine model was obtained. All models were driven to demonstrate lateral bending motion. From the results obtained from motion capture data, maximum bending angle in normal case was 10 degree higher than that of the scoliosis case. In other words, Rib or spine flexibility in normal case is higher than the case with scoliosis and it confirms that scoliosis subjects experience less mobility compared to normal ones.

Figure 4 shows the thorax angle in lateral bending motion for normal and scoliosis subjects. The angular displacement results of the ribs in scoliosis and normal models from multi-body simulation system was also in agreement with the previous findings which predicted less flexibility of the scoliotic spine as compared to the normal one. In addition, as depicted in Figure 5, hip angle in frontal plane of the normal model is much higher than the scoliosis model in the maximum left and right bending. It is also in agreement with this claim that the mobility of the scoliosis spine is less than the normal one.

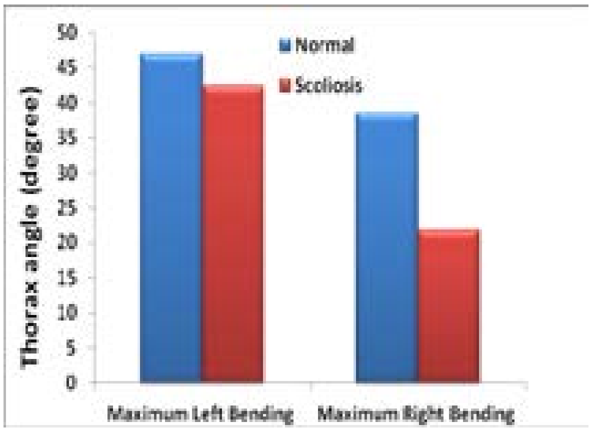

Fig. 4 Thorax angle in Lateral bending motion

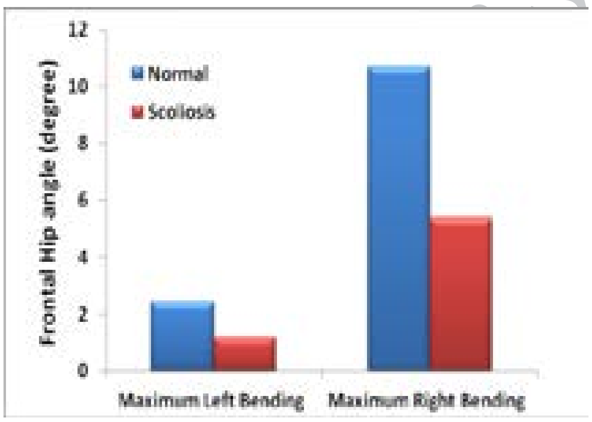

Fig. 5 Hip angle in frontal plane in lateral bending motion

3.2. Evaluation of the muscle activity in lateral bending motion

The strength of a muscle is the maximum force that can be produced by this muscle. The activity represents the percentage of the maximum force that the muscle is producing. This activity can be assumed as a constant for the whole muscles considered. Therefore, the muscle activity is defined as a value between 0 and 1. When the load exceeds the strength of the muscles in the model, the inverse dynamic analysis returns maximum muscle activation 1. If the muscle configuration is insufficient to balance the load even with overloading of the muscles, the inverse dynamic analysis will fail. This means that omission of muscles in the model may lead to inability to complete the analysis, even when these muscles can be presumed not to carry loads that are important to the overall balance of the spine.

From results obtained from the simulation; the muscle activation value in scoliosis model is higher than that of the normal one. One set of the muscles most affected by the changes in posture were the abdominal group muscles. Figures 6 and 7 shows the neck muscle group activation in left and right side of the human model in left bending, standing and right bending motion. The Neck muscle group includes: Scalenus Anterior, Scalenus Posterior, Scalenus Medius, Splenius Cervicis, Splenius Capitis and Sternocleidomastioeus muscles.

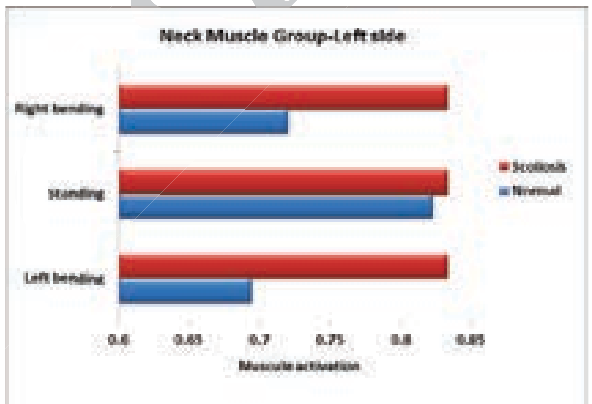

Fig. 6 Neck Muscle Group activation in left side of the human body

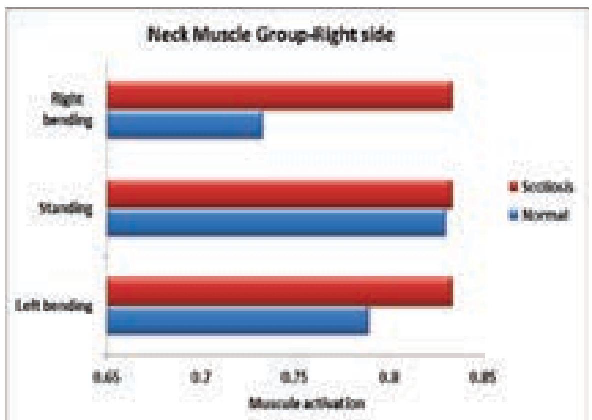

Fig. 7 Neck Muscle Group activation in right side of the human body

Figures 8 and 9 shows the Abdomen muscle group activation in left and right side of the human model in left bending, standing and right bending motion. The Abdomen muscle group includes: Rectus abdominis, Obliquus Exenus Abdomial and Obliquus Internus abdominal muscles. As can be seen in these figures, duo to the asymetry of the spine, the muscle activation in the scoliosis model is much higher than that of the normal one.

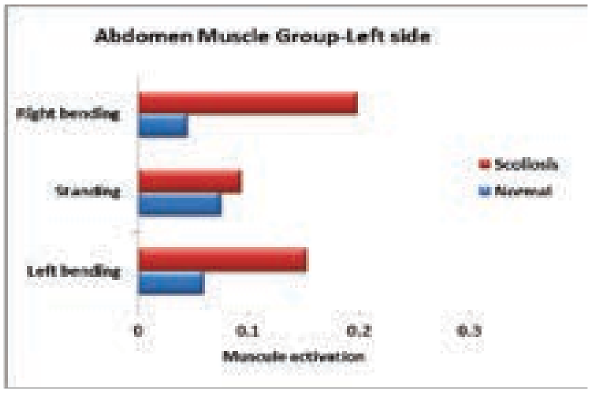

Fig. 8 Abdomen Muscle Group activation in left side of the human body

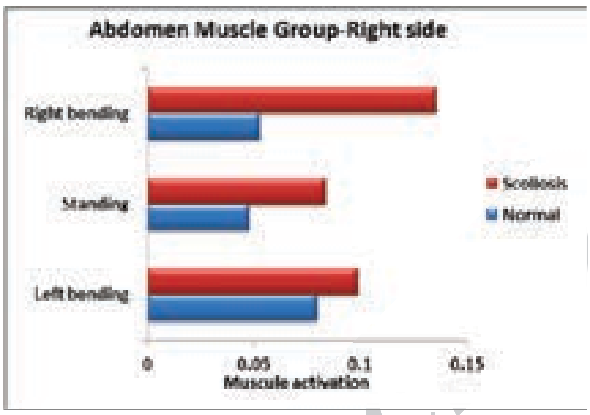

Fig. 9 Abdomen Muscle Group activation in right side of the human body

3.3. Comparison of the force and torque distribution in lumber part of normal and scoliosis model

The average joint forces and joint torques for lumber region during left bending, standing and right bending posture of scoliosis and normal models were found in this study. Magnitude forces of lumber region in right and left bending and standing posture were shown in Figures 10-12. As can be seen in Left bending motion, the magnitude force of scoliosis model in L3 to Lower torso is near to the value of the normal model. But in standing and right bending motion the magnitude force value in all lumber joints of scoliosis model are much higher than that of the normal model.

This may cause pain during bending for a scoliosis subject. This may be due to the left curvature that the scoliosis model has in the lumber region. The subject who participated in this study had two curvatures in her spine, one in the right side in the thoracic region and the other in left side in lumber region. When the subject tends to the left side, the curvature in the lumber side leads to compression of the vertebrae in this region.

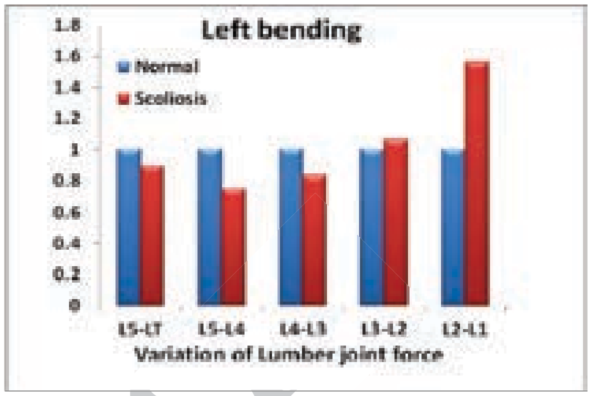

Fig. 10 Lumber joint forces in left bending motion

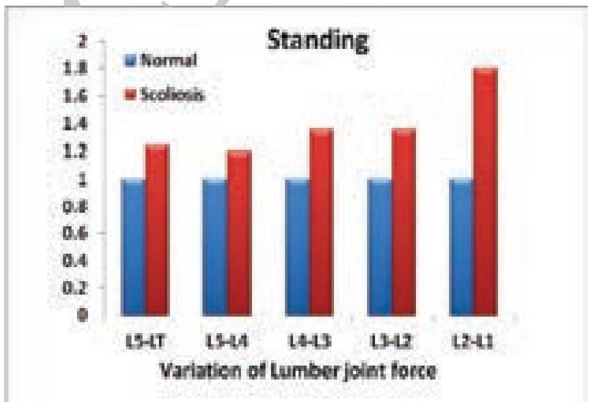

Fig. 11 Lumber joint forces in Standing Posture

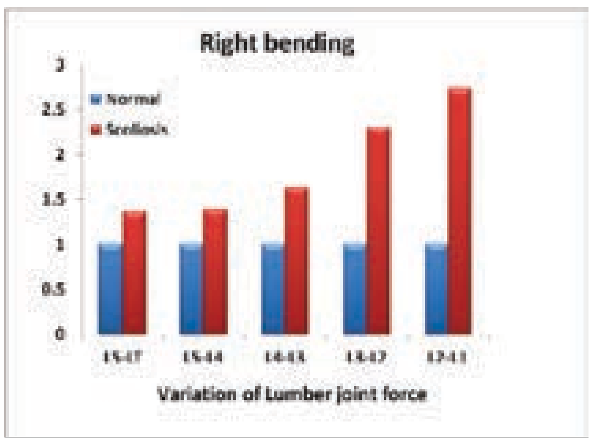

Fig. 12 Lumber joint forces in right bending motion

Generally in the lateral bending motion, the joint forces were increased from L1 to L5 in both models. However, the magnitude of the forces in scoliosis model is much higher than that of normal one. Figure 13 shows that the lumber joint force in scoliosis model is between 1 to 2 times of that of the normal one. As can be seen in this figure, magnitude force of the scoliosis model in lumbo-sacral joint is more than 2 times that of the normal one.

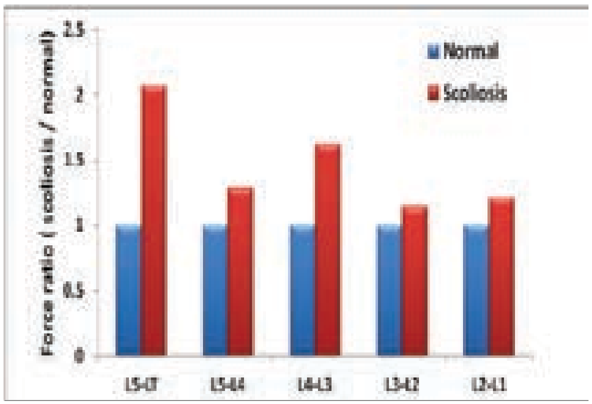

Fig. 13 The variation of lumber joint force in scoliosis model compare to the normal model

Sagittal, lateral, and twisting torques were also compared in the same way among the models. The results show that torques also increase in all lumber joints of scoliosis model compared to the normal one in all three phases, left bending, standing and right bending. Because of the symmetry of the spine in the normal model in standing position, sagittal, lateral and twisting torques are near zero. However, in scoliosis model, due to the asymmetric mass distribution, these torques are considerable.
Figure 14 shows the joint torque ratio of scoliosis model respect to that of the normal one at lumbo-sacral joint in three planes. Since the Cobb angle in thoracic region is in the right side of the subject the joint torques in left bending should be increased comparing to the normal model because of the mass distribution that exists in the right side.

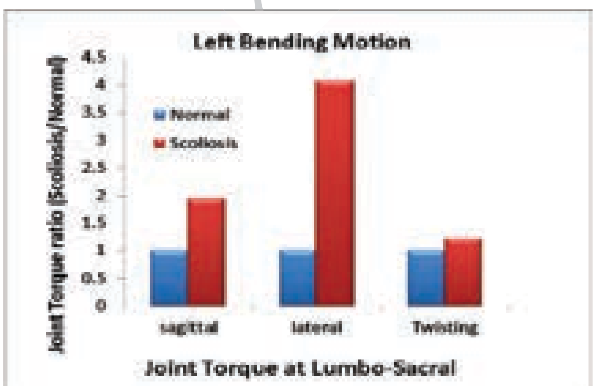

Fig. 14 Joint torques force at lumbo-sacral joints

As can also be seen, the maximum variation occurs in the lateral plane. This is because of the curvature that scoliosis model has is in right side of the body and the vertebrae that are not in the center line can produce extra torques on vertebral discs. In addition, the motion is also in lateral plane.

4 CONCLUSION

We have built a fully discretized bio-fidelity simulation model for analysis of scoliosis conditions in various patients. This paper has presented an overview of our ongoing work towards building multi-body musculo-skeletal scoliotic spine models for investigating various medical applications.
These models can be useful for incorporation into design tools for wheelchairs or other seating systems which may require attention to ergonomics as well as assessing biomechanical behavior between natural spines and those that have undergone spinal arthroplasty or spinal arthrodesis.
Furthermore, these models can be combined with analysis tools to help surgeons to examine kinematic behaviors as well as force distributions around scoliotic spines and to propose possible surgical plans before spine correction operations. Trajectories and forces can be computed for various postures and the models can be used to assist surgeons in pre-operative planning and post-operative treatment.
In summary, this research has investigated the effect of scoliosis condition on spinal angles, muscle activation and joints force by the application of motion data capturing and virtual musculo-skeletal modeling in lateral bending motion. This work is ongoing and aims to contribute to the understanding of complex scoliotic spine biomechanics. So far this particular research study has only been carried out on a single scoliosis subject. More scoliosis subjects will be involved in the future to study the spinal biomechanics in different motion on different scoliosis types.
The result of this study showed that the loads at the lumber joints in scoliosis model are considerably higher than the loads of normal model in lateral bending motion. This may lead to a better understanding of corresponding pain during some activity in scoliosis subjects. It is hoped that this result will help future investigations on scoliosis to understand furthermore its development as well as improved treatment processes.

5 ACKNOWLEDGMENTS

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