IMPROVING SCOLIOSIS REHABILITATION

by

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Abstract

IMPROVING SCOLIOSIS REHABILITATION

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Scoliosis, a disease that mainly affects children, is a deformity of the spinal column in all three dimensions. Treatment for scoliosis aims at straightening the spine with minimal discomfort to the patient. Current treatments involve wearing a polymeric brace that based on its design, exerts force on the spine in order to straighten it. The shape of the brace and the locations where the brace should apply forces are defined based on the experience of the orthotics specialist. Currently used braces are not equipped with sensors to monitor the loads generated and possibly aid in the rehabilitation process. The rehabilitation procedure could be improved if the brace design process is based on more fundamental analysis and the forces exerted by the brace could be monitored during rehabilitation. In this research, simplified two dimensional and three dimensional finite element models of the spinal column including the ribs and sternum were developed and used to qualitatively study the effects of brace forces on the spinal deformation using finite element analysis tools. These models were used to understand the spinal deformation as a function of the magnitude of force and application location. In addition, an actual brace was successfully outfitted with sensors (strain gauges and pressure) and interfaced with a data acquisition system to acquire the strain on the brace under loading. The results of this research effort will be employed to improve scoliosis rehabilitation and to gain a better understanding of the rehabilitation process.

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Chapter 1

Introduction to Scoliosis

Scoliosis is a disease that generally affects younger individuals with no warning and no reasoning; such as genetics from previous generations. Idiopathic scoliosis is the deformation of the spine in three dimensions and accounts for approximately 70% of all scoliosis cases. [1] Most people associate scoliosis with the bending of the spine in and out of the sagittal plane; however, in some cases it twists from the lumbar vertebrae up to the cervical vertebrae as well as deforms in the anterior or posterior directions. The severity of scoliosis is determined by the magnitude of the Cobb angle. The Cobb angle is found by extending lines tangent to the vertebrae and determining where the intersection is located. The angle at which the two lines intersect is the Cobb angle. [2]

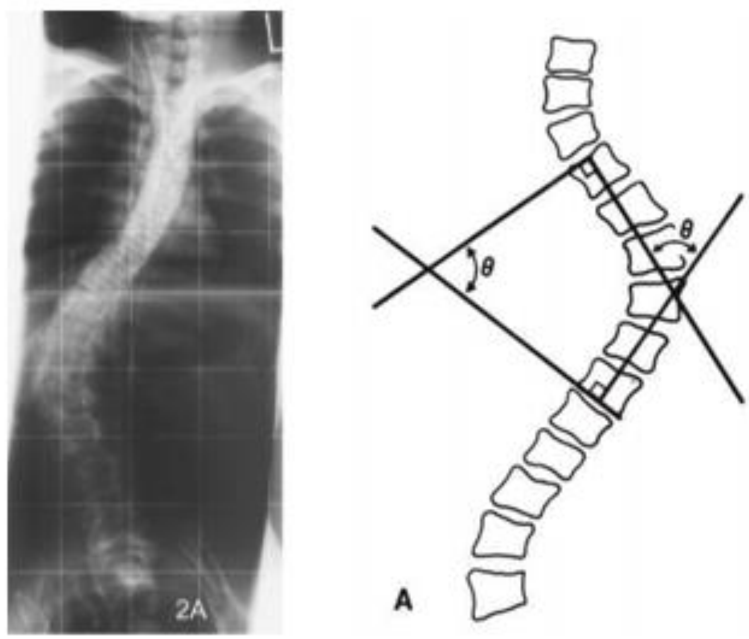

Figure 1.1: (Left) X-ray results of a scoliosis patient (right) Cobb angle deformation diagram [2,3]

This angle is found by taking x-rays of the patient throughout treatment and then determining how much the rehabilitation is helping. A patient is deemed to require bracing if the Cobb angle is more than 40 degrees; consequently, angles smaller than 40 degrees are generally left alone or treated with a night time brace. [1,4,5] On the opposite extreme, surgery is common when the Cobb angle reaches angles in excess of 70 degrees. Due to scoliosis being found predominately in children, schools across the United States perform routine exams to determine which children are at risk. The test involves the child bending at the waist while the nurse checks the spine for any signs of curvature as well as checking if the shoulders are at the same height. This procedure exposes the shape of the spine and shows a possible height difference in the child's shoulders. While the method is effective, some schools test children with x-rays to determine if the child has scoliosis. Despite the more profound accuracy of using x-rays, it is not being used as often and is currently under scrutiny for subjecting children to harmful radiation with a small portion of the population having scoliosis. [6]

1.1 Scoliosis Treatment

There are three main ways of treating the different severities of scoliosis: letting the patient grow out of it, bracing, and surgery. Cobb angles less than 40 degrees typically allow doctors and specialists to do nothing and hope that time and growth spurts will take care of any small curvature. Alternatively, they could treat it by providing the patient with a personalized brace to wear only at night. Since patients are generally younger when scoliosis is detected there is the possibility, if the Cobb angle is small enough, that when the patient grows, the spine will realign itself. The use of the night-time brace is prominent when the scoliosis curve is on the threshold of bracing or not. Doctors believe that using the night brace will help realign the spinal column without causing the patient more discomfort than necessary. The night brace also helps the children physiologically by the child not having to wear the brace during the day while at school where other students will be aware of their condition and possibly make fun of them. The corrective abilities of bracing rehabilitation are shown in Figure 1.2.

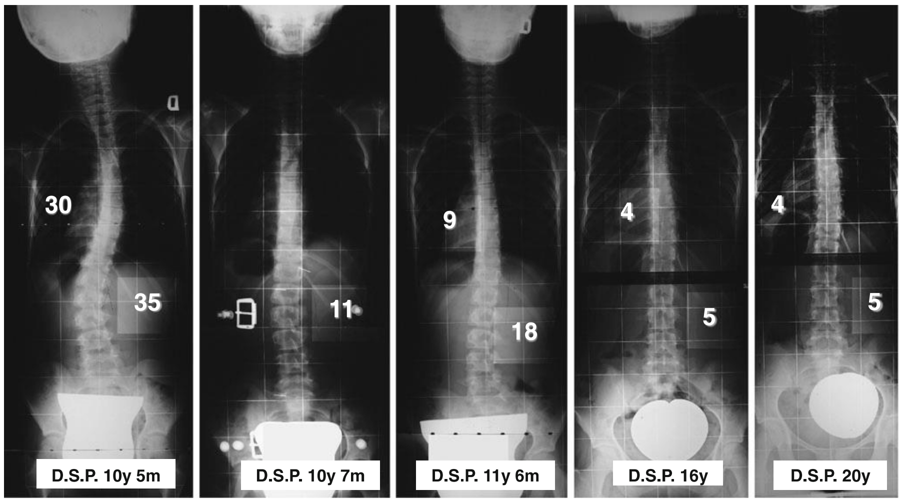

Figure 1.2: Timeline of x-ray results from scoliosis rehabilitation [7]

In the most extreme circumstances, surgery is the last treatment of scoliosis and is often the last resort for doctors and specialists. In most cases, Cobb angles that are over 70 degrees will not be corrected by a brace. To correctly address the increased Cobb angle, surgically implanted plates and screws are normally attached to the spine. Surgery is used to correct the curvature the best it can, but in most cases, only a minor amount gets corrected. The main purpose of the surgery is not to correct, but to ensure the curvature does not advance. During surgery, intervertebral discs are removed from the spinal column and bone graft material is used to replace the discs before they are screwed down. [8] The biggest concern when performing corrective spinal surgery on these patients is the possibility of leaving the child disabled if the curvature cannot be corrected without serious complications. Though surgery is a last resort for doctors, it does stop the curvature from progressing and getting worse. With extreme cases of scoliosis, surgery is required to save the patient's life. Since the spine is bending and twisting throughout the body, organs can grow in abnormal shapes, be crushed, or punctured by the ribs or vertebrae.

Though scoliosis does not stunt the overall growth of the child, having extreme curvature can cause problems on the inside of the body. [9]

The middle portion of scoliosis treatment, where Cobb angles range from 40 to 70 degrees, is treated with the use of some type of brace. The three most commonly used types are the Charleston brace, Boston brace, and Cheneau brace which are shown in Figure 1.3.

Figure 1.3: Left to right; Charleston brace, Boston brace, Cheneau brace [10,11]

1.1.1 Charleston Brace

The Charleston brace is used in mild cases of scoliosis where the Cobb angles range from 20 to 40 degrees. As stated, when the doctor believes that the patient has a mild form of scoliosis, but normal growth by the patient will not be enough to combat the problem completely, the patient must wear a brace at night. The Charleston brace bends the patient's spine in the opposite direction of the curvature. The brace starts at the buttocks of the patient, hugging the hips for structural support, and then continues curving up to underneath the patient's arms. [10]

1.1.2 Boston Brace

The Boston brace is the most common brace utilized for rehabilitation with scoliosis patients. This brace wraps around the lower portion of the back, resting on the hips, and stops just below the armpit of the patient. There is support material attached to the inside of the brace that will push on the spine and ribs to correct the curvature. The support material is primarily used to push on the spine to align it, but is made of aliplast foam to make wearing the brace more comfortable for the patient. [10] A Boston brace that was previously used at Texas Scottish Rite Hospital for Children (TSRHC) was donated to this research effort for experimentation by outfitting it with sensors and collecting appropriate data.

1.1.3 Cheneau Brace

The Cheneau brace is more commonly used in European countries compared to the popularity of the Boston brace in the United States. Though it is similar to the Boston brace, it is more personalized for treatment of the patient. This brace contains regions of convex and concave curves to create additional pressure on the spine. The brace allows for the rotation of the pelvis and shoulders to account for the twist of the spine, while still maintaining pressure to push the spinal curvature and align properly over the sacrum. [11]

1.2 Brace Treatment

At TSRHC, when a new scoliosis patient comes in the hospital, the first thing that the doctors and specialists do is take x-rays of the patient’s spine to determine the severity of the curve. That information is then used to create a custom rehabilitation regiment for the patient that would range from bracing to surgery or just waiting to see if growing will realign the spine. If bracing is required, a personalized brace is fabricated based on the patient’s body and fitted with aliplast foam padding to create support locations; therefore, when the brace is tightened it will push against the spinal curvature to correct it. Once the brace is fitted, the doctor and/or specialist will teach the child and his or her parental guardians how to put the brace on every day and what guidelines they need to follow to ensure that the brace does its job. The only way of determining if the brace is tight enough is to make sure that the doctor or specialist's fingers cannot fit anywhere between the brace and the patient's body. Unfortunately there is no finite value of force or pressure that is used to determine if the brace is tight enough. The patient generally has to wear the brace 22-23 hours a day until he or she naturally stops growing. Therefore, it is imperative that the child keeps the brace tight throughout the day due to the straps loosening during daily activities in order to maximize rehabilitation. The child needs to ensure that if they take the brace off, when they put it back on that the brace is retightened to the designated tightness. Currently, doctors or specialists will mark on the vest straps to show the level of tightness they should be experiencing every week. The doctors and specialists' experience shows that because of how often the patient is wearing the brace the markings quickly fade leaving them with no reference for retightening the straps. Therefore, if the brace could be outfitted with sensors, a finite number from a sensor reading could be determined for the patient as a tightening reference. This could eliminate the need for marking the straps with estimations of where the straps should be pulled to and start the process of determining what forces are exerted on the body from wearing these sensor outfitted braces.

1.3 Future Treatment

Scoliosis treatment is only monitored when the patient returns to the hospital to have an x-ray taken. This research aims to determine a sufficient and reliable way to monitor strains in real time as the means to improve the rehabilitation and treatment. Currently, a scan of the patient's spine is input into modeling software to analyze the various locations of where support needs to be installed. The primary objective of this research is to attach sensors on the brace to record stress and pressure values that the brace is experiencing. These sensors will record information at regular time intervals ensuring that the spine is experiencing the necessary forces required to correct the curvature. The second objective is to sync the sensors with actuators that will control the straps that tighten and loosen the brace automatically. For most braces, two or three straps are required to close and secure the back of the brace. Actuators could be used to control each strap that will increase or decrease the pressure required. Having the straps at different tensions on the back of the brace will allow individual changes depending on the data collected from the sensors. With all of these components working together, the knowledge that could be gained about scoliosis is limitless. Since the actual force experienced by the body is something that is relatively unknown to the doctors and specialists, this information could possibly be employed to develop new treatments for scoliosis. Considering the technology that is available today, sensor outfitted braces make it easier for the hospital to track progress and potentially lower the time required for the patient to be in the brace. New customized braces could be made depending on the severity of the curvature for each patient, with emphasis on monitoring critical regions. Combining these concepts together could potentially save the patient and family time and money from going to the hospital for checkups as well as help the hospital treat more patients.

1.4 Outline of Thesis Document

In this document, finite element models will be developed in chapter 2 and analyzed in order to gain a better understanding of how force magnitude and application needs to be applied to simulate scoliosis. The finite element models include a 2D model, a 3D model with ribs and a sternum, and a 3D model with a brace. A set of four loads and different force application locations were analyzed. In chapter 3, a donated Boston brace is outfitted with sensors (strain gauges) and a data acquisition system to demonstrate real time data acquisition when the brace is under different loading scenarios. Chapter 4 summarizes the information collected from these two different sections and proposes future work to advance the rehabilitation of scoliosis.

Chapter 2

Spinal Modeling and Finite Element Analysis

2.1 Spine Modeling

A solid model of the spine was developed similar to the spine models found in [4, 5, 12, 15, 16, 17, and 18] to better understand the various components of the spine and how it responds to controllably applied forces through finite element analysis software. In general finite element software does not have extensive three dimensional modeling capabilities for creating complex models for analysis. A three dimensional model of the spinal column was drafted in Creo Parametric 2.0 (PTC Inc., Needham, MA) to later be implemented into a finite element analysis software. The available computer system used in the mechanical engineering department CAD lab was not able to perform finite element analysis of the 3D spine; therefore, a 2D model was developed from the original 3D model. The spine models developed detail the Cervical vertebrae starting with the third vertebrae down to the last Lumbar vertebra based on the human spinal column as shown in Figure 2.1.

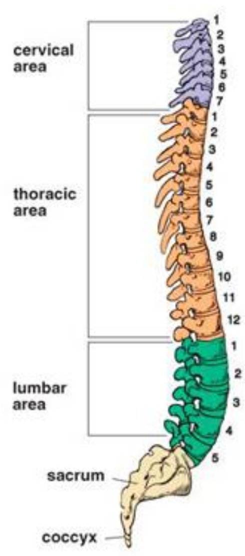

Figure 2.1: Vertebrae regions in the spinal column

The first two vertebrae were skipped due to the fact that they are fused to the skull and do not have or experience significant movement in any of the Cartesian directions centered at the bottom of the fifth lumbar vertebra. The thirteen thoracic and five lumbar vertebrae were modeled and assembled with the cervical vertebrae but the sacral and coccygeal vertebrae were excluded due to the vertebrae being fused together in conjunction with the pelvis. The model of the spine is simplified by modeling the vertebrae as cylinders with the articulated faucet joints being excluded. Along with the vertebrae, bone marrow found inside each of the vertebrae that could improve structural integrity was also modeled. The dimensions of all the vertebrae were determined by taking the average values of the dimensions from cadavers that ranged in age from 55 to 84 years old and are detailed in Table 2.1. [20] Due to the difficulty of finding dimensions of vertebrae and disks in children, the values found for adults were used in the simulations.

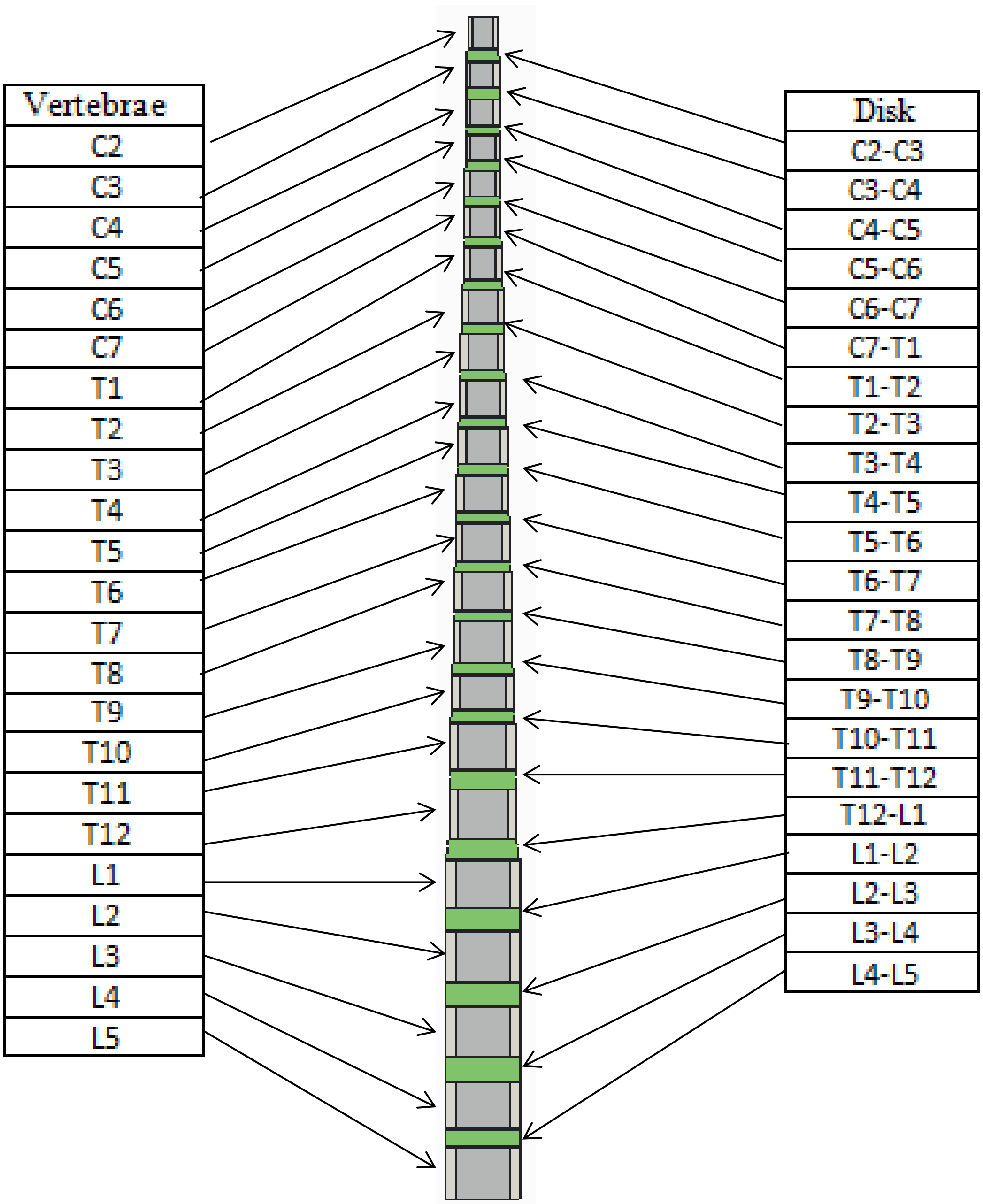

Figure 2.2: 2D spine model with the components of the spinal column

Table 2.1: Vertebrae heights and widths used in 2D and 3D models [20, 21]

Vertebrae	Height (mm)	Diameter (mm)
C2	16.00	14.00
C3	13.00	15.00
C4	13.00	15.00
C5	13.00	15.75
C6	13.00	16.25
C7	15.00	16.35
T1	16.50	17.55
T2	17.30	18.80
T3	18.40	20.05
T4	18.90	21.30
T5	18.35	22.55
T6	18.60	23.80
T7	19.55	25.05
T8	19.95	26.30
T9	21.35	27.55
T10	17.75	28.80
T11	24.20	30.05
T12	25.55	31.3
L1	25.00	34.00
L2	26.00	34.00
L3	26.00	34.00
L4	24.00	34.00
L5	27.00	34.00

The disks were modeled as trapezoidal cylinders to ensure the top and bottom of each disk make a smooth transition between the vertebrae. The top and bottom surfaces of the disks were defined to match that of the vertebrae that it would be attaching to; though, the disk heights did vary from disk to disk with the dimensions shown in Table 2.2. In addition to the disks, each end plate that attaches the vertebrae to the disk was modeled as being of the same diameter as the vertebrae and a thickness of 0.6 mm. [16]

Table 2.2: Disk heights used in the 2D and 3D models [20, 21]

Disk	Disk Height (mm)	Superior Width (mm)	Inferior Width (mm)
C2-C3	6.10	14.00	15.00
C3-C4	5.80	15.00	15.00
C4-C5	4.50	15.00	15.75
C5-C6	4.40	15.75	16.25
C6-C7	4.70	16.25	16.35
C7-T1	5.00	16.35	17.55
T1-T2	4.50	17.55	18.80
T2-T3	4.70	18.80	20.05
T3-T4	4.70	20.05	21.30
T4-T5	4.80	21.30	22.55
T5-T6	5.30	22.55	23.80
T6-T7	5.20	23.80	25.05
T7-T8	4.90	25.05	26.30
T8-T9	4.80	26.30	27.55
T9-T10	5.90	27.55	28.80
T10-T11	6.10	28.80	30.05
T11-T12	9.30	30.05	31.30
T12-L1	10.30	31.30	34.00
L1-L2	11.50	34.00	34.00
L2-L3	11.80	34.00	34.00
L3-L4	12.70	34.00	34.00
L4-L5	8.80	34.00	34.00

This simplified model depicts the spine as a single column of cylinders instead of the general curve in and out of the coronal plane. The simplified spine model provides for faster simulations without compromising the overall understanding of the stress and deformation experienced by the spine under loading. The model of the spine was developed in Creo 2.0 Parametric and then imported and analyzed in the finite element software package ANSYS (ANSYS Inc., Canonsburg, PA). The material properties used in the model of the spine in ANSYS are shown in Table 2.3. The values used for the finite element analysis were found by comparing literature references that ran simulations to determine the most accurate material properties or experimentally tested elements that were retrieved from cadavers. As expected, the values for the material properties vary drastically from source to source due to the high variability from person to person.

Table 2.3: Material properties used for the 2D and 3D ANSYS analysis

Element	Young's Modulus, MPa	Poisson’s Coefficient	Reference
Vertebra	12000	0.3	16
Marrow	500	0.3	16
End Plates	100	0.4	16
Disks	500	0.3	16
Ribs	5000	0.1	15
Sternum	10000	0.2	15
Rib Cart	480	0.1	15

2.2 Two Dimensional Modeling

2.2.1 ANSYS Model

The 2D model of the spine was generated by taking a center cutout of 0.5mm thick from the 3D model. The 2D model was imported into ANSYS Workbench 15.0 to be analyzed under static loading conditions. The finite element model was simplified by defining the constraints for each of the vertebrae, disks, and end plates as bonded. This condition was implemented since the spinal column is held together with numerous nerves, such as the spinal cord that runs between the vertebrae and the articulate faucet joints, as well as the numerous ligaments, tendons and muscles that wrap around the spinal column. Though the 2D model could have been analyzed as a 2D surface with an applied thickness it was treated as a three dimensional structure with a 0.5 mm thickness. Therefore, instead of analyzing it in ANSYS as a surface with surface elements, solid elements were applied to the structure. The mesh was defined with a solid brick 20-node element with mid-side nodes defined in ANSYS as a Solid186 element. Using this 20-node element, it is possible to accommodate up to quadratic displacements in the model and large deformation. The 2D model mesh consists of 28392 nodes and 2756 elements as shown in Figure 2.1.

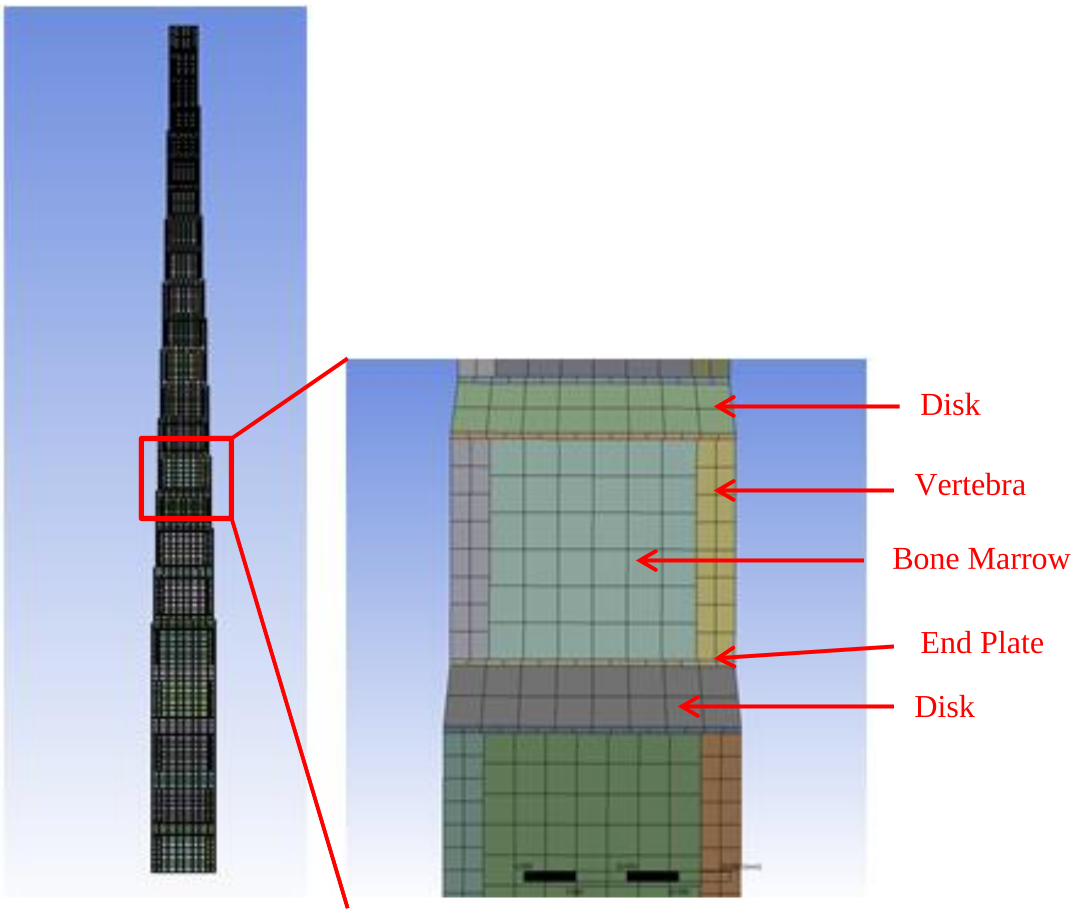

Figure 2.3: Fine mesh used in the 2D spine model

The bottom of the L5 vertebra has been set to a rigidly fixed displacement boundary condition that prevents the bottom of the spine from translating in either the x-axis or y-axis or rotating about the z-axis thus preventing rigid body motion or translation. Fixing the L5 vertebra in this manner, illustrates the fact that the vertebra and the sacral and coccygeal vertebrae are rigidly attached to the pelvis. When the spine is pushed in a lateral manner, the L5 vertebra should be fixed because any change in that direction will be due to the cervical vertebrae region translating up or down. The C3 vertebra at the top of the spine was set to a displacement boundary condition to demonstrate this scenario. This boundary condition fixes the top of the spine from translating in any direction but the y-axis (the axis that runs down the length of the spine). To analyze how the spine responds to lateral loading conditions, no additional fixed displacement boundary conditions were applied. All of the boundary conditions that were applied to the 2D model are presented in Figure 2.2.

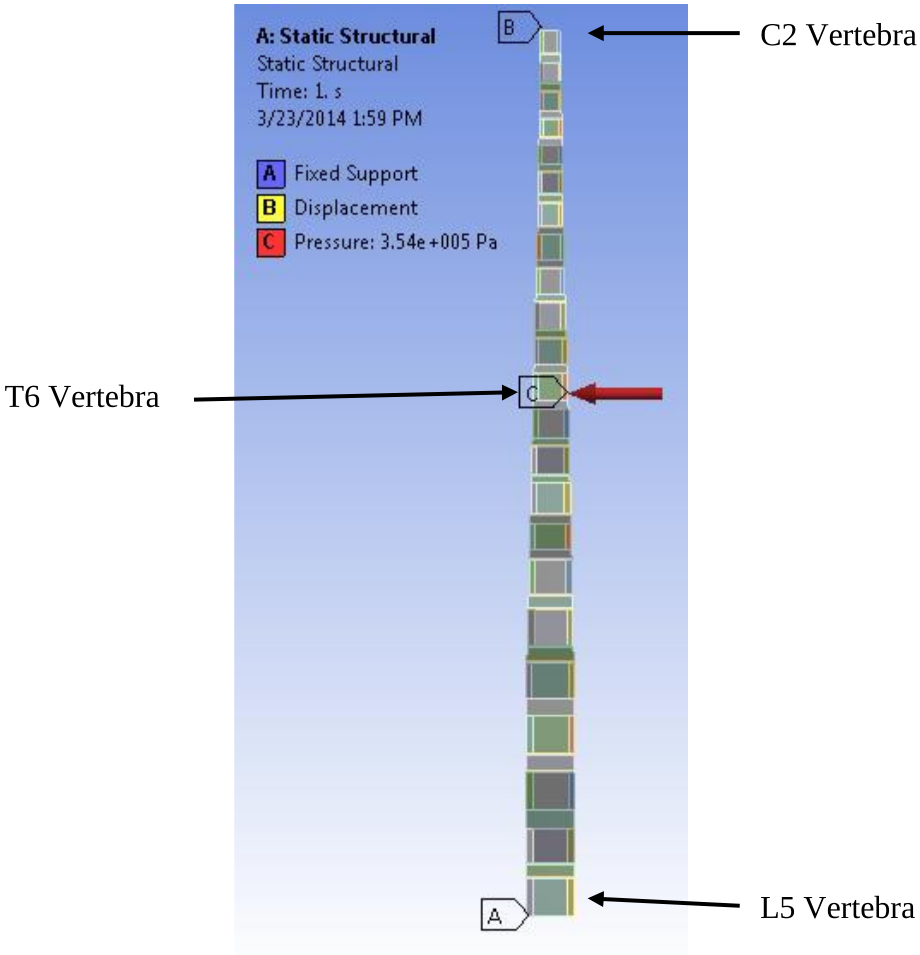

Figure 2.4: 2D spine model with load and boundary conditions

2.2.2 Simulating Loads on the Spine

In order to gain an understanding of how the spine responds to forces exerted in the lateral direction, various loads were applied to the T5 vertebra down to the T7 vertebra. This includes all of the disks and endplates between these vertebrae. This region was modeled to simulate a thoracic scoliosis curve that affects the vertebrae in the middle of the thoracic region and to gain an understanding on how the spinal column responds to applied forces. The loads applied to the spine were applied as pressures in ANSYS. The overall area where the loading was taking place was calculated and the equivalent pressure was applied to that specific region. The equivalent von Mises stress and the total deformation of the spine were obtained for each ANSYS simulation. The padding on the inside of the brace has been determined to experience a force between 20 to 70 Newtons. [16] In order to gain an understanding of how the spine reacts to these different forces, a 20 N and 70 N force were used based on the information found in the literature. A 50 N force was also added as a close midway point to capture the middle of the range, and for extreme measures a 100 N force was applied to see how the spine reacted to a load outside of the experienced range. The maximum stress and deformation in the spine for each loading condition is shown in Table 2.4. The locations of the maximum stress and deformations results were at the T6 vertebra which is the middle vertebra of where the forces were being applied.

Table 2.4: Stress and deformation results from the 2D testing

Force (N)	Maximum von Mises Stress (MPa)	Maximum Deformation (mm)
20	48.8	24.7
50	122.1	61.9
70	171.1	86.7
100	244.2	123.7

2.2.3 Discussion of the Results

When analyzing the results, the first concern is why the spine deformed 123 mm (4.8 in) and experiences such a high stress value. These values are not physically possible or survivable. The relationship between the force applied and the stress or deformation does give an idea of what the spine is experiencing. The deformed shape and stress distribution of the spine for the extreme loading conditions of 20 N and 100 N are shown in Figures 2.5 and 2.6.

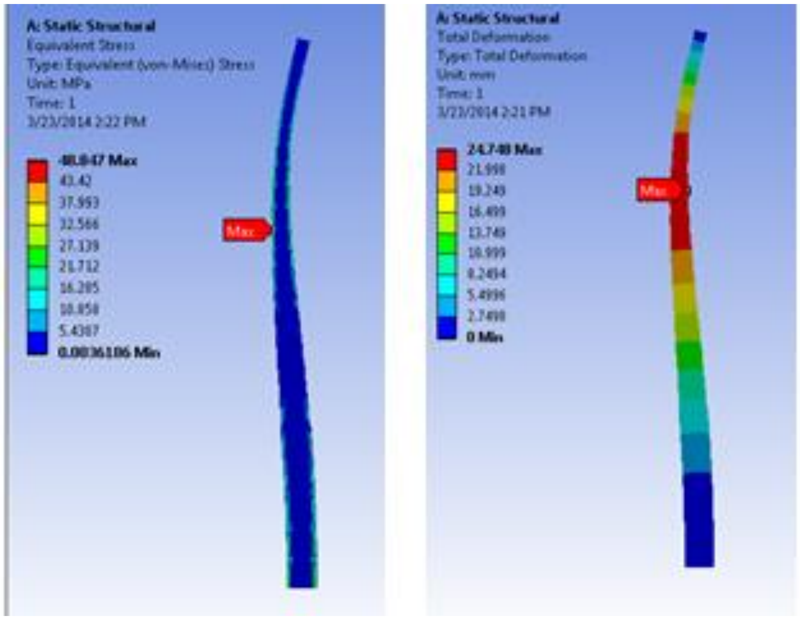

Figure 2.5: Stress and deformation distributions of the 2D spine under a 20 N loading

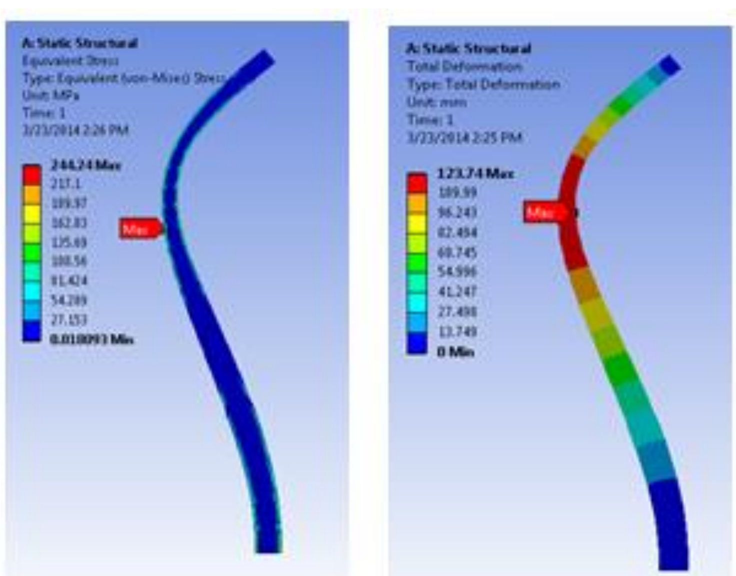

Figure 2.6: Stress and deformation distributions of the 2D spine under a 100 N loading

The results of the 2D model show a thoracic curve that would be similar to curvature found in a patient with thoracic scoliosis. However, even though the shape of the curve is similar to that of an actual patient, the results are too extreme. The 2D model does allow for an understanding of how the spine responds to these applied forces, and a Cobb angle of 66.4 degrees was generated from the results of applying 100 N. In order to gain a better understanding of how these forces affect the spine a 3D model that includes the sternum and ribs to transfer the load to the spine needs to be developed.

2.2.4 Validation of Results

Validation of the spine subject to these various lateral applied loads can be performed using beam equations to find the deflection. The spine can be analyzed as a simple beam that has a constant cross sectional diameter of 24 mm, which is approximately the average diameter of all the vertebrae. For the simplest analysis, the beam can be simply supported with a load applied to the center and using the vertebrae material properties shown in Table 2.3. Equation 2.1 is used to solve the deflection in the beam with E as the material's Young Modulus, L the length of the beam, P is the applied force, and I is the cross sectional moment of inertia.

$$v = \frac{-PL^3}{48EI} \qquad 2.1 \ [24]$$

Analyzing the beam under these boundary conditions show deflection results extremely similar to the deformation found in Figures 2.5 and 2.6. Though the boundary conditions for the 2D spine model are not simply supported but fixed, making the problem more of a statically indeterminate problem. An equation for the spine has been generated to show how the spine will deform as reported in [1].

2.3 Three Dimensional Modeling

2.3.1 ANSYS Model

The 3D model of the spine is similar to the 2D model with new complexities, including ribs that would realistically show how the spine would react to pressure from the brace. In the 2D spine model the pressure is applied directly to the side of the spine where in reality the brace would be applying pressure to the ribs that would then transfer the load to the spine. The 3D spine model includes a simplified model of the ligaments, tendons, and cartilage that would be holding the ribs to the spine and were modeled as single elements as shown in Figure 2.7.

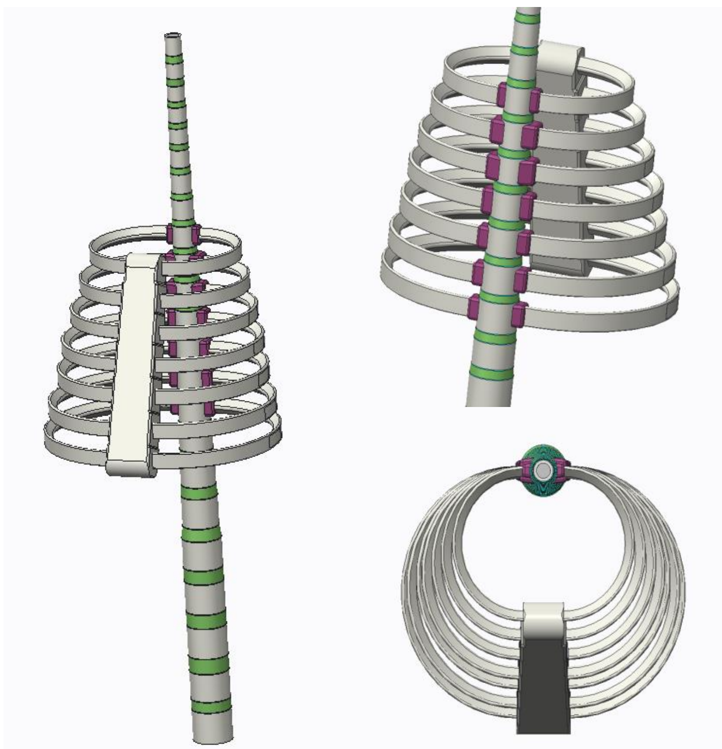

Figure 2.7: Isometric, top, and back view of the 3D spine model

The 3D model includes a simplified model of the sternum to complete the rib cage as well as transfer loads from one side to the other. The Solid186, 20-node, solid element was used in the meshing of the 3D spine model. These elements consisted of not only the brick elements but also tetrahedral elements with mid-side nodes. Though the three dimensional model is not expected to have the severe deformation of the 2D model, it is expected to have quadratic deformation that the Solid186 elements will help capture. The coarse mesh was used due to the complexity of the model and to ensure that the available computer systems in the MAE CAD lab would be able to analyze it. There are 138130 nodes and 31320 elements in this model of 149 different parts and is shown in Figure 2.9.

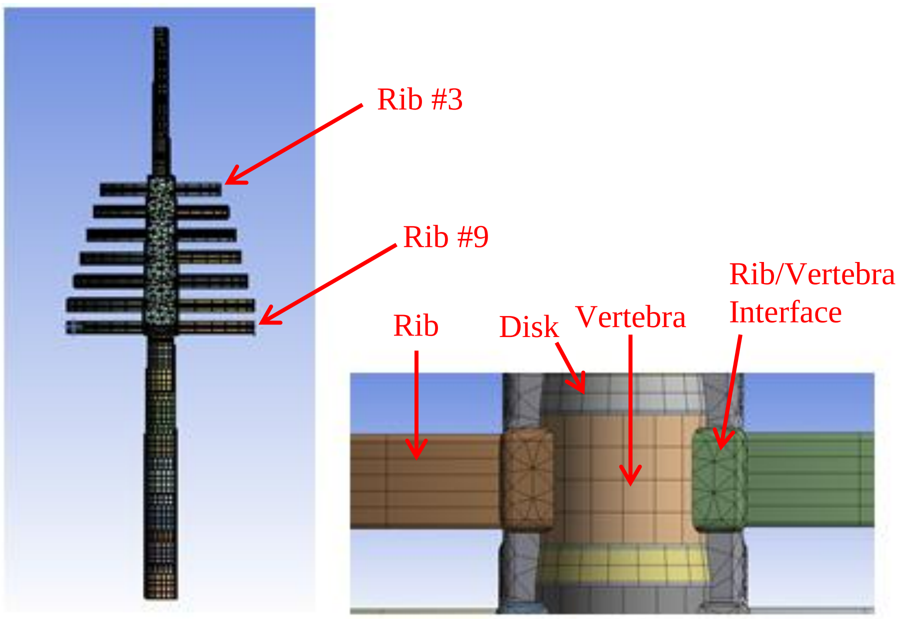

Figure 2.8: 3D model mesh distribution with focus on T6 vertebra

As described in the 2D modeling of the spine, the L5 vertebra is once again under a fixed displacement boundary condition. The bottom of the spine will not translate in either the x-axis or the y-axis, and it will not be allowed to rotate about the z-axis. This is done to model how the bottom portion of the spine is joined to the pelvis and prevent any motion. The C3 vertebra is held under a displacement boundary condition, allowing the spine to contract in the y-direction as is the case when the lateral force is applied to the side of the model. Pressures were exerted on the ribs instead of the vertebrae and now fixed displacement boundary conditions were added to oppose the pressure to simulate the brace. In order to compare the results from the 2D analysis to the 3D analysis, ribs five, six, and seven have varying pressures applied to the left side of the body. However, to simulate what the brace will be applying to the rib cage, ribs three, four, eight, and nine will be held to the same fixed displacement that the L5 vertebra is experiencing. This scenario models how the brace would be acting on the spine by padding pressing on ribs five, six, and seven and the area around ribs three, four, eight, and nine would be fixed as shown in Figure 2.10. Opposite of ribs five, six, and seven there are no boundary conditions as would be the case for simulating a brace. In the brace design, the region opposite of the built up region is left open due to the fact that the body needs a place to go if pressure is being exerted on the opposite side.

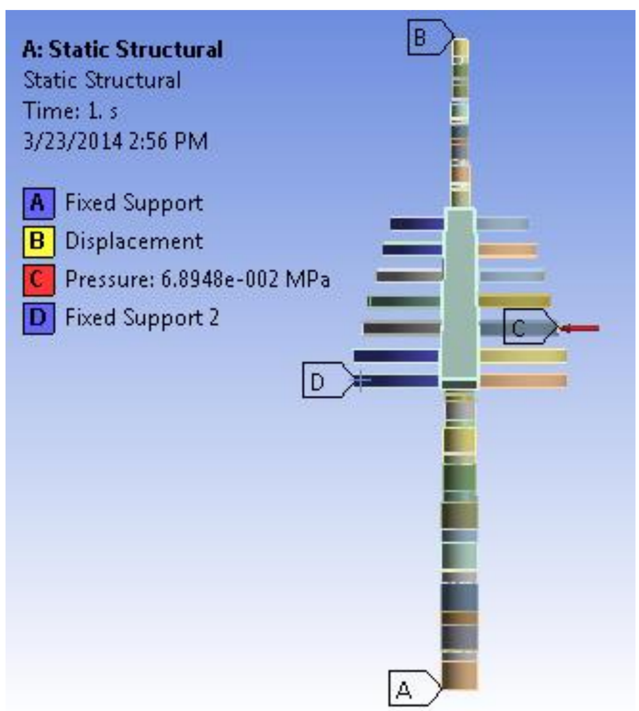

Figure 2.9: Boundary conditions for the 3D spine model

2.3.2 Simulating Loads on the Ribs

The test loads of 20, 50, 70, and 100 N were applied as pressures on ribs five, six, and seven similar to the 2D spine model. The equivalent von Mises stress and the total deformation are the two quantities desired from this analysis. While these results will show major deformation and stress located in the ribs, the main focus is on how the spine reacts to the applied pressures. In the body, the ribs will have several muscles, ligaments, tendons, and other tissue keeping them from deflecting significantly. Also the ribs are against organs like the lungs, heart, stomach, and liver that will also be there to resist the compaction of the rib cage. Table 2.5 illustrates the results from the 3D analysis under the various applied forces.

Table 2.5: Maximum stress and deformations on the spine from the 3D model

Force (N)	Maximum von Mises Stress (MPa)	Maximum Deformation (mm)
20	0.17	0.02
50	0.46	0.04
70	0.57	0.06
100	0.83	0.09

2.3.3 Explanation of the Results

The von Mises stress and deformation results from the 3D analysis under the same forces are clearly reduced from the values in the 2D analysis. The locations of the maximum stress and deformations are in the same location on the T6 vertebra. These results clearly show the importance of modeling the spine with the ribs and sternum if transverse loadings are to be applied. Just as shown in the 2D model, there is a linear correlation between the applied force and the output stress and deformation.

The 3D model provides a clearer understanding of what the spine experiences under these forces as compared to the 2D model. Von Mises stress and deformation of the spine for the two extreme loading conditions of 20 N and 100 N are shown in Figures 2.10 and 2.11. The deformation scaling on these ANSYS results is under a true scale therefore deformations will not show up as clearly as in the 2D modeling results; however, the stress and deformation distributions increase with amplified force as expected.

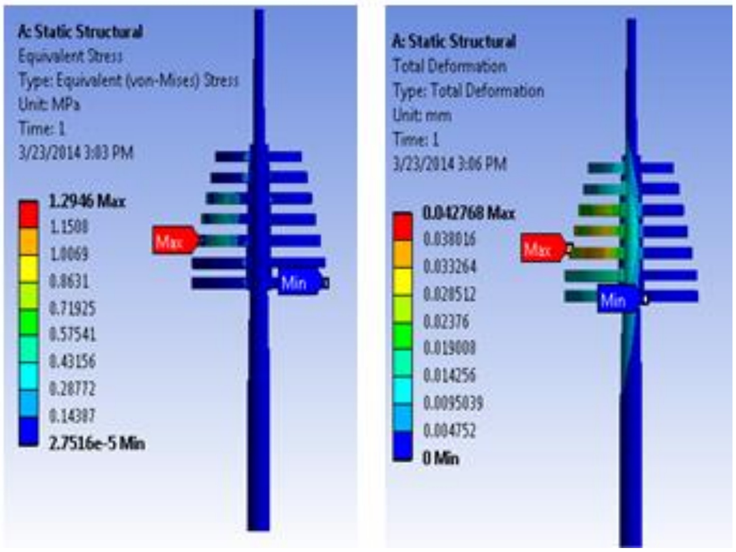

Figure 2.10: Stress and deformation distributions of the 3D spine under a 20 N loading

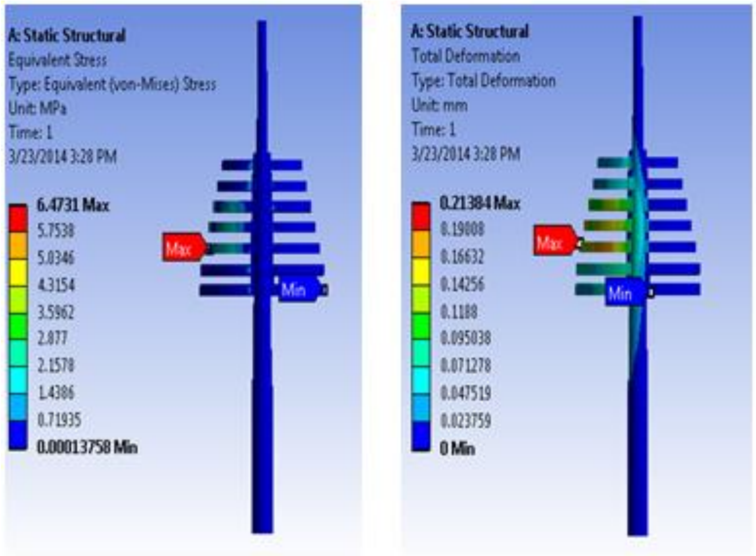

Figure 2.11: Stress and deformation distributions of the 3D spine under a 100 N loading

The spinal deformation ranges from the top of the first lumbar vertebra to the third thoracic vertebra. This indicates that with the pressure applied to three ribs, the deformation of the spine covers a much larger region because the force being distributed throughout the body.

This model could be employed to determine an initial guidance on how to aid in the rehabilitation of scoliosis. A Cobb angle was drawn and evaluated from the results of the 100 N applied force and found to be approximately 4.7 degrees. Since this model was tested under a static condition only the stress and deformation results will be able to assist in the understanding of this disease. However, if this model was tested under dynamic loading conditions a more distinct deformation in the spine would appear and a more accurate Cobb angle could be calculated from the body enduring a load for a longer duration. These stress and deformation results would provide doctors and specialists a better understanding of what the body is experiencing when the child is wearing the brace and can adjust rehabilitation procedures as needed if there is any concern of damage to other regions of the body.

2.3.4 Validation of Results

The spine and rib cage analyzed with these different applied loads gives a better understanding of how the spinal column responds when under pressure. The rib cage was initially loaded with symmetric loads to ensure that there would be a symmetric stress and deformation distribution throughout the rest of the body. After applying the 20 N load to both the left and right sixth rib, the stress and deformation distributions were symmetric down the center of the rib cage and throughout the spine. This in combination with the validation from the 2D spinal modeling, shows that the model is responding to these various applied loads as expected. This model of the spine includes the rib cage which the 2D model validation did not have; however, it did deform as expected and as shown in the validation that was used with the 2D model.

2.4 Three Dimensional Brace Modeling

2.4.1 ANSYS Model

The goal of the 2D model used in the beginning, was to determine how the spine responded to having pressures exerted on the vertebrae. Then, to gain a better understanding of how the spinal column reacted to these applied forces the rib cage was added and the analysis was performed in three dimensions. In order to model the effects of the brace more representative for a child with scoliosis, a brace was modeled to fit over the rib cage. This brace demonstrates how the applied force is distributed from one or two ribs throughout the rest of the body and into the spinal column. In this model, the same 3D rib cage and spine model was used with the only modification being the brace surrounding the rib cage. An extruded surface was added connecting each rib together to create a solid surface covering the rib cage as shown in Figure 2.12.

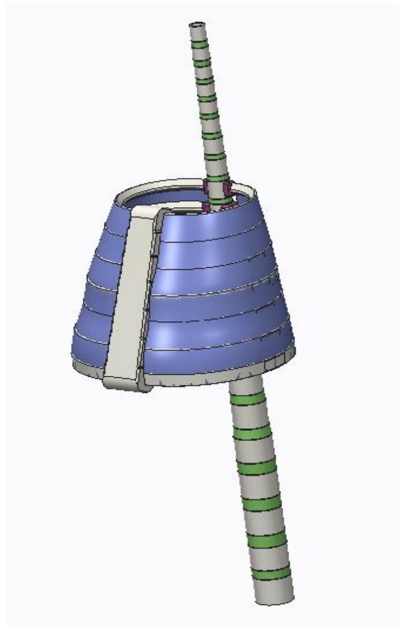

Figure 2.12: Isometric view of the 3D model with the brace

Though the sternum extrudes out of the brace, which would not be a realistic case, this still provides an accurate model of what would happen with the brace-ribs-spine pressure interaction. The surface was then given a thickness of 5 mm, the same thickness of the brace that was donated by the TSRHC. The material properties of the brace were a Young's Modulus of 1500 MPa and a Poisson's coefficient of 0.3. [5]

This model only differs from the original 3D model with the addition of the brace. Therefore, the same Solid186 elements were used in this analysis as well. Based on the results from the 3D model, without the modeled brace there will not be any major deformations in the curvature of the spine due to the lateral applied forces. However, there will be some quadratic deformation that having these elements will identify. The same combination of solid brick elements and tetrahedral elements are generated to complete the mesh for the analysis with the brace as shown in Figure 2.13. There are 138335 nodes and 30778 elements in this model with an average run time for this simulation of approximately 15 minutes on a Windows 7 HP Compaq

Elite 8300 CMT. This PC was equipped with an Intel i7-3770 CPU at 3.40 GHz, 8 GB of RAM, and a 64-bit operating system.

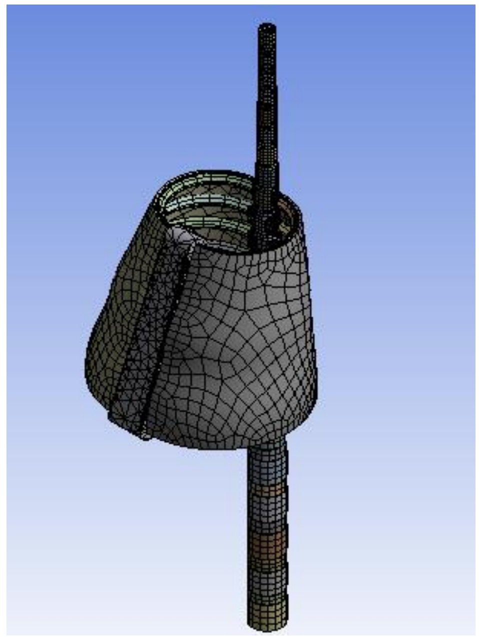

Figure 2.13: Meshing results for the 3D model with the brace

The same 20, 50, 70, and 100 N forces were applied to determine if there was the same 1:1 ratio that was found in the 2D spine model as well as in the initial 3D spine model and to determine if there is any significance in attaching the brace to the rib cage. Figure 2.10 from the first 3D spine model analysis shows how the boundary conditions were defined on this model. The bottom of the L5 vertebra is fixed to eliminate all degrees of freedom on that end of the spine. The top of the brace is only free to translate in the y-direction as well as rotate about the z-axis. The forces were applied to the same fifth, sixth, and seventh vertebrae while the other ribs were held fixed simulating what the brace would naturally be doing. In addition to the three ribs experiencing the lateral applied forces to simulate a thoracic scoliosis curve, the eighth and ninth vertebrae had forces applied to them to simulate the thoracic lumbosacral scoliosis in a separate scenario. In thoracic scoliosis the curvature of the spine occurs in the thoracic region and thus the reason for applying pressures to ribs five, six, and seven. One of the other forms of scoliosis is thoracic lumbosacral scoliosis that occurs when the curvature targets the thoracic region as well as the lumbar and sacral regions; to model this, ribs eight and nine were the ribs where the pressure was applied. Ribs three, four, five, and six were subject to a fixed boundary condition, simulating the brace being on that side. To treat thoracic lumbosacral scoliosis a Thoracic Lumbosacral Orthosis or TLSO is used for rehabilitation. A TLSO is commonly referred to as a Boston brace, which is the same brace donated by the TSRCH. Analyzing the spine under this specific loading condition will provide a better understanding of how stress and deformations are distributed with the Boston brace.

2.4.2 Simulating the Loading of the Two Braces

Once the forces were applied, the maximum stress and deformation were calculated once again along with their maximum locations. Table 2.6 details the results found from the simulations while analyzing it as a thoracic scoliosis curvature.

Table 2.6: Stress and deformation results for the 3D model as a Thoracic brace

Force (N)	Maximum von Mises Stress (MPa)	Maximum Deformation (mm)
20	0.036	0.009
50	0.091	0.023
70	0.128	0.032
100	0.180	0.046

The location of these maximum stress and deformation results stayed fixed throughout the various loadings conditions. The maximum stress value was found on the anterior and posterior sides of the T6 vertebra, and the maximum deformation was located on the left side of the T6T7 disk.

Simulating the model as if a TLSO was being worn by the child will not only be beneficial to understanding how that specific brace works but also make it possible to compare with the original thoracic scoliosis curvature. Table 2.7 shows the results from applying the forces to the bottom of the rib cage instead of in the middle for a TLSO.

Table 2.7: Stress and deformation results for the 3D model with a TLSO

Force (N)	Maximum von Mises Stress (MPa)	Maximum Deformation (mm)
20	0.089	0.020
50	0.222	0.049
70	0.304	0.069
100	0.436	0.099

When the forces were applied to the lower ribs, the locations of the maximum stress and deformation naturally moved to these lower locations. The maximum stress was located on the posterior side of the T9 vertebra and the maximum deformation occurred on the left side of the T9T10 disk.

2.4.3 Explanation of Results

The results of these forces as functions of stress and deformation are similar to the previous results in there is still the almost 1:1 ratio between applied force and stress or deformation as observed in all loading and model scenarios.

This illustrates that under these loading conditions the spine, ribs, and brace all stay within the elastic region. Therefore, whether it is a simple 2D model of the spine, a 3D model of the spine and rib cage, or a 3D model of the spine and rib cage incased in a brace there is almost a linear relationship between the applied force and the von Mises stress or deformation. Figures 2.14 and 2.15 show the results for the 20 N applied force and Figures 2.16 and 2.17 show the results for the 100 N applied force. For all of the figures, the brace was hidden to better show how the ribs and spine responded to these loads instead of the brace.

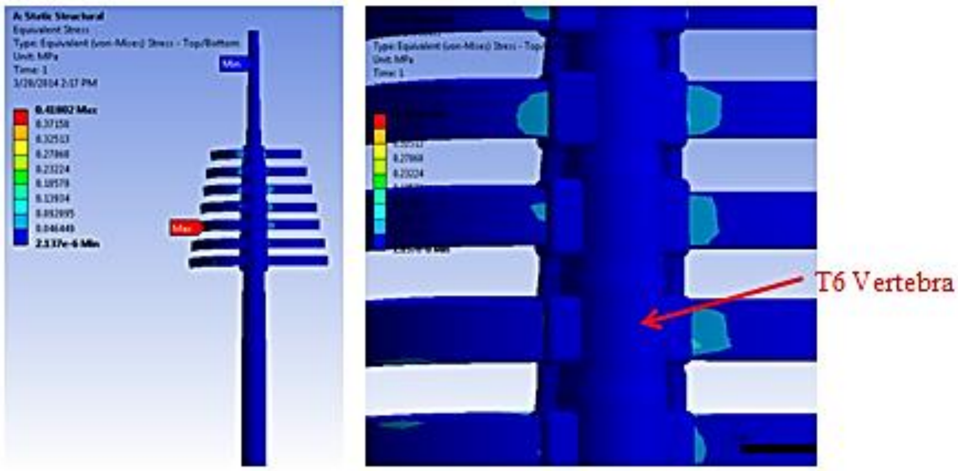

Figure 2.14: Stress results of the spine under 20 N; maximum stress on the spine is found on the T6 vertebra

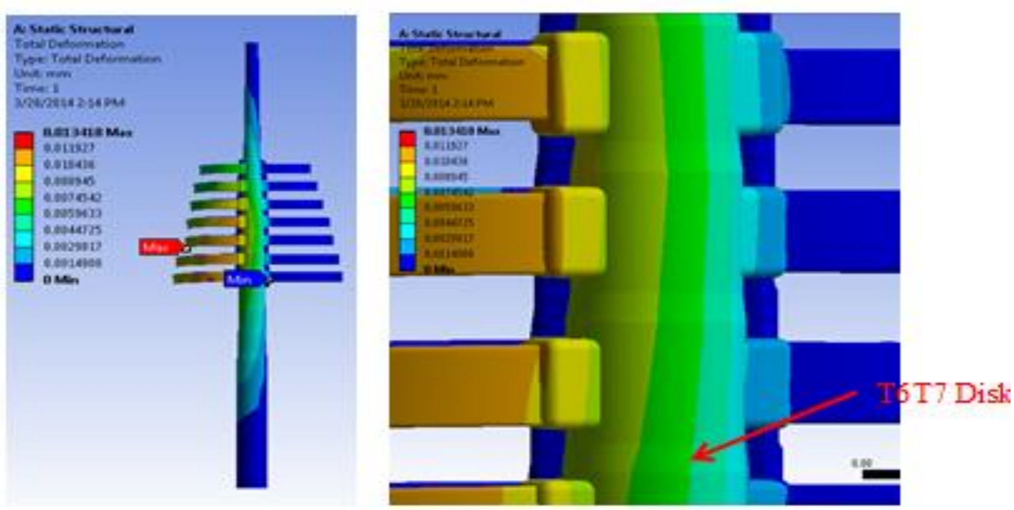

Figure 2.15: Deformation results of the spine under 20 N; maximum deformation of the spine is found on the T6T7 disk

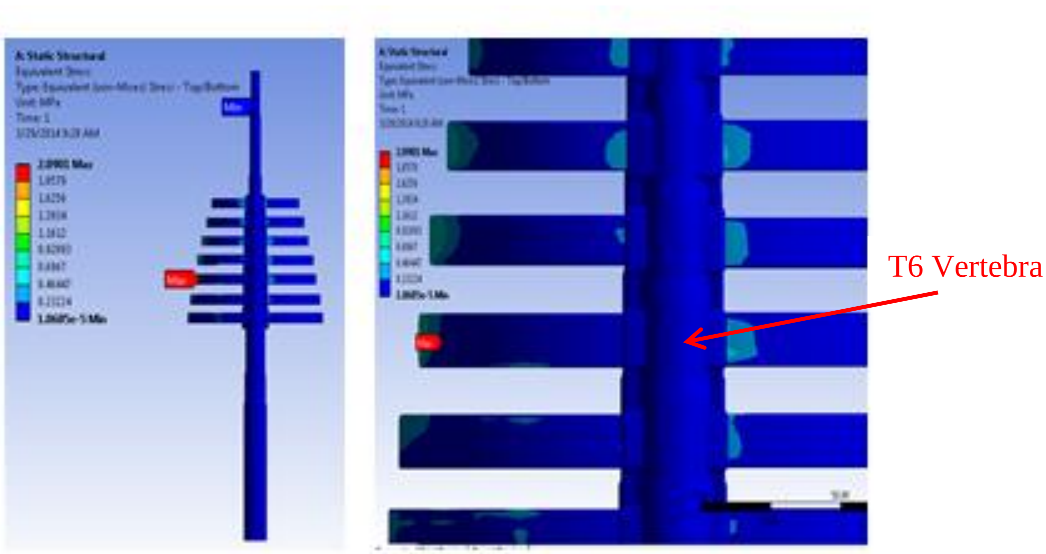

Figure 2.16: Stress results of the spine under 100 N; maximum stress on the spine is found on the T6 vertebra

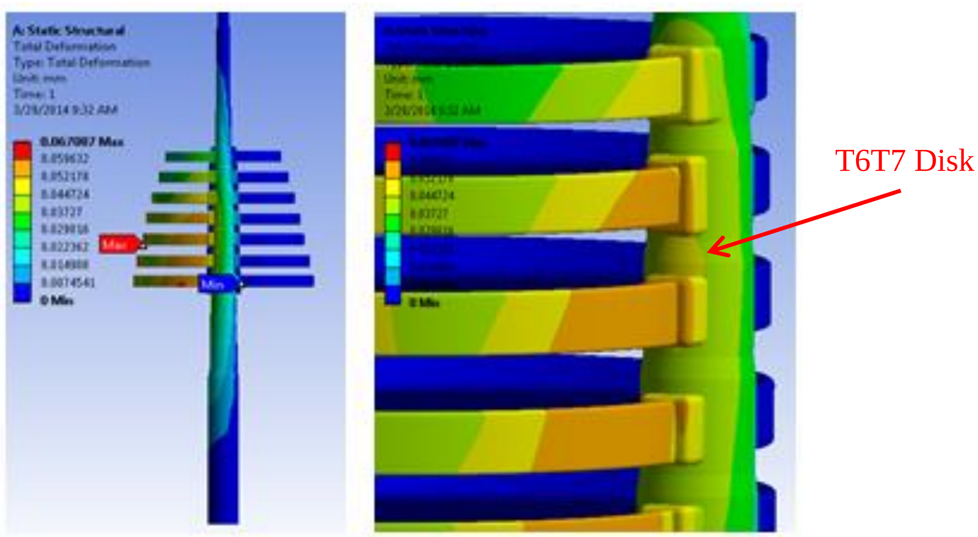

Figure 2.17: Deformation results of the spine under 100 N; maximum deformation of the spine is found on the T6T7 disk

The deformations in the spinal column range from the middle of the cervical vertebrae down to the middle of the lumbar. This shows a wider range of affected vertebrae than without the brace. This even further verifies that having the brace modeled with the spine and rib cage give better indications of how the spinal column reacts to lateral applied loads. The brace distributes the applied pressures throughout the rest of the body letting it have a bigger impact on the spine. Though the maximum values or locations of the stress and deformation do not vary from modeling the spine with or without the brace; having the brace attached to the rib cage distributes the applied loads throughout the spine.

The TLSO bracing results as shown in Table 2.7 indicate an almost one hundred percent increase in maximum stress and deformation results compared to simulating the brace as a thoracic scoliosis curve. As with the other models, when there is a pressure applied to the lateral side of the body the pressure is felt directly on the spine. This is only the case because soft tissues and organs which would normally absorb some of the energy were not modeled. Though having the tissue and organs would make for a more realistic case, the level of detail would be too extreme. Having different material properties changing from person to person and then from age to age also makes it extremely difficult to get a true model of how the curvature would respond to various loads. Simply modeling the disks, bones, and some tissue will provide a sufficient idea of how the spine would respond to various loading scenarios. The model that includes the brace attached to the rib cage gives the best representation of how the spine will respond to various applied loads. A better understanding of how the stress and deformation are distributed throughout the spinal column can easily be attained from this final model. A Cobb angle of 4.58 degrees was calculated from the thoracic brace scenario and a Cobb angle of 4.35 degrees was found from the TLSO. Applying more pressure to the side of the body would increase the Cobb angle to a more noticeable value or if the simulation was run under a dynamic loading scenario for longer durations. These models illustrate how the stress and deformation are distributed throughout the rest of the body instead of focusing on obtaining noticeable Cobb angles.

2.4.4 Validation of Results

The spine with brace simulation analyzed different applied loads to give a better understanding of how the spinal column responds when under pressure. The same validation experiment that was used on the 3D spine model without the brace was used here again to verify that there would be equal distributions. The brace was initially loaded with symmetric loads to ensure that there would be a symmetric stress and deformation distribution throughout the rest of the body. After applying the 20 N load to both the left and right areas where the sixth rib is located the stress and deformation distributions were symmetric down the center of the brace and throughout the spine. This in combination with the validation from the 2D spinal modeling shows that the model is responding to these various applied loads as expected. Though this model includes a rib cage and brace, the spine will still show characteristics of the deformation that was experienced in the 2D model.

2.5 Overall Results from Modelling and Simulation

When analyzing the Cobb angle generated from the various loads and comparing it with information from other sources, the developed model is close to yielding similar results. In a 2000 publication by Gignac et al., [17], the initial Cobb angle from evaluating 20 patients with scoliosis showed a 32 degree curvature in the thoracic region with a standard deviation of 6 degrees. When a Boston brace was worn by the patient the Cobb angle dropped down to 30 degrees with the same standard deviation of 6. This shows that when the patient wore the brace their Cobb angle decreased by 2 degrees when strap tensions are on average of approximately 60 N. [5] The Cobb angle that was derived from the spine under the applied load of 100 N was determined to be approximately 4.7 degrees; therefore, with a 50 N load the Cobb angle is expected to be 2.4 degrees. This is extremely close to the results that were calculated having a correction of 2 degrees with the patient wearing the Boston brace reported in [17]. The patient wearing the brace and having it straighten the spine is similar to the static load of 50 N used in the simulations. This shows that this model yields reasonable results on how the spine will be corrected by simply wearing the Boston brace.

The information in the same 2000 publication by Gignac et al., [17], has spine corrections for the thoracic-lumbar region as well. The average Cobb angle for the patients that had a thoracic-lumbar scoliosis curve was 34 degrees with a standard deviation of 7 degrees. When the patient was put in the Boston brace under the same average strap tension as reported in [5] the results showed a decline in the angle of 3 degrees. Comparing this change in Cobb angle with what was calculated in the TLSO, the results show there is a difference of less than a degree. This indicates that the modeling of the TLSO to simulate a thoracic-lumbar scoliosis curve is accurate with actual data from patients and can predict how the spine will move when a brace is put on.

The analysis of how the brace immediately affects a thoracic and thoracic-lumbar curve was reported in a paper written by Desbiens-Blais et al., [5]. The results from this publication show a change of 16 degrees in Cobb angle with a standard deviation of 10 degrees for a thoracic scoliosis curve when the patient wore their TLSO. When analyzing the thoracic-lumbar scoliosis curve there was an improvement of the Cobb angle of 13 degrees with a standard deviation of 6.5 degrees. The two models that had the brace surrounding the rib cage showed a 2.29 degrees and 2.18 degrees improvement in the Cobb angle respectively. Though these numbers are lower than the averages that were found in this analysis there is a slight resemblance and proves that the models are close to modeling the spinal change.

Analyzing these results compared to the information that was found in these two publications indicate a high level of confidence in the models. Though there is not a lot of difference between the final results in Cobb angle between modeling the spine with or without the brace the overall stress and deformation distributions are much greater. If the model was run under a dynamic loading condition that showed how the spine would correct day to day a better understanding of the entire rehabilitation process could be found. These models closely resemble what happens to the spine when the normal load of 50 N is applied to the brace, and a better knowledge of treatment can be understood.

Chapter 3

Outfitting a Brace with Sensors and Data Acquisition

3.1 Introduction to Data Acquisition

Data collection of strains values that a scoliosis brace is experiencing while being worn is in the initial stages. New research collects pressure values from inside the brace and the tension in the straps. [20 and 21] Using these measurements, a new or better understanding of what the body is experiencing during rehabilitation is being determined. This chapter will focus on the procedures followed in order to obtain qualitative measurements from the brace with the use of strain gauges and pressure sensors.

The strain gauges used in the experimentation are from Measurements Group, Inc. and are a combination of unidirectional lateral strain gauges and forty-five degrees rosette strain gauges. All of the strain gauges were 120 ± 0.4% Ohm gauges with a nominal gage factor of 2.07 at 24 degrees Celsius. In addition, two FlexiForce force sensors (Tekscan, South Boston, MA) were used in the initial stages of collecting data from the brace. These force sensors are the FlexiForce A201 model sensors, 7.75 inches long, have a <±3% linearity error and a force range up to 440 N. The data from the sensors was acquired, processed, and displayed by developing custom VIs (virtual instruments) in LabVIEW 2013 Student Edition from National Instruments. The USB 6009 data acquisition device with 14 bit analog to digital conversion was used to acquire and import into LabVIEW. A 14 bit data acquisition implies is a resolution of 1/2^14 on a -10 V to 10 V range.

Experimentation started by first acquiring signals for one strain gauge on a custom developed analog circuit breadboard and then acquiring signals from three strain gauges with the use of bridge completions modules. The brace outfitted with sensors was tested under extreme compression and expansion loading scenarios and by having a person that does not have scoliosis wear the brace to obtain information when the brace is actually on a patient. Since the brace was not custom made for this person only qualitative data was obtained.

A bridge completion module made by Omega Inc. was introduced to simplify the circuit instead of having the breadboard to read changes in the voltage. This device took the place of the circuit breadboard and simplified the wiring used in collecting data. Once the bridge completion module was connected and its results matched that of the breadboard, two more bridge completion modules were added to collect data from all three lateral strain gauges. The use of these devices to obtain three signals from the brace is the first stepping stone in collected information from the brace in real time.

3.2 LabVIEW Data Acquisition with Breadboard

For the initial data acquisition setup, three rosette strain gauges were attached vertically down the center of the brace as shown in Figure 3.1. The rosette strain gauges were installed to measure the expansion or compression of the brace primarily in the hoop direction. The rosette strain gauges were selected because they could be used to measure deformation in different directions due to any arbitrary deformation of the brace.

The breadboard design was equivalent to the setup used in the MAE 3183 course, Measurements II lab, at the University of Texas at Arlington. [26] This setup acquired the raw voltage data from the strain gauge, amplified it though an operational amplifier, and displayed in LabVIEW. A combination of various resistors (120, 10k, 75k, 100k, and 1M Ohm), a 0.1 µF capacitor and a quad operational amplifier all manufactured by NTE Electronics were used in creating the circuit. The NTE987 quad operational amplifier was the center piece of this circuit board by taking the original signal, amplifying it through a combination of resistors, and then sending the amplified data to LabVIEW. The prototyped board is shown in Figure 3.2. A 5 V external power supply regulator was initially used in the experiment to guarantee that a constant excitation voltage was being supplied to the breadboard.

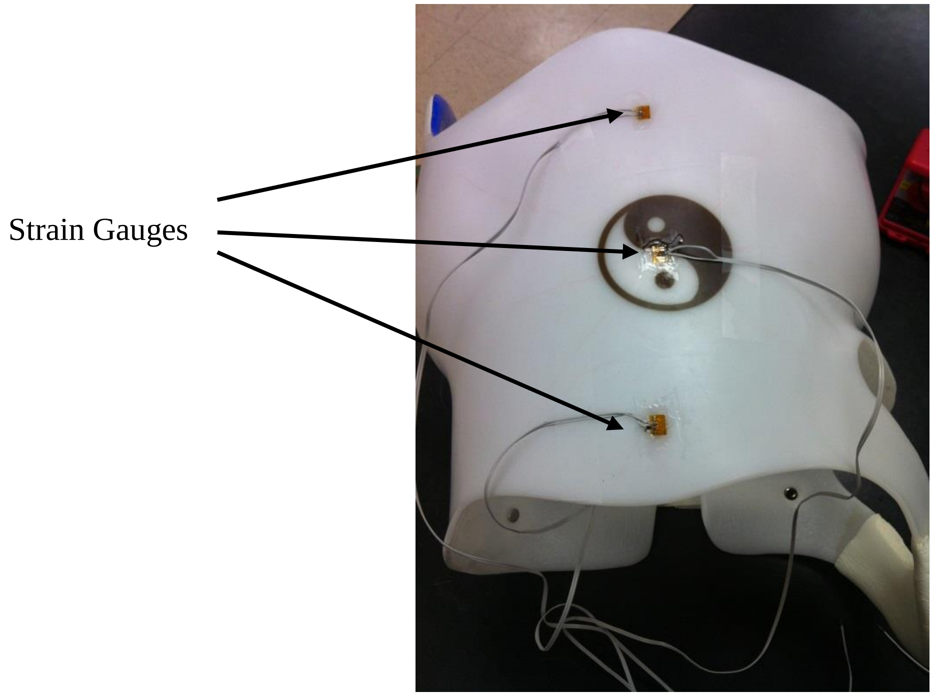

Figure 3.1: Boston Brace outfitted with three strain gauges

In order to ensure that there is an accurate calibration of the strain gauges and DAQ set up, it is recommend to always use a constant voltage regulated external voltage supply or a voltage regulator. This circuit setup would also be used to gather data from the two FlexiForce pressure sensors that could be attached to the inside of the brace underneath the padding. Each of the sensors was placed underneath the padding of the brace to obtain information about how much force the side of the brace was applying to the body. These locations were intentionally targeted because the padding build up is used to push against the spine to correct scoliosis. Qualitative voltage readings were collected from the force sensors indicating that the forces applied to the body by the brace could be estimated from these readings with the correct sensor calibration.

Once the strain gauge registered a change in voltage due to deformation in the brace, and the signal was amplified, it was further processed in digital form in a LabVIEW VI.

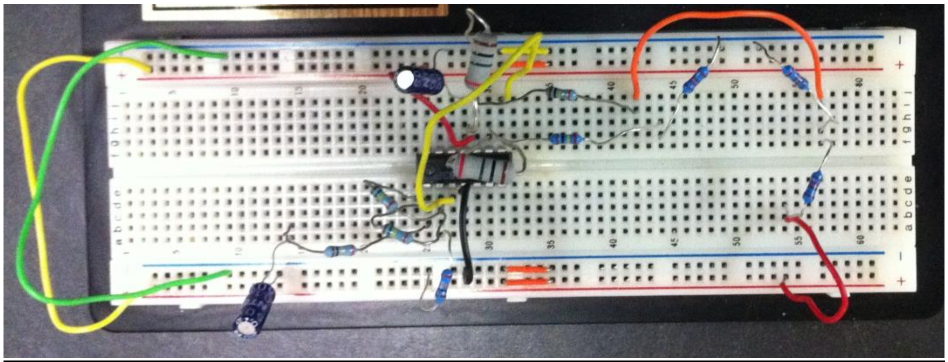

Figure 3.2: Data acquisition circuit on the breadboard

The LabVIEW VI used in collecting this single strain gauge signal (voltage) is shown in Figure 3.3. The original collected data is digitally filtered to remove unwanted high frequency noise with the use of a Butterworth filter available in LabVIEW since no filtering is performed in hardware.

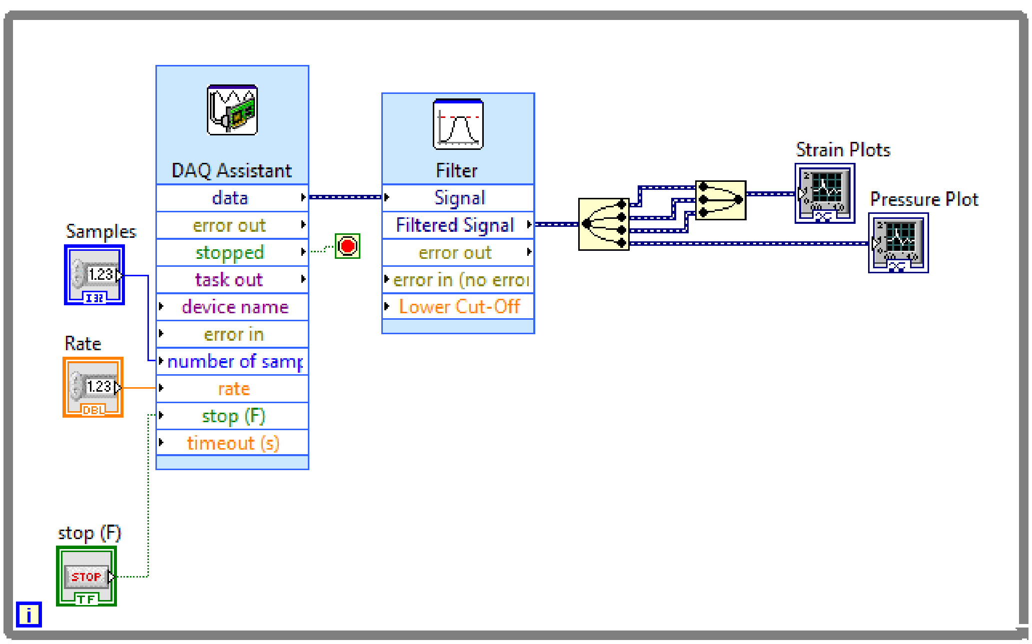

Figure 3.3: Initial LabVIEW VI used with the initial breadboard design

The selected Butterworth filter is a third order, low pass filter, with a cutoff frequency of 7 Hz. The Butterworth filter will ensure that only the low frequency strain gauge signals will be displayed and recorded while the high frequency noise will be eliminated.

For the initial phases of data collection from the brace, the center lateral strain gauge was the only one used and connected to the breadboard. This initial phase of collecting data from the brace showed how the strain changed depending on how the brace was flexed. Five different loading scenarios were used and strains recorded to gain a better understanding of the brace flexing and what strain it experiences. The five loading scenarios are the following: leaving the brace on the table to ensure it does not experience any strain, manually compressing and expanding the brace to extreme conditions, fitting the brace on a human subject, and then putting the brace on the human subject and tightening it past comfort levels.

3.3 Data Acquisition with Breadboard Results

The strains (after converting the acquired voltage to strain) obtained from the bottom, center and top-strain gauges are shown in Table 3.1.

Table 3.1: Data Acquisition Phase I strain results

μStrain	Brace Compressed	Subject No Padding2	Subject No Padding1	Brace Rest	Brace Expanded
Bottom	-7510.14	-	-	0	5024.86
Center	-4367.26	-1091.04	-339.33	0	3540.31
Top	-140.48	-	-	0	-130.29

This information shows the results from the brace resting on the table with zero strain, and then when the brace was expanded and compressed. For the expansion, the brace was pulled apart at the back by hand, expanding it as wide as possible without causing any damage to the brace. The brace was also compressed by pushing on the sides manually to be as compacted as possible. In both of these cases the actual force or pressure that was exerted was unknown. These two extreme scenarios were performed to develop a range of values for a patient wearing the brace.

The other two loading scenarios of Subject No Padding2 and Subject No Padding1were obtained by actually putting the brace on a human subject. The subject did not have scoliosis and was too tall for the brace to function correctly but small enough that qualitative values could be obtained from a human patient. For the human subject only the center strain gauge was used for data collection. Since the available brace was not custom made for the human subject, the top of the brace was not resting against the subject's chest and the bottom portion of the brace where the strain gauge was located was resting against the subject's belt. The center strain gauge was deemed to be the best candidate for acquiring reasonable qualitative values for strain that the brace was experiencing. The results from the center strain gauge for each of the five loading scenarios are shown in Figure 3.4

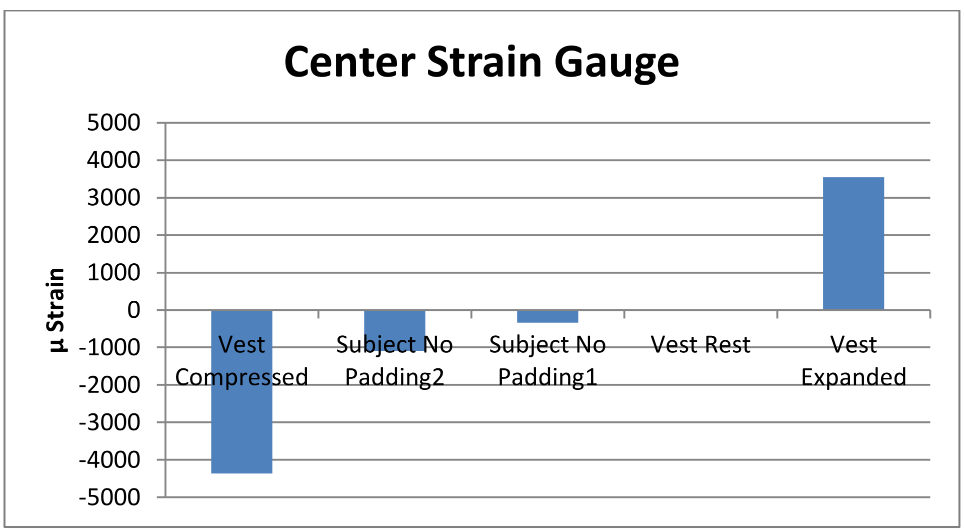

Figure 3.4: µStrain results from the center strain gauge

Therefore, based on the results from using the breadboard it can be concluded that the procedure and methodology for attaching the strain gauges, interfacing the strain gauges with the breadboard, and later with LabVIEW are correct. Qualitative data can be recorded from the brace under different loading scenarios,-processed and recorded.

3.4 LabVIEW Data Acquisition Bridge Completion Modules

After confirming that data could be obtained from the three strain gauges and the usage of a single breadboard, more strain gauges were attached to the brace along with the use of three bridge completion modules. These modules replaced the Wheatstone bridge and did not amplify or filter the signal.The usage of three completion modules allows three signals to be obtained from different strain gauges simultaneously to better understand what the brace as a whole is experiencing. Six new uniaxial strain gauges were attached alongside two rosette strain gauges in different orientations along the centerline of the brace as shown in Figure 3.5 with two more uniaxial strain gauges on the left side (not shown).

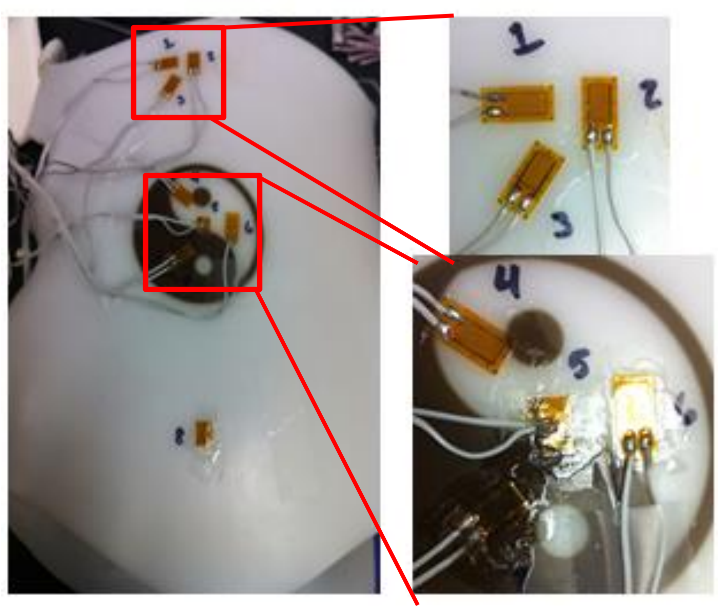

Figure 3.5: Additional strain gauge locations and orientations

The strain gauges attached vertically down the front of the brace would provide a clear stress distribution of what the body experiences from the brace when lateral forces are applied. In order to still gain a better understanding of how the brace responds to these loads the strain gauges were arranged in formations similar to rosettes since it was difficult to solder on all the interface soldering pads of the small rosettes. The new strain gauge arrangement with the uniaxial gauges will help provide a better understanding of how the top portion of the brace adjusts to lateral forces compared to other regions. The grouping of lateral strain gauges that surround the second rosette are positioned as such to gain a better understanding of how the middle portion of the brace differs from the top. There are two strain gauges that run up and down the brace in the axial direction and are used to calculate the hoop stress that the brace is experiencing. No major strain values are expected from the brace in the axial direction, but axial strain values are needed when calculating the hoop stress. The bottom of the brace is only monitored by a single rosette strain gauge that is measuring lateral changes. The last grouping of strain gauges is a combination of measuring lateral and axial changes on the left side of the brace. Though there is not expected to be significant strain located on the side of the brace based on the loading scenarios, the two strain gauges were positioned laterally and axially to measure data on the hip of the patient. This was deemed to be a key location based on input from patients and specialists at the Texas Scottish Rite Hospital for Children.

The signal from the strain gauges is sent to the bridge completion module as shown in Figure 3.6, and is then transferred through the USB 6009 data acquisition device to be processed and displayed in LabVIEW.

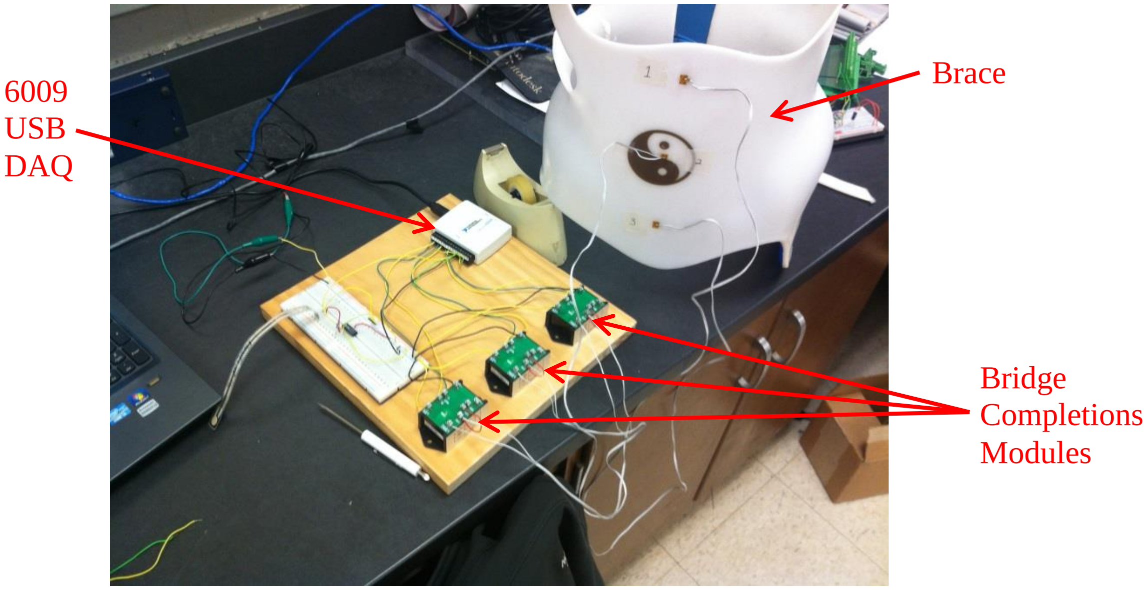

Figure 3.6: Bridge completion module setup

The signal is processed by the same Butterworth filter and then displayed in the LabVIEW VI. Once the average values have been obtained they are converted into strain values using Equation 3.1:

$$\varepsilon = \frac{-\,4\,V_r}{GF\,(1 + 2\,V_r)} \qquad (3.1)\ [25]$$

with *GF* the gauge factor and *Vr* the read voltage from the strain gauge. Finally, the hoop strain value is used in conjunction with the strain value from of the strain gauges that run axially on the brace to calculate the hoop and axial stress. As additional quantitative data is collected from the brace, realistic values of stress can be determined. The hoop and axial stress values could be calculated using Equations 3.2 and 3.3 respectively:

$$\sigma_h = \frac{E\,\varepsilon_h + v\,E\,\varepsilon_a}{1 + v^2} \qquad (3.2)$$

$$\sigma_a = \frac{E\,\varepsilon_h - \sigma_h}{-v} \tag{3.3}$$

with E the material Young Modulus, v the material Poisson's ratio, and ε_h and ε_a are the hoop strain and axial strain respectively.

The Boston brace is not a perfect cylinder but to obtain values for the hoop stress experienced by the patient, the assumption of the brace being a cylinder is used. Equations 3.2 and 3.3 are equations for stresses in cylinders and will provide a reasonable value for the stresses experienced by the brace. Once the stresses are determined, the equivalent pressure and the pressure that was applied could be calculated using Equation 3.4 with t being the thickness of the brace and d its diameter.

$$P = \frac{2\,\sigma_h\,t}{d} \qquad (3.4)\ [24]$$

Calculating the hoop and axial stress and then the applied pressure would be invaluable information to understanding scoliosis rehabilitation. In order to obtain these stress values, quantitative data must be collected and compared to the qualitative results that were sought in this research.

The LabVIEW VI used to acquire the data, filter it with the use of the Butterworth filter, display the current values, and then calculate the stresses experienced is shown in Figure 3.7. The inner loop shown in Figure 3.7 was developed to perform different calculations based on the Boolean switch that was installed on the front panel in LabVIEW. The switch was installed to be toggled on and off depending if an axial strain gauge was connected to the third bridge completion module as well as a lateral strain gauge connected into the second bridge completion module. When the two conditions were true, the LabVIEW VI would perform the calculations to determine the strain and stress values on the brace.

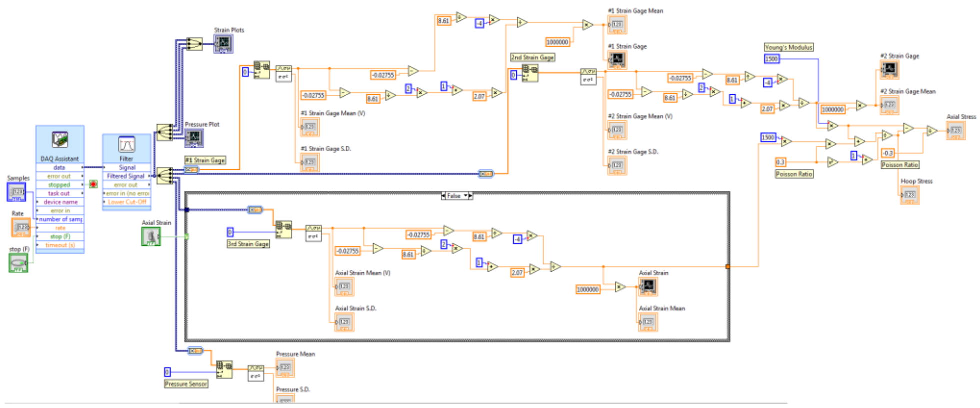

Figure 3.7: Lab VIEW VI developed for analyzing the data acquired with the bridge completion modules

This setup of the bridge completion module was presented to the Director of the Orthotics Department and the Residency Director, Education Coordinator, and Clinical Orthotist at the Texas Scottish Rite Hospital for Children on February 20, 2014. The results of that meeting were positive with the accomplishments thus far, such as attaching strain gauges to the brace and acquiring data from sensors. During the meeting emphasis on being able to collect data from key locations was discussed for the next phase of data collection from the brace. These key locations need to be identified in order to determine the stress values applied to the portion of the brace that hugs the hips. One of the biggest complaints by the patients is the pain experienced when the brace hugs the hips too tight. Though it is necessary to ensure that the brace is secure, determining the forces that are experienced in these locations could potentially lead to new design ideas and rehabilitation that would minimize or eliminate this concern.

3.5 Bridge Completion Module Results

The same test scenarios performed using the breadboard were repeated using the bridge completion modules. The brace was compressed as tight and expanded as wide as possible, lay un-deformed on a table, and then had a human subject that does not have scoliosis wear it. Gauge #8 was damaged during setup and preliminary testing and its results are not presented in this section. The results of the different testing scenarios are shown in Table 3.2.

Table 3.2: μStrain measurements from all strain gauges on the brace

Strain Gauge	Orientation	Compressed	Subject	Rest	Expansion
#1	Lateral	-3939.76	950.44	0	2166.16
#2	Axial	-189.94	1025.40	0	2209.59
#3	Forty-Five	-300.34	-739.64	0	-958.15
#4	Forty-Five	-2008.39	-730.82	0	1573.10
#5	Lateral	-3681.11	-66.72	0	4813.24
#6	Axial	479.65	-118.03	0	-765.34
#7	Forty-Five	-2854.35	-974.18	0	1463.10
#9	Axial	-46.19	1421.21	0	2487.95
#10	Lateral	-1362.02	-1609.13	0	1012.15

Since there are nine strain gauges and only three bridge completion modules, data collection had to be accomplished in three different stages. In order to keep the scenarios as close to the same as the others, each of the four loading scenarios was performed for all working strain gauges. After the first group was tested under one of the loading scenarios, the next group of strain gauges were connected with the help of an assistant and subjected to similar loading conditions as before. Once that specific loading scenario was completed, the next scenario was performed for all strain gauges throughout the rest of the loading scenarios. Therefore, it needs to be established that not every loading scenario was repeated exactly the same and one group of strain gauges could have been under a slightly different load. The best attempt to avoid applying different loads was performed by testing each scenario sequentially. The top grouping of the strain gauges were collected first with all three strain gauges attached to each of the three completion modules. The next grouping of strain gauges was the fourth, fifth, and sixth strain gauges as designated on the brace. Consequently, the last group of strain gauges that was tested was the seventh gauge in the middle of the brace along with the lateral (nine) and axial (ten) strain gauges that were attached to the left side of the brace.

In a realistic scenario where a patient is wearing the brace for rehabilitation purposes, the lateral strain gauges are deemed to be the most important since their measurements could be used to estimate the hoop stress and forces applied or exerted by the brace on the body. Therefore, the results from the lateral strain gauges are presented in Figures 3.8 to 3.10.

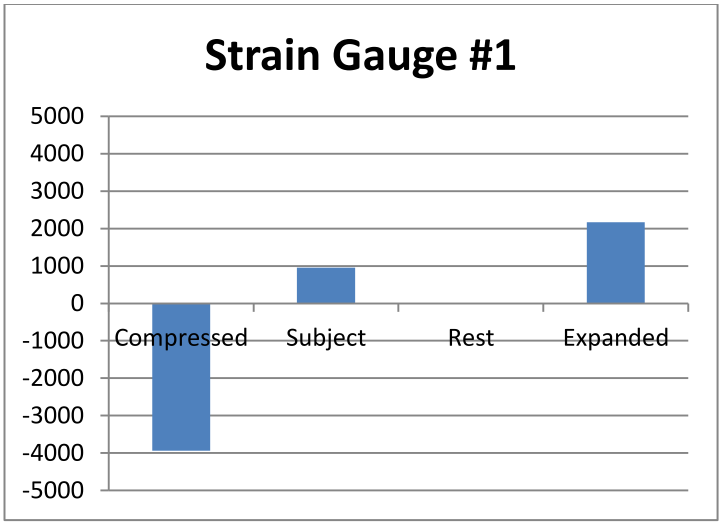

Figure 3.8: μStrain responses from strain gauge #1

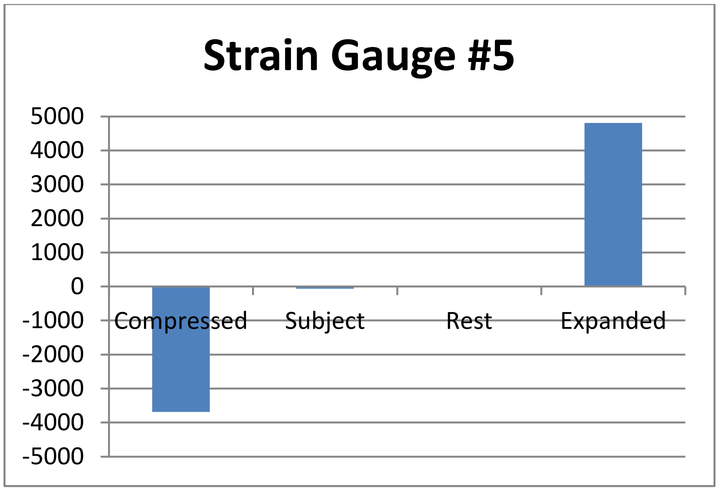

Figure 3.9: μStrain responses from strain gauge #5

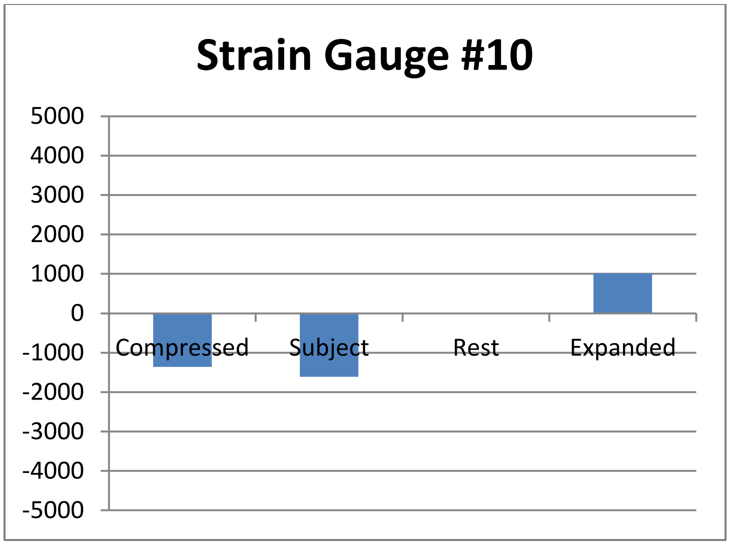

Figure 3.10: µStrain responses from strain gauge #10

The results from strain gauge #10 seem suspect but indicate the expansion of the brace in the posterior-anterior direction. The results show an increase in strain with a subject simply wearing the brace instead of obtaining data from the brace by expanding or compressing. Both the expanding and compressing scenarios of the brace are mainly concentrated on affecting the front of the brace instead of the side. Patients wearing the brace will cause it to expand in the posterior-anterior direction more than the sagittal direction.

The axial strain gauges were used to calculate the hoop stress experienced by the brace. Determining the stress would be a giant step forward in determining what forces and pressures the body was experiencing when the patient was wearing the brace. Figures 3.11 and 3.12 show how the front of the brace responds to axial strains while Figure 3.13 shows how the side of the brace will respond based in the loading scenario. Strain gauges #2 and #6 show similar results in how the front of the brace responds to axial deformation but they are polar opposites indicating the body may be pushing the brace out in the vicinity of strain gauge #6. Strain gauge #9 is similar to strain gauge #10 in that the compression and expansion of the brace targets the front of the brace instead of the side. This provides a better indication of the axial strain of a patient wearing it rather than how it responds to compressional and expansion loads.

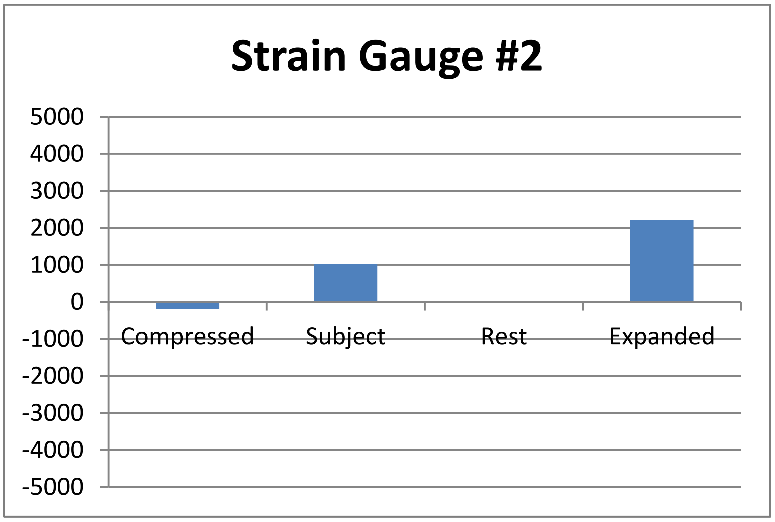

Figure 3.11: μStrain responses from strain gauge #2

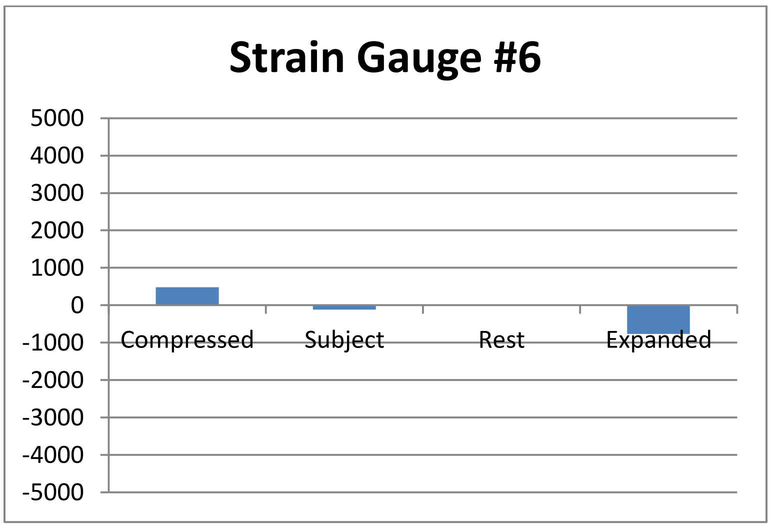

Figure 3.12: μStrain responses from strain gauge #6

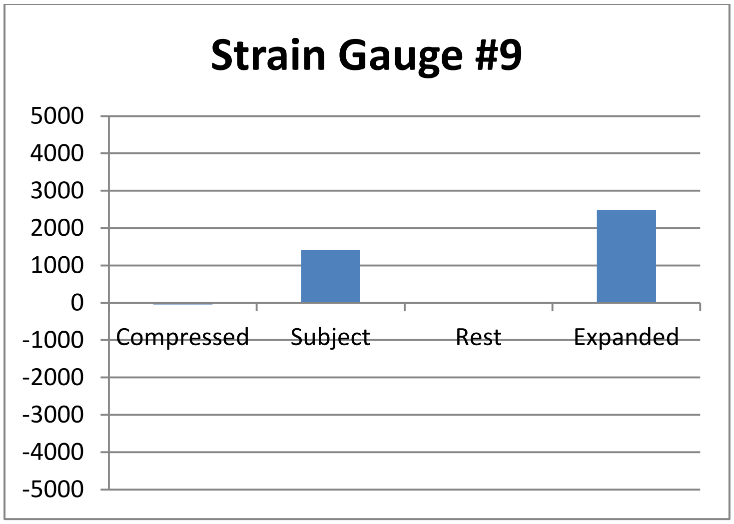

Figure 3.13: µStrain responses from strain gauge #9

The last three strain gauges show how the brace responds to compression and expansion loads on approximately forty-five degree angles and, as expected, both compression and expansion are sensed. Figures 3.14 through 3.16 show the general shape of strain distributions from compressed through expanded scenarios.

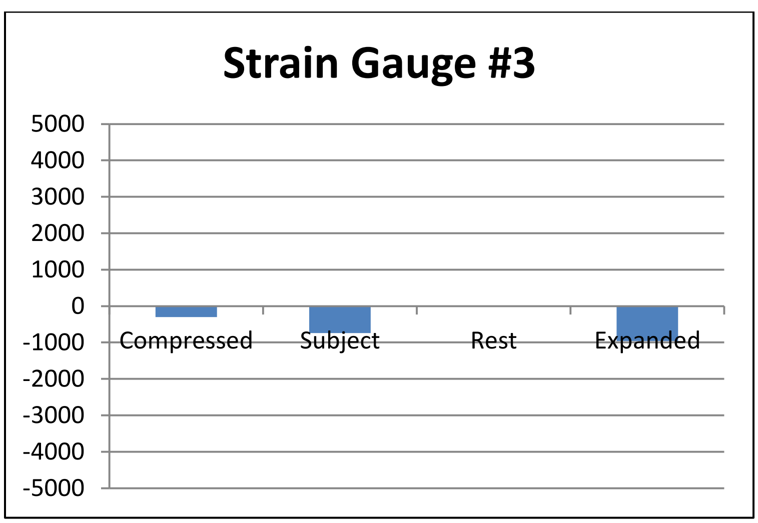

Figure 3.14: µStrain responses from strain gauge #3

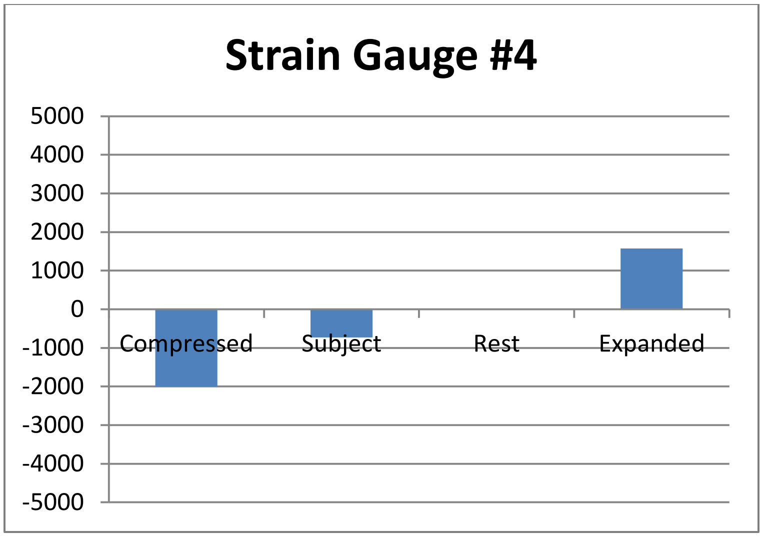

Figure 3.15: µStrain responses from strain gauge #4

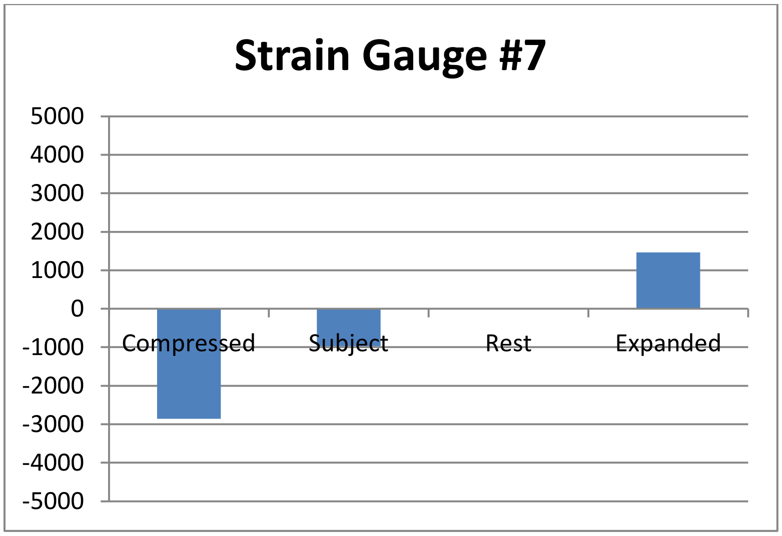

Figure 3.16: µStrain responses from strain gauge #7

Stress values for the brace could be determined using Equations 3.2 and 3.3. Obtaining the stress values that the brace experiences could make it possible to use the information in the FE models and predict the rehabilitation progression.

The stress results shown in Table 3.3 should be taken as purely qualitative information. This shows the procedure that could be followed to obtain quantitative values from the brace. In this scenario, it was observed that the values of the strain gauges did fluctuate and should only be considered as qualitative information. Due to the irregular shape of the brace more research needs to be performed in understanding how the brace responds to various applied loads before-numbers obtained from the brace are used for rehabilitation purposes. Table 3.3 shows the simulated stress values at the top, middle, and left side of the brace from the subject's data; note that there are certain conditions relating to the human and brace geometry as already discussed in Section 3.1. This information is to illustrate the procedure that could be followed to calculate stress values from quantitative data. These locations were targeted due to their importance in determining what is happening when forces are applied to the side of the body.

Table 3.3: Stress and pressure values experienced by the brace from the human subject

	TOP	MIDDLE	SIDE
Hoop Stress (MPa)	1.73	-0.14	-1.63
Axial Stress (MPa)	0.91	-0.14	2.71
Applied Pressure (MPa)	0.67	-0.06	-0.63

These results show how much the top of the brace is compressing the body compared to how much compression is felt in the middle. Though there will not be a stress value generated from the bottom of the brace a reasonable understanding of how this region responds to rehabilitation forces can be determined. Along with computing the stresses experienced in these specific regions the pressures associated with the hoop stress was calculated using Equation 3.4.

The discussion in this chapter represents the beginning stages of obtaining real time data from sensors attached on the brace. These results show how the brace is applying pressure to the patient and where maximum rehabilitation efforts need to be focused. Gathering data from a scoliosis brace and determining the pressures that are applied to the body will make it possible to constantly monitor the tightness of the brace. Ensuring the brace is always pulled to a certain level will aid and speed up the rehabilitation process for patients with scoliosis.

Chapter 4

Conclusion and Recommendations

4.1 Conclusion

The information found by this research is one of the first steps into improving scoliosis rehabilitation. The 2D and 3D spine models are extremely beneficial to understanding the stress and deformations that the spinal column experiences when applied loads from the brace occur. Though the 2D model does provide results that are extreme, it can still show how the spine will deform with applied forces to the lateral side of the body. The 3D model of the spine without the brace provides results that are exceptionally close to data collected from publications in the open literature. The 3D model can be the first stepping stone into developing a more complex model that can more accurately predict the desired applied loads. The proximity to the actual results obtained from scoliosis patients remains as close as the original 3D model when the brace is applied to incase the rib cage. Modeling the rib cage with the brace attached does not significantly change the Cobb angle with different applied forces but it does drastically change the stress and deformation distributions throughout the rest of the spinal column. Data collected and reported in [17] shows that with a thoracic scoliosis curve the 3D model of the spine without the brace is within 92 % of the actual results and within 87 % for the 3D model that has the bracing attached. For the thoracic-lumbar scoliosis curve the 3D model with the brace is within 85 % of the measured results. The data reported in [5] also depicts data from scoliosis patients with thoracic and thoracic-lumbar scoliosis curves. The 3D models without and with the bracing attached to the rib cage are within 15 % and 14 % respectively for patients that have a thoracic scoliosis curve. The TLSO brace spine model is within 18 % for the patients that have the thoracic-lumbar scoliosis. All of this reemphasis the validity of these models and shows that the models can be used to understand how much the spine will initially change when the patient first puts on the rehabilitation brace.

In order to independently use the models for analyzing how the brace will initially deform to applied loads there needs to be a way to determine what loads are being applied to the patient. Outfitting the brace with strain gauges was the first stepping stone into obtaining real time data from the brace as the patient is wearing it. The nine different strain gauges that are spread all over the brace will make it possible to obtain different readings from these various locations and determine the loads that the patient is under. Combining the information that is obtained from the brace and determining the pressures experienced can then be imported into the 3D models of the spine to determine what correction will take place. Being able to record the data and then immediately analyze it and evaluate how the spine will react is crucial to improving scoliosis rehabilitation. The results of this research can assist in continued development of obtaining real time data from the scoliosis brace, and in developing more complex models of the spine and the rest of the body to gain a better understanding of what will happen under various applied loads.

4.2 Future Recommendations and Research

The future research that can be performed by employing the procedures developed in this research is virtually limitless. The spine and brace model shows how they interact to lateral pressures. Specifically, the spine shows how the pressure is distributed from the ribs to the spine and sternum. With a more complex model the accuracy of the simulation would be even higher. Tendons, ligaments, cartilage, etc. could be modeled and incase the spine and disks while attaching to the ribs, thus providing a better indication of the damping that takes place when the ribs are compressed. Also using the models and applying more detail to the sternum such as the cartilage, tendons, and muscles would also provide a better indication of the damping that would happen on that side of the rib when it is compressed. Lastly, the spine could be modeled as its natural curve in and out of the coronal plane. This would show how the actual spine deforms when it is being pushed with a lateral pressure or force. The data analysis concept could be made smaller and more compact to be attached to the brace or worn in a small pouch. With a smaller, more compact data acquisition system there could be the possibility of obtaining more strain and pressure values from the brace to be analyzed.

In the future, the brace could be equipped with a micro-controller that would have programmed strain and pressure values that it needs to obtain and hold constant. The controller would obtain the data from these sensors, analyze the data, and then communicate with actuators that are attached to the back of the brace controlling the tension on the straps. This would be a closed loop system that is obtaining a reading every thirty minutes or hour and then controlling the brace as needed. The programmed strain and pressure values would be sent from either the hospital or the finite element model of the spine. Combining the strain and pressure values a required strap tension could be obtained and then sent to the brace controller for execution.

Combining all of these and implementing it on the brace could make it possible to track the progression of scoliosis and possibly reduce the treatment duration. The amount of time that the patient and guardians have to spend in the hospital could be drastically reduced and the amount of time they have to wait to get x-rays and amount of radiation endured to determine how treatment is progressing could be reduced as well. The possibilities and the knowledge that could be gained from this research would be priceless in treating and understanding scoliosis treatment and rehabilitation.

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Biographical Information

Ryan Neufeld graduated Cum Lade with his Bachelors of Science in Mechanical Engineering in the spring of 2012 after a three and a half collegiate career. During the completion of his bachelors Ryan join the mechanical engineering fast track program to accelerate the completion of his master’s degree. Ryan has an interest in aircraft, and controls systems, and hopes to apply skills gathered in his thesis work to broaden his capabilities of working within the aerospace industry. Currently Ryan holds an internship with Triumph Aerostructures and plans to continue in this field upon completion of his master’s degree. In the future Ryan plans to work through the corporate ladder and become a manager of research departments. He would also like to aid the military by moving into a manager position within the Department of Defense. The medical research that Ryan is conducting into scoliosis will allow him the opportunity to manage and conduct research into fields that he has not worked in before which will give him skills in overseeing a research department with many projects over various areas.