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Analyzing the Biomechanical Nature of Thoracic Kyphosis and Other Mid-Sagittal Spinal Deformities Using Finite Element Analysis

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ANALYZING THE BIOMECHANICAL NATURE OF THORACIC KYPHOSIS AND
OTHER MID SAGITTAL SPINAL DEFORMITIES USING FINITE ELEMENT
ANALYSIS TECHNIQUES

A Thesis
Presented to
the Graduate School of
Clemson University

In Partial Fulfillment
of the Requirements for the Degree
Master of Science
Bioengineering

by
Jaylin M. Carter
August 2017

Accepted by
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Dr. Jiro Nagatomi
Dr. Delphine Dean

ABSTRACT

Thoracic kyphosis is the mid-sagittal misalignment in the human thoracic spine. Occurring in both adults and children, this spinal deformity is caused by the likes of poor posture, genetics, osteoporosis and intervertebral disc degeneration. This disease results in the patient having a rounded or hump back appearance causing strain on muscles, internal organs and improper walking gate.

Corrections for this condition involve surgical implantation of metallic hardware to straighten the patient's posture. However, this treatment does not come without its own drawbacks such as a retrogressive forward head posture (FHP), which can occur, post-surgery. With the assistance of computer aided design and finite element analysis, we propose to link the cause of FHP to the surgical realignment of the thoracic spine.

DEDICATION

I dedicate this work to my family, friends and teachers.

ACKNOWLEDGEMENTS

I would like to thank my advisor Dr. Guigen Zhang for his guidance in my research. I am also thankful for all my lab mates for helping me in my work. I would also like to thank Dr. Timothy McHenry of the Greenville Health System for introducing this challenge to

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CHAPTER 1

GENERAL INTRODUCTION, ANATOMY OF THE SPINE AND KYPHOSIS

1.1 Introduction

The vertebral column is a vital and highly dynamic structure in the human body. It allows for flexible mobility of the body, support for the head and torso and protects the life sustaining spinal cord. By being a highly critical structure in the body, damages sustained to the spine will be amplified producing systemic effects on the body's health.

Because humans have adopted a primarily upright posture, they are prone to spinal disorders. Most spinal diseases are the symptoms of, or caused by misaligned curvatures in the spine. Kyphosis and scoliosis are two common spine misalignment types that present themselves in the sagittal and coronal plane respectively. The anatomical nature of kyphosis will be discussed alongside the current understanding of its causes and methods of treatment.

The basic building block of the human spine comprises the vertebrae; bone with a cylindrical body and various bony processes protruding from it. Vertebrae are stacked on one another to form a vertebral column. In between each vertebra is viscoelastic padding called intervertebral discs. The disc acts to adsorb compressional forces and add flexibility between the vertebrae.

1.2 Spinal Anatomy

The spine is classified into 5 major regions, cervical, thoracic, lumbar, sacral and coccyx (Fig. 1). Only cervical, thoracic and lumbar are able to move freely for that sacral and coccyx vertebrae are fused together. Cervical, which composes the neck, is constructed of 7 vertebrae and designated C1-C7. Thoracic, which supports the thoracic cavity, has 12 and designated T1-T12. Lumbar, which makes up the lower back contains 5 and designated L1-L5. The curves in each of these three mobile regions help support the weight in the body in the most efficient way possible. These three curves also have specific names: cervical lordosis, thoracic kyphosis, and lumbar lordosis, which relate to spines directional concavity (Fig. 2). A patient's thoracic kyphosis can be both referred to as an anatomical feature or a diseased state.

When a patient has kyphosis in the sense of a disease, it means that there is an over or under curvature in the thoracic curve. When this curvature becomes incorrect it can lead to misalignments in the patient's lumbar and cervical lordosis.

An individual vertebra is comprised of two primary structural regions; cortical and trabecular bone. Cortical bone is made of dense bony tissue which serves to protect and add structural support to the vertebrae. The cortical bone allows for the vertebrae to interlock with other vertebrae by attaching to the vertebral disc. It also functions to protect the spinal column and be the insertion point for various ligaments and muscles. The trabecular bone (also referred to as cancellous bone) is a lattice of inter weaved bone used to add structural support and keep the bone relatively light. It is

located in the interior of the vertebrae. Both regions are critical for proper movement in the skeletal system.

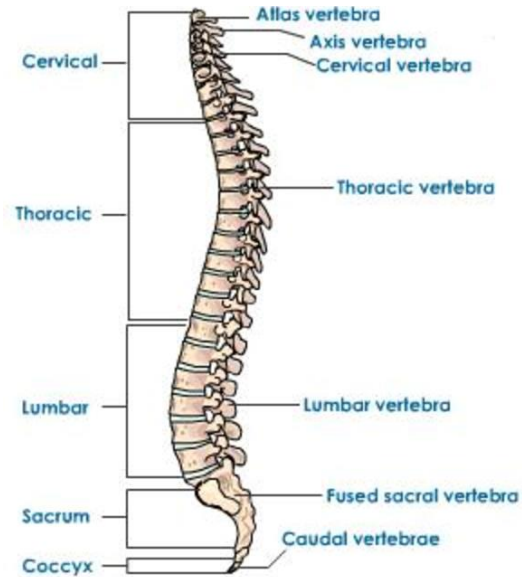


Figure 1: Regions of the Spine

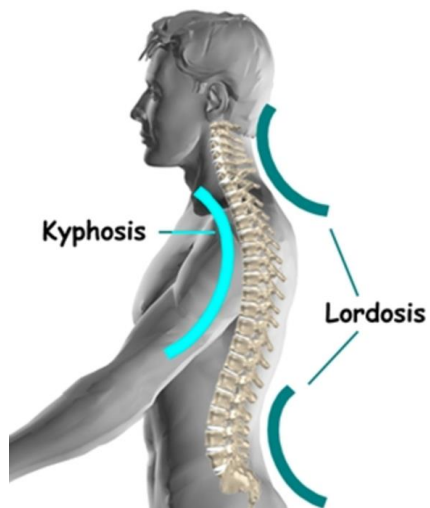


Figure 2: Kyphotic and Lordotic Curves

1.2.1 Vertebrae

The body of the vertebrae (Fig. 3) is the central hub that connects the vertebrae to its respective discs and the spinous processes. Protruding from the vertebral body are two pedicles, which form a hole with the lamina. This hole creates the vertebral foramen, a channel through which the spinal cord runs with adequate protection. From the lamina are the transverse processes, spinous processes, superior and inferior facets.

Inferior and superior facets on each vertebra form a cartilaginous joint that connects together adjacent vertebra. This helps stabilize the spine and prevents it from over straining.

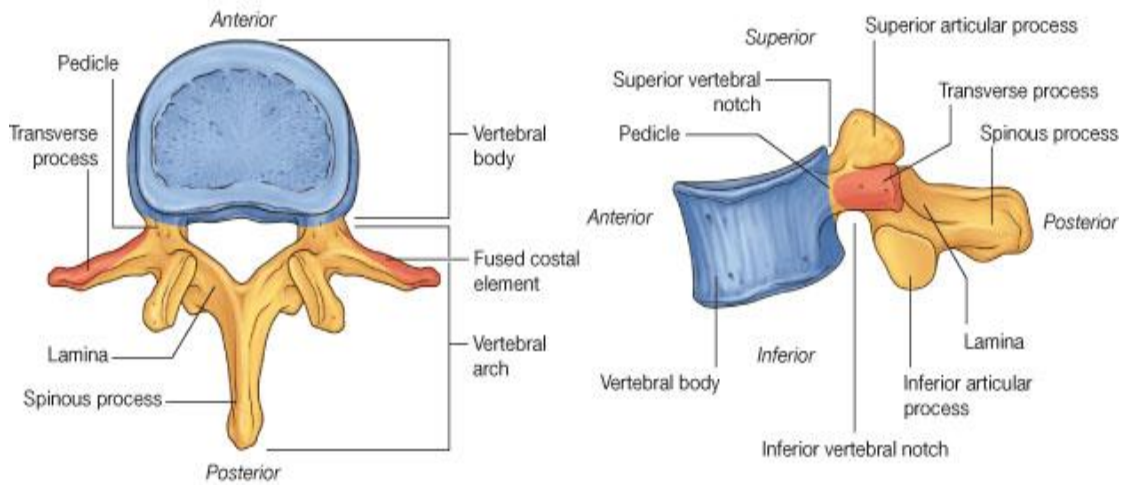


Figure 3: Vertebrae anatomy (thoracic)

1.2.2 Ligaments

The transverse and spinal processes are important for being the insertion and origin point for the various ligaments. The yellow Ligamentum Flavum (LF) is attached in between each of the lamina of adjacent vertebrae along the length of the spine. It specializes as a spinal stabilizer and spinal cord protector. When the body leans forward it will expand and thin. In contrast it contracts and thickens when the body leans backwards. Under some circumstances the LF will become too thick, this is a condition called Ligamentum Flavum hypertrophy. This can put immense pressure on the spinal cord causing pain and spinal damage (Fig. 4).

The Interspinous Ligament (IL) runs between the spinous processes of the vertebrae (Fig. 4). They help keep the intervertebral foramen clear of obstruction while the spine is flexing forward. In cases where IL is weak the intervertebral foramen can shrink pinching the nerves that exit through the foramen. This dysfunction is repaired with minor surgery.

The Anterior Longitudinal Ligament (ALL) runs along the anterior outer body of the spine touching the periosteum of the vertebral column (Fig. 4). It blankets both the vertebrae themselves and the intervertebral discs. Its primary purpose is to prevent over extension of the spine while the body is flexing backwards.

Similarly, the Posterior Longitudinal Ligament (PLL) lies on the posterior outer body of vertebrae medial to the pedicles (Fig. 4). It too blankets the vertebrae and intervertebral discs making it a continuous band from the base of the skull to the

sacrum. The PLL counteracts the ALL in that the PLL prevents the spine from over extending to far forward. Ligaments play a crucial role in the natural curvature and elasticity and flexibility that are innately characteristic to the spine.

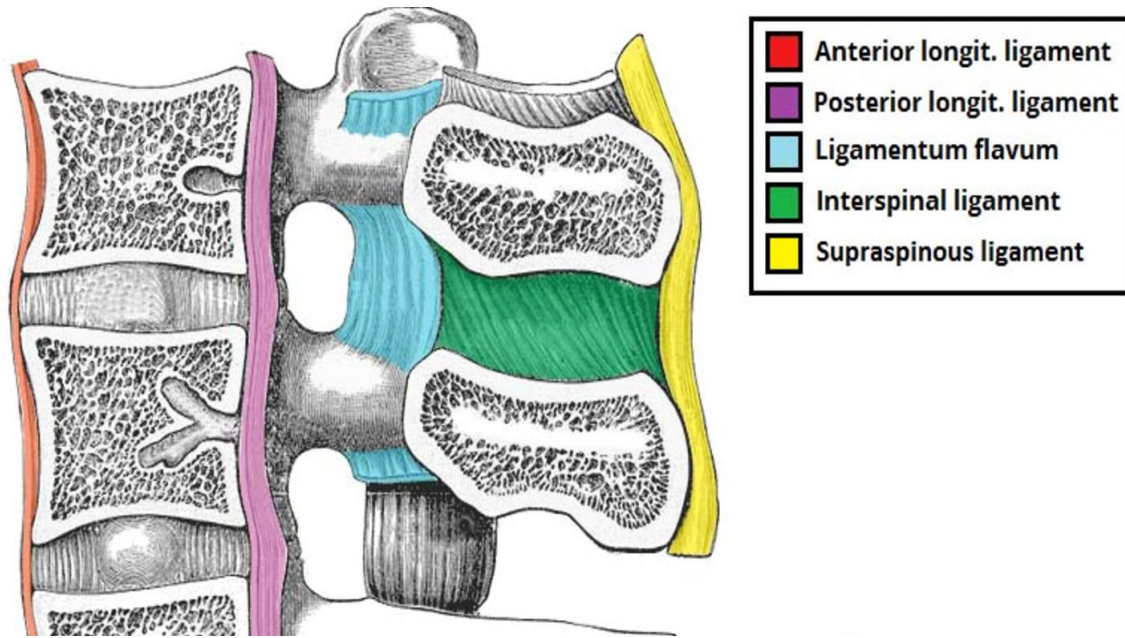


Figure 4: Layout of crucial spinal ligaments LF, IL, ALL, PLL

1.2.3 Intervertebral Disc

The intervertebral discs are fluid filled shock absorbers that help absorb the cyclic loads such as walking, running and jumping. Because of their unique constructions, the discs are stiff enough to add support to the spine but can become flexible on demand. As soon as the vertebral cartilage becomes displaced fluid is released giving the disc its gelatinous properties.

The intervertebral disc is essentially an enclosed membrane. Inside the membrane is the annulus made up of fibrocartilage. The fibrocartilage is created out of

collagen II and I. The collagen types are distributed in such a way as to establish both flexible and rigid regions of the disc. The fibrocartilage helps to evenly spread out stress along the body of the vertebrae in order to prevent fracture and damage. In the middle of the annulus there is the nucleus pulposus that's filled with interstitial fluid.

Throughout the day the fluid is compressed out of the nucleus and then reabsorbed when the body is resting. Fluid is also found in the annulus but is re-adsorbed and therefore most concentrated in the nucleus (Fig. 5). This incompressible fluid is mainly comprised of water and proteoglycans. The proteoglycans are mainly large sugar molecules allowing for the reduction of friction making the fluid slick and friction reducing.

Different types of injuries can cause structural problems to the intervertebral discs. A vertebral disc herniation creates a weak point in the membrane allowing the contents of the disc to spill outside from between the vertebrae. Even an increase in age can lower flexibility. Old age can prevent the natural re-adsorption of fluid into the annulus making an older person's posture shorter over time.

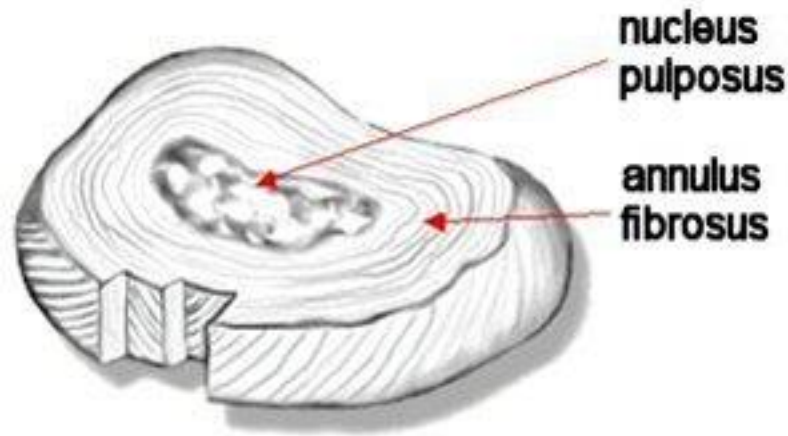


Figure 5: Nucleus Pulposus and Annulus Fibrosus.

1.3 Material properties of the Spine

Material properties governing the internal and external infrastructure of the human spine are crucial to the patient's overall health. Because of the various functions the spine plays, its flexibility and rigidity help the spine support and move the body while maintaining optimal posture. Healthy vertebral bone will display very different mechanical properties from that of osteoporotic bone. The same goes for the intervertebral disc where healthy cartilage will exhibit different compressive and mechanical properties than those properties of arthritic cartilage. The mechanical properties of specialized ligaments that line and support the spine are also crucial for the proper alignment of the spine. The elastic moduli of these ligaments are responsible for maintain posture for they act as a spring and cable system providing structural support for the spine.

There has been speculation that the cortical and trabecular regions have similar material properties despite a wide density differentiating each section of bone. Both

sections are composed of the same non-organic mineralized hydroxyapatite, however differences in densities can play a role when it comes to fracturing from non-compressive movements. The anatomy of the human spine is specialized for maximum compressive forces axially along the body of the vertebrae, evident by how a patient's rotational movement has been known to cause large stress concentrations as well as fractures. This can be a product of the traceable and cortical bone having different natural material properties. Experiments have shown that when a spine is upright and in its natural compressive states, the material properties of both trabecular and cortical bone are equivalent [1]. The material properties of vertebral bone at a macroscopic scale are dependent on the orientation and directional loading; therefore, making it anisotropic. Any type of mechanical testing where the spine is upright is difficult to do with most animal models because vertebrae are horizontally aligned as opposed to being vertically stacked. Human models are usually the most ideal.

Vertebral bone has also been seen to decrease in anisotropy with increased bone density and elastic moduli. It has been observed that elastic moduli in cortical and trabecular bone are relatively more or less the same when loaded axially. However, when the spine is transversely loaded or loaded at an off center angle (such as during spinal rotation) it becomes apparent that the moduli are in fact very different due to heightened stress levels within the bone (this has been known to lead to degeneration of the intervertebral discs [2, 3]). Vertebrae elastic modulus ranges between 10 and 20 GPA. Stark differences in elastic moduli between bones that are not loaded in its intended direction could cause them to fracture.

A series of ligaments and tendons are responsible for the stabilization and recoil of the spine. Because these tendons are not loading bearing they will have much lower elastic moduli than their bone and disc counterparts. These ligaments however will exhibit large amounts of strain in order to compensate and readjust for the curvatures of the spine. The stabilization tendons exhibit viscoelastic mechanical properties and are also subject to creep after loaded for extended periods of time. High strain over an extended period of time in cans also cause kyphosis.

For the SSL, putting extended amounts of stress in areas in front of the ligament can lead to over straining. Those patients who already have an abnormal kyphosis curvature will be shifting most of the weight in their thoracic cavity anteriorly anyways causing center of mass to move forward and downward [4]. This added weight would add creep motions that will eventually over strain the ligament only worsening kyphosis if left untreated. The ligaments found in the spine are also highly anisotropic. Due to fine fibers running parallel to the sagittal axis of the spine, the highest stress energy is found along this path. This can be proven by looking at the differences in ligament deformity in kyphosis compared to another spine misalignment disease called scoliosis. Scoliosis is the misalignment of the spine in the coronal plane. Upon observation of several cadavers exuding kyphosis and scoliosis curves it is seen that the ligaments in the scoliosis curve are thicker and more robust than those in the kyphosis cadaver. When comparing sagittal and coronal straining of spinal ligaments. The ligaments in the kyphosis cadaver were thinner and weakened by stress from a misaligned spine. Experiments also prove longitudinal have a higher modulus [5]. Ligaments also have

anisotropic properties are most present to the sagittal plane and will have little to no effect on other planes. When studying the midsagittal nature of kyphosis, it becomes unnecessary to study stress concentrations in multiple planes for that there is no significant strains for ligaments traveling in those orthogonal directions.

The intervertebral disc is well equipped to handle the constant and cyclical compressional loading needed by the spine. Cartilage alone is not capable of meeting the loading demands required to walk and lift. Intervertebral discs do not only show linear elastic properties because, due to its behavior when loading, it resembles viscoelastic properties.

The intervertebral disc must be able to absorb impact and promptly return to its original shape without losing significant energy. Studies have shown that constantly loading an intervertebral disc will increase its hysteresis curve representing a higher loss of elastic energy and over time diminishing the longevity of the disc (however, pre straining a vertebral disc before adding large loads have been proven to extend its longevity). The elastic moduli of intervertebral disc in the lower spine have shown the ability to increase in elastic moduli because of high load bearing duties.

Vertebral disc has also been proven to be quite durable. Even if the annulus fibrosis is lanced, the disc is able to maintain most of its mechanical strength after numerous tests. The osmotic pressure of the viscous disc fluid allowed for ample compression as long as it did not leak out of the joint capsule.

The viscoelastic properties in the intervertebral disc result from the elastic properties of the cartilage and the displacement of fluid from the nucleolus pulpous. The viscoelastic properties allow large loads to shift fluid within the disc. When loads are relieved, the fluid is allowed to recirculate the disc restoring original shape. This biphasic swelling property allows the spine to redirect and support external loads will still maintain the ability to be flexible. For reference, the modulus for nucleus pulpous is nearly 3 times higher than that seen in the annulus fibrosis with moduli at 100KPA and 30 KPA respectively [6].

1.4 Diagnosis, Symptoms and Pathology

Clinically good posture can be rated by looking at the C7 plumb line on the spine. On a cross-sectioned midsagittal radiographic image, the C7 plumb line is an imaginary line going straight down from the C7 vertebra. If the line passes straight through the sacrum, the patient is considered to maintain good posture and has a regular spinal curve. Anything grossly in front or behind the patient is considered miss-aligned.

A patient's thoracic kyphosis should normally be in the ranges of 10 degrees to 40 degrees (while lumbar lordosis is usually 40 degrees to 60 degrees. Anything less than 10 degrees, patients will exhibit a condition called sway back; the patient's spine will start leaning backwards and the C7 plumb will move posteriorly to the sacrum. If the curvature is over 40 degrees, patients will show a hunch back appearance called thoracic kyphosis (in the disease sense). The C7 plumb line will be anterior to the pelvis and in response the pelvic girdle will rotate forward in order to compensate for the off

set in weight. The abnormal rotation of the pelvis will cause strain on muscles of the posterior thigh. This over strain can lead to pain in the legs and back. Because the irregular curvature is not optimal for the body, muscles are in constant and futile struggle to return the spine to its regular shape.

The irregular curvature not only forces the pelvis to rotate but also forces the head downward. The downward angle of the head is measured as the brow, chin angle. The brow chin angle is the measure between the brow/chin and sagittal vertical axis (SVA) [7]. In good postured patients this angle is usually at or near 0° and so parallel to the C7 plumb line. Anything greater will trigger a reflex called the eye righting or the labyrinthine righting reflex. The inner ear detects non-horizontal posture when the head moves forward and will make the head readjust itself. Muscles in the neck and thoracic vertebrae will attempt to adjust by pulling the head back and upwards. Because this is not a natural body position, these muscles will begin to strain and fatigue causing major pain and discomfort to the patient [8]. If not treated this posture can become habitual. Even if the patient's thoracic posture is corrected the head can still lean backwards causing permanent stress on the vertebrae.

There have been attempts to reposition the head to naturally be in the upward forward position. By attaching weights to headgear, a person's head can be re-trained to return to a more natural curve. Currently these methods are only effective for people with extreme cervical lordosis. Patients on average saw about a 34% improvement in the cervical spine when using a 5-pound weight and a 31% improvement with using a 3-pound weight. People with cervical kyphosis also fared to do better than those with

cervical lordosis. It is not understood why there is a change in performance between those with cervical kyphosis and cervical lordosis.

The result of kyphosis does not stop at the muscle fatigue. Because the patient is unable to lift their heads up they are unable to communicate as easily as someone without kyphosis. With a head being down it becomes a struggle to make eye contact. Patients will force themselves to move upright which pains them in the process. Neck and leg muscles are not to be constantly contracted for extensive periods, so patients ultimately revert to their hunched positions and are sometimes embarrassed by not being able to make eye contact with a friend or loved one. Patients can also be put on a pain medication regiment, which can also decrease morale.

Kyphosis can cause several other symptoms that plague the body such as difficult breathing. The curve can decrease available volume in the thoracic cavity limiting one's lung capacity. Difficulty breathing makes these patients fatigued and short-winded making it difficult to carry out one's day. Irregular curvature can also cause herniation of the vertebral plate. A herniation can then put pressure on a nearby nerve causing pain and nerve damage to the patient. Thoracic kyphosis can also cause the dislodgement of an entire vertebra, which can cause extreme pain, little to no movement in the body and can account for added stress on the spine. The irregular curvature thoracic kyphosis creates also makes many patients feel self-conscious of their appearance. Thoracic kyphosis has several treatments that can in turn relieve a lot of these side effects.

In order to prescribe treatments one must understand the mechanisms by which kyphosis comes into existence. Thoracic kyphosis is split between its causes in children and in adults. In children kyphosis is termed Scheuermann's disease. Scheuermann's disease is seen as the irregular growth of a child's vertebral column at the onset of puberty. Different vertebrae will grow at abnormal rates forcing them to become wedge shaped. It is not sure what cause the vertebrae to start growing independent of each other but there are theories that link the disease to being hereditary. Children are also at risk for kyphosis from poor posture. It causes damage by impacting the spine's developing morphology.

Adult patients get thoracic kyphosis when they have degenerations in their intervertebral discs. The cartilage that originally provided support will no longer be able to absorb shocks. Degenerative disc diseases (DDD) is the degradation and eventual disappearance of the shock absorbing vertebral discs. DDD and its symptoms will inevitably cause other spinal conditions. Lack of motion in the spine will increase its rigidity and can cause rubbing of the vertebrae causing discomfort for the patient. This rubbing can cause damage to nerves due to increase in pressure and from resulting bone spurs from the vertebral body.

Normally the fluid filled vertebral disc would give the patient an upright posture. Disc degeneration alters shape making the disc compress irregularly. Hernias can also create similar effects. With less disc fluid now underneath the vertebrae, the disc has a lesser ability to keep the vertebrae upright after being compressed leading to increasing degrees of kyphosis.

Spondylolisthesis is a result of DDD. Normally the intervertebral disc keeps the vertebrae aligned with one under the other. However, Spondylolisthesis occurs when vertebrae slip past one another. A vertebra (usually the L5) may slip too far forward allowing the spine to destabilize resulting in spinal mal-alignment and the onset of pain. A single vertebra has the potential to offset the rest of the spine, and the lower the lower the vertebrae is the more of a catastrophic effect it may have on the rest of the spine.

Osteoporosis is another disease that can potentially result in thoracic kyphosis. Because this disease is the loss of bone density, pressure can cause fractures easily. The weight of the body will crush the vertebrae. Weight from the patient's body will cause the vertebrae to wedge. The wedging will cause the entire spinal column to produce the aforementioned hunch back appearance. This cause of kyphosis is highly associated with women post menopause. Because inefficient amounts of hormone are no longer being secreted the density of bone suffers.

Even some cancers can cause thoracic kyphosis as a side effect. Large tumors can misalign vertebrae and cause them start curving abnormally. Also various treatments and the cancer itself can have effects on bone porosity making it easily susceptible to breaking. It is possible that patients in the hospital having extended periods of stay are susceptible to bed sores. Because the spine curvature is abnormal, some patients may experience more severe bedsores than others because of added pressures where the curvature is over extended. The shape of the bed could further degenerate the patient's posture the longer they sit, that is why it is still important to remain active. The inability

to move is a major part of irregular kyphosis development. So the development of kyphosis is not only effected by disease and genetics, but posture, activity and your environment can play a very significant role.

1.5 Treatment of Thoracic Kyphosis

Treatment of kyphosis is improving to better comfort the patient while still providing the most effective treatment. Back braces have been commonly used to support the spine without the need for surgery. These types of braces surround the entire torso allowing for equal weight distribution. The rigidity and back support helps the wearer maintain good posture. Back braces and physical therapy can be an asset for patients who have more mild cases of kyphosis. Pain medication, spinal cord stimulation and spinal blocks are also options that can help patients relieve pain during treatment of the spine.

The brace however is more pertinent for mild cases of thoracic kyphosis. Braces cannot stop the progression of thoracic kyphosis because it does not have the strength or control over the spine in order to cause significant change to the spine.

Spinal fusion is another option for spinal correction. Spinal fusion as the name suggests is the growth of 2 vertebrae into a single rigid bone to combat the spinal misalignment. The procedure first involves removing sections of bone either at the facet joints or the vertebral body of two adjacent vertebrae. Upon removal of the bone, a surgeon will then cut away at the damaged intervertebral disc until it is removed completely. Finally, the surgeon will then inset a bone graft in between the vertebrae

forming a rigid bridge between them. The bone graft can be harvested from another part of the patient's body during the surgery or from a bone bank before the surgery. Once bone is inserted into the void bone from both vertebrae will grow together forming a fused section of vertebrae. By building fused vertebrae, the flexible disc that can cause mal-alignment is removed in its entirety giving the surgeon power to reshape the spine in its correct alignment. However, with increased alignment comes a loss in flexibility in whichever region the vertebrae have been fused.

External hardware and implants have been developed as a means of repairing spine curvature in more extreme cases. Surgeons can plastically deform metallic rods to create perfected curvature of a patient's spine. Rods and plates can either be placed anteriorly or posteriorly to the spine. Different levels of hardware are also available to the patient. Sometimes the entire spine doesn't need to be fixated with implants, but can sometimes be better off with just certain portion with hardware on it. If a surgeon only has part of the spine connected to an implant they must consider possibilities of high levels of stress concentration between spinal segments with and without implants. If the moduli of the bone and metal do not match, high levels of stress resulting from a difference in material can cause the bone to potentially crack [9]. Even the holes drilled into the vertebra for the placement of the implants are a call for concern. Screw holes that are drilled too close in proximity can result in overlapping stress, which can also crack bone.

Re-alignment of the spine during severe cases of thoracic kyphosis have proven to be a truly invasive task. The correctional surgery can take upwards of nine hours to

complete and is followed by months of rehabilitation as the patient acclimates to a new posture. Thoracic kyphosis realignment can be achieved by using metal hardware to treat bone misalignment [10].

Hardware is divided into three major modules. The anchoring members or bone screws are designed to interface between bone and metal. These screws are threaded to allow for maximum surface area between bone and metal. After bone screws have been set and are aligned, longitudinal elements are attached. Longitudinal elements are essentially two metallic rods parallel to each other running the length of the spine. The rods are plastically deformed and shaped by the surgeon to match closely to what spine angle the patient requires. Implants can be an alloy largely consisting of titanium, stainless steel, or cobalt chromium.

Transfixation elements are also included to help distribute weight within the longitudinal elements. By bridging stresses between both elements, load is dispersed and evenly distributed between both rods providing for the longevity of the implant throughout its service life.

When it comes to researching kyphosis animals are not ideal. Most animals on earth have a spinal vertical axis that is perpendicular to the force of gravity, so the loads and strains carried from upright human vertebrae are wildly different from quadruped pig or rat. Currently there is extensive research going into how the spine curvature can affect other systems and how it can better be treated without actually doing trials on patients first. Computational modeling techniques is a powerful tool used to look at the

mechanical properties of bone in the vertebral column, how different increments of curvature can control stress concentration in bone, and how back muscles can react to electrical stimulation to lessen the fatigue effects brought on by kyphosis. These computational simulators are very powerful in measuring and calculating mechanical properties like stress, which can become a very daunting task. Radiographic images can be converted into highly detailed 3D renderings. With proper coding, an engineer (or surgeon) can use this data to simulate tensile forces needed to correct patient's back, or how their back will react to treatment.

1.6 Surgery and Complications

Proper placement of bone screws is crucial for proper alignment. Anchoring members are implanted through the pedicles of the vertebrae for maximum support. Guide holes are created before anchoring member is introduced. These guide holes are approximately 60% of the final diameter of the actual anchoring member, while the screw itself is 80% of the pedicle diameter. The depth of the anchor member can reach nearly 75% of the entire length of the vertebrae. Cancellous bone must be contained on all ends of the member to promote proper attachment so proper drilling depth is must be obtained to achieve this [11].

Numerous failure modes are present both during the installment of mechanical hardware operation and post-operative. With any type of surgery, infection is always a risk. Large portions of the back are exposed to the environment, which always runs the risk of foreign entities entering the body. Infection however was less than 2% of patients

in a study reporting the effectiveness of thoracic kyphosis, with most cases resolved through drainage.

There were also issues when it came to misaligned screws. Screws could be too superior or inferior to the target point on the vertebral column and could potentially be called back in for corrective surgery. Screws can also lose interface with the bone by becoming loose. Loosened screws could be the result of the angular deformity itself. High stress exhibited from attaching the screws to rod could cause screw detachment when utilizing vertebrae-to-rod procedures. The opposing forces of kyphosis can resist the rigidity of the longitudinal rod forcing bone screw to pull from their holes.

There is also the potential danger of the longitudinal member to fail as well. Varying levels of hyper kyphosis (kyphosis deformity over 40°) heightens the chance of rod breakage and failure in the implant. Stress concentrations will eventually propagate outside of the bone screws and will have the opportunity to form stress fractures in the metal. The metal used in the rod (whether it is titanium, steel or cobalt chromium) is plastically deformed to fit the patient's aligned spine contour. Plastically deforming a metal will in most cases strengthen the metal. However constant bending of metal will mix grain boundaries making the metal susceptible to failure [12].

Pedicle fracture is also a formidable concern. Fractures can increase the diameter of screw holes, which can invite loosening of the bone screw. This usually happens intraoperative and may require the full removal of screw. This has not been a major issue for older patients. However younger patients with softer more immature

cortical bone are at a higher risk fracture making this procedure potentially more dangerous for children.

The complexity of diseases makes mid sagittal curvature misalignments an ailment solved by many different solutions. Because even the very nature of juvenile versions of kyphosis is unknown clinicians and engineers are given the opportunity to imagine inventive solutions. Everything from the weighted headgear [13] to complex computer models can be used to help treat the disease. And solving these types of problems will only chain effect into other areas of the disease. Because kyphosis is an amalgamation of other diseases I.e. osteoporosis, cancer, etc., understanding kyphosis can be another key to our full understanding of some of these other diseases. Even other types of spinal misalignments like scoliosis may too benefit from the advances in treatment and diagnoses of kyphosis. Large technologic advancements like spinal sensors can be created to detect alteration in a patient's spine giving a surgeon real time data on its progression. The possibilities and opportunities are near endless when it comes to the solutions needed to heal mid-sagittal miss-alignments.

1.7 Remaining Problems of Interest Dealing with Thoracic Kyphosis and Hypothesis.

Mechanical fixation has proven to be the most effective method in spinal realignment for patients exhibiting thoracic kyphosis. Although this type of realignment has been praised as the golden standard for disease treatment, its performance can sometimes suffer with the most extreme case being rod breakage [14]. Most issues plaguing the implant occur postoperative. After several months after surgery, some

patients will start to display fractures in their cervical region do to what is called forward head posture (FHS) [15]. FHS is an irregular downward posturing of the cervical spine where the cervical lordosis is near zero (Fig. 6). It is unknown why patients start to exhibit this deformation after surgery. Many surgeons speculate that FHS is habitual, meaning that the patient's head will try to return to its original position because that is what the brain has recognized as normal. However other surgeons speculate that FHS may have some role in establishing biomechanical stability by reliving itself from proximal junctional kyphosis (PJK) (also proximal junctional failure). PJK occurs when two spinal regions of contrasting stiffness form a stress concentration at their interface [16]. This can hurt the body, so the body will naturally try to correct itself. So at the interface between the flexible cervical spine and the rigid thoracic spine a major stress concentration will build between the C7 and T1 vertebrae and FHS may just be the body's natural response.

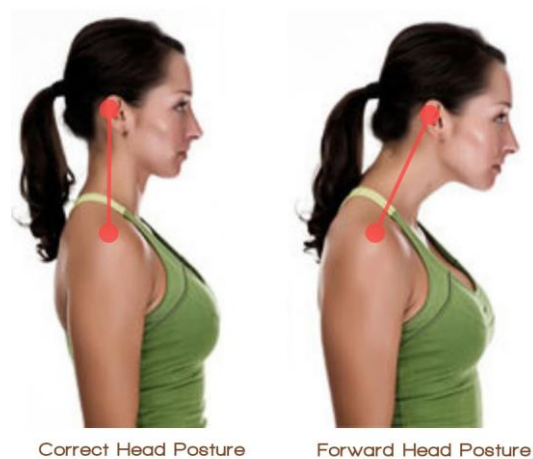


Figure 6: Forward Head Posture (FHP)

Surgeons are incapable of measuring forces acting on the spine during surgery and are therefore unable to calculate and analyze stress concentrations. Because of the rather complex geometry of the spine it is difficult to calculate stress concentrations. Locating areas of high stress in the spine can give insight as to why particular portions of the spine can be over stressed and how different implant designs can be utilized to lessen the impact in these regions.

FHP can also be a result of how the surgeon sets the spine. Getting the spine perfectly aligned is a challenge in itself, so it is expected that the surgeon can over correct or under correct the spinal curvature. By over correcting the spine the surgeon pushes the spine slightly past the target angle. Similarly, by under correcting the spine, the spine angle is slightly below or target angle.

Phenomena like fractures in the cervical vertebra and FHS can better be explained if one had better representation as to what forces are at play (internal and external), the material properties and stresses interacting inside the spine. Surgeons are unable to directly calculate stress with tools they currently have during the course of a surgery and are unaware of what types of forces they are applying to patient during surgery; and naturally do not know what forces are present.

The objective of this study is to create detailed stress profiles in the spine that are controlled by internal physical parameters. Parameters will include the patient's preoperative curvature, material properties of the spine/implant, the mass of the patient's head and whether the surgeon under corrects, over under corrects, or gets the

spine at its exact target angle. By creating these profiles, an explanation for the causes of FHS can better reveal itself.

We hypothesize that a forward head posture post operation is habitual in nature when the thoracic spine is over corrected or under corrected because increased stress concentrations from proximal junction kyphosis at and near the C-7 makes a head forward positioning highly unfavorable.

The use of computational models will provide theoretical values of stresses and enacted on vertebral geometry [17]. Computer renderings of the skull, cervical spine, thoracic spine, and lumbar spine will comprise the computational models. This finite element modeling will be used to calculate the complex nature of the spine.

CHAPTER 2

CRITICAL ANALYSIS AND EXPERIMENTATION WITH FEA.

2.1 Introducing Finite Element Analysis to Solving Complex Computational Models

Finite Element Analysis (FEA) solves physics related problems by solving governing partial differential equations (PDEs) using numerical methods through domain discretization, selection of proper types of elements for field quantity interpolation, algebraic linearization, and finding of approximate solutions to the PDEs. Finite element analysis can be used to solve complex problems including heat transfer, fluid flow, electromagnetism, and structural mechanics, among others.

FEA will be the most appropriate method for analyzing stresses found in a patient's spine. The virtual nature of the model's geometry gives the user a technically accurate rendition of the spine without running physical experiments. Different aspects of the model can be parameterized to show a large combination of different properties found in patients including preoperative spinal angle, material properties of the vertebrae and the mass of the skull. By changing material properties of the spine, the experimenter can represent different disease states inside patients like Osteoporosis by altering the vertebrae's density.

The stresses resulting from the deformation of the spine is not something easily quantified and visualized without using theoretical calculations from an FEA model. Stress is a calculated value and is traditionally done with the use of strain gauges. FEA is

more feasible because the amount of strain gauges needed to develop a workable stress profile won't be as detailed as a computerized model.

FEA models can be rendered at very high detail, but if only one aspect of the model needs to be inspected simplified models can be used. The geometry used in FEA modeling can exist in all 3 dimensions. Most structural analysis modeling involves 3D geometry, however there are still many cases where a 2D rendering will suffice. For example, if an analysis is being performed where the user is only interested in stress in one plane of the object, a 2D rendering will be proficient. By using a 2D model, the geometry is greatly simplified and calculation time is significantly reduced. In this study a simplified 2D rendering of the human spine will be used to study and visualize the stresses in the spine during and post-surgery. FEA could be performed on 3D spinal model, however since kyphosis is isolated to the mid sagittal plane, the experiment only calls for a 2D model. Future studies that want to study the structural stress on patients with both kyphosis and scoliosis, a 3D rendering would be advised. The final construction of a 2D spine will be further simplified to isolate the structural points of interest for this experiment.

FEA modeling software COMSOL 5.2a update 1 will be used to run this simulation. The geometry of the spine will be programmed within COMSOL's proprietary geometry creator. The physics and boundary conditions will all be set and analyzed through the structural mechanics physics module.

2.2 The Simulation

2.2.1 Construction of Spinal Geometry and Spine Curvature

By obtaining one 2D slice of the spine, the structures in question are isolated and can allow for more in depth studies. By excluding the rest of the 3D geometry one does run the risk of not recognizing potential trends that occur in other sections of the spine. However, by keeping important detailing factors regarding shape, physics and material properties in mind a simplified 2D model can be conceived that is thoroughly complex in its own right.

COMSOL's proprietary geometry package allows for parameterization of models geometric features. The spine rendered in COMSOL has the ability to have its dimensions altered internally making the spine's size height and initial curvature truly variable. COMSOL does have the ability to import geometries from third party Computer Aided Design (CAD) software including Solid works and CREO. Originally the model was going to be constructed in one of the aforementioned software, however that would counteract COMSOL's ability parameterize key aspects for the experiment for this and future experiments.

Several iterations of the spine were created until a suitable version was constructed. Amongst all versions the shapes and internal construction of the spine remained relatively the same. Each feature of the geometry is divided into individual "parts". These parts are built once and can be retrieved an infinite number of times. The parts go as follows; skull, cervical, thoracic, lumbar and sacrum. Parts represent each

vertebral bone type and skull. The parts are essentially stacked on top of one another until a recognizable profile of the spine is generated.

In construction of individual vertebrae types, a planar outline of each vertebra was made. The original vertebrae included all the major anatomical components of the cervical, thoracic, and lumbar vertebra. The vertebra was made as a combination of at least two major geometrical parts. The body of the vertebra is made of the rectangular block feature in the geometry ribbon in COMSOL. The simulator's user can alter the length and height of the block. As previously mentioned, this gives the user the complete ability to completely describe the model to a wide range of patient dimensions. The posterior of the vertebrae's body is constructed with a polygon tool that is able to create free-formed shapes. The geometry formed from the polygon represents the pedicles, facets, lamina, spinal and transverse processes of the spine. However, the current iterations of the model would do away with this rear facing geometry. The physiological purpose of this section of the spine serves functions of protecting the spinal cord or preventing the spine from over exerting. For the simulation these remaining portions of the vertebrae would only serve aesthetic purposes on the virtual spine. With a lack of functional purpose relevant to any variables or characteristics of the simulation it was voted to neglect these remnants in order for the simulation to run faster.

The sacrum is also constructed using the polygon tool. The shape of the sacrum does not have any significant mechanical implications on the spine, therefore the dimensions of the sacrum is not a variable parameter. The sacrum is a base point where

the rest of the spines geometry references to and ultimately governs the shape of the spine. The base of the sacrum is located at Cartesian coordinate (0,0).

Lastly the skull is constructed in the similar method as the sacrum. All these segments of bone will all appropriately be given the similar material and mechanical properties when fully assembled.

Once all parts are constructed, it is time to assemble. Between each part there will be the intervertebral disc. Rectangular blocks that overlap the boundary of the adjacent vertebrae represent the intervertebral discs in this model. The purpose of this was to have the shape of the disc change shape when the user programs a new misaligned curvature value. Programming the dimensions of the vertebral disc to change shape in relation to the rotational movement of a vertebrae is needed to test structural stress at various angles. In order to solve this the vertebral disc block geometry simply stands underneath the two vertebrae covering almost all the area in between them. When the rotational orientation of the vertebrae changes it will cover more or less of the vertebral disc block's geometry leaving behind a shape that represents the displacement of the vertebral disc when the disc is deformed.

The curvature pre-set programmed by the user utilizes the rotational feature native to each vertebral part type. Each vertebra has an axis of rotation located at the bottom left corner of the vertebral body. At this corner, each vertebra can be rotate incrementally to a specified degree. In order to make all the vertebrae rotate just enough to add up to its respected angle, the number of vertebrae located in that section

of spine must divide that angle. In the thoracic spine, 12 will divide the angle of curvature. Every time a vertebra rotates by its divided value, it will displace all the vertebrae above it the same number of degrees, and those vertebrae above have also been rotated the same number of degrees. When all of the vertebrae add up the number of degrees they have been rotated, the accumulated values will equate to the desired angle for that section of spine. The column of intervertebral disc that sits underneath the vertebrae mimic this motion so that most of their area still remains in the region in between each of the vertebrae.

The sacral slope also has a similar effect. When the angle of the sacrum slope changes it rotates around its body at the center of the coordinate plane. This movement will in turn change the direction the spine leans as well further allowing the user to replicate thoracic kyphosis scenarios. Using this method of rotating the spine allows for replication of thoracic kyphosis as well as changes in curves for the other sections of the spinal column.

Geometry of the hardware implants are also closely associates with the parameters of the spinal rotation. The thoracic vertebral part is modified to allow the inclusion of bone screws. A single rectangular profile is subtracted from the posterior end of vertebral body. A second rectangular element is added and the two are integrated into one object. The edge of the bone screw is filleted in order for better attachment between the screws and longitudinal elements.

It was decided not to add threading to the screws because of issues with meshing the model. Because the bone screw interface is not of primary concern some lack of data from this portion of the model could be condoned. The sharp edges of the threading would create a naturally very highly concentrated mesh requiring higher computational resources. The longitudinal element is an outline of the thoracic spine when its angle is at 40 degrees. In surgery this represents the shape surgeons will plastically deform a longitudinal element to before attaching the spine. After outlining the profile of the longitudinal element with a polygon tool the, element is left in that fixed position in 2D space. When the spine is set to a new curvature, the longitudinal element will not move. The screw at T12 is the only segment of the longitudinal element that is attached to spine the throughout the whole experiment.

The spine and all the bone screws inside of it are set as a single part but have distinguished regions that contain different property values. The spine itself is a union but the spine and longitudinal element are assemblies that must be adhered together. By making the two separate entities the relationship between two foreign objects will be amplified. The contacting physics between the rod and the spine will be better studied and be able to better represent what happens outside of a non-virtual environment.

2.2.1 Movement of the Spine

The spine model will start in its diseased kyphosis curvature before it is set to move (Fig. 7). Without any forces acting on the spine, it will remain a stationary static

object. The curvatures of the spine are set at the initial values listed in the table 1. All curves are clinically healthy except for the thoracic region.

Table 1: Starting Angles and Slopes for Spinal curves

Spine Region	Initial Angle /Slope (°)
Cervical	20
Thoracic	60
Lumbar	35
Sacral slope	20

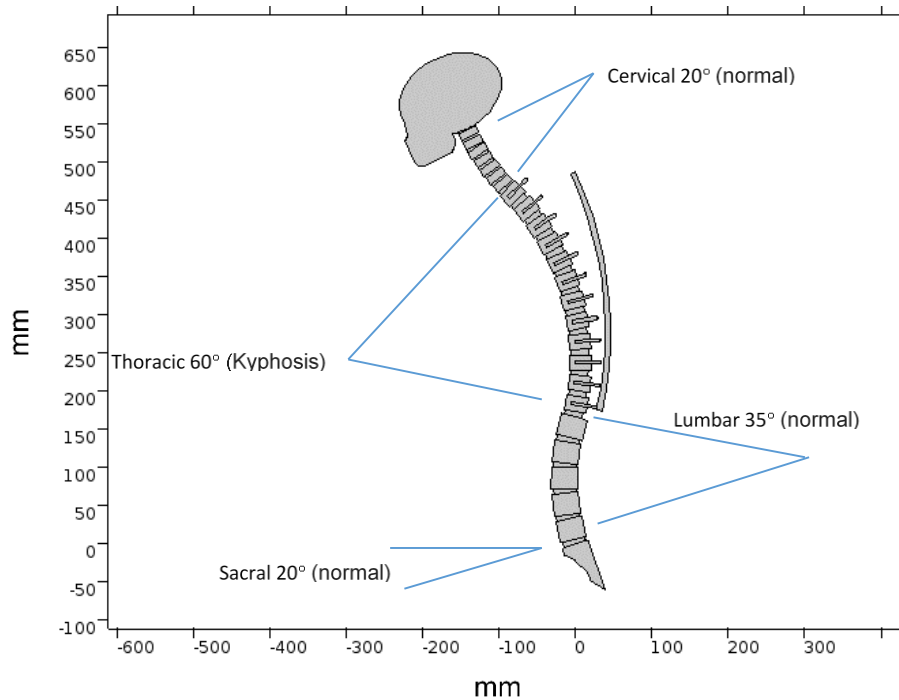


Figure 7: Geometry of Spine Model Prior to Simulation Execution.

Each region is positioned to a normal angle except for the thoracic spine, which will change shape from 60° to 40°

The longitudinal element is stationary and permanently connected to the spine at the T12 vertebrae. The longitudinal element is designed to be at an exact 40° angle

(Fig. 8). The spine will incrementally move along a predefined path straightening the spine (Fig. 9).

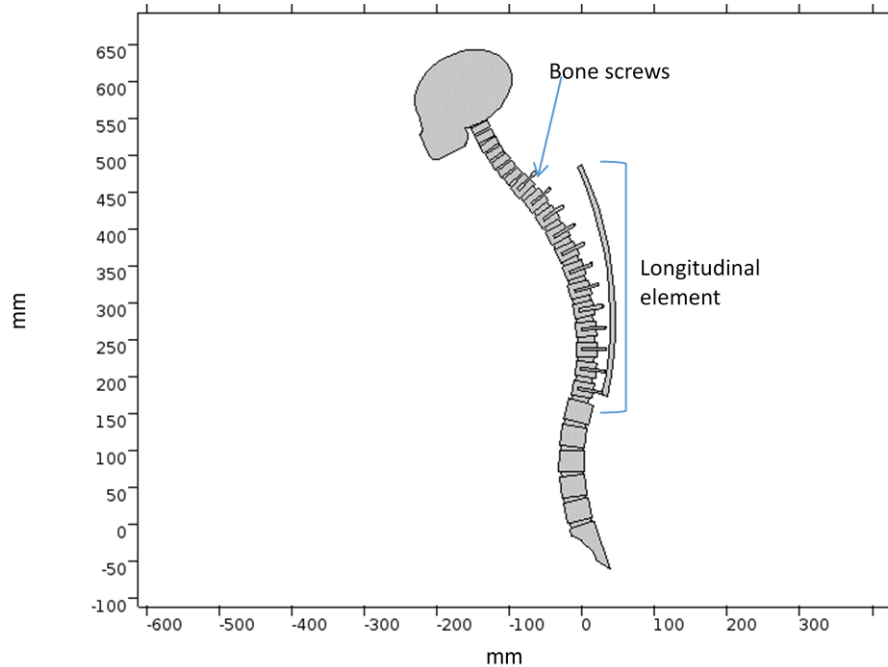


Figure 8: Hardware

When the thoracic spine migrates from 60° to 40°, the bone screws in each of the 12 thoracic vertebrae will interlock with the longitudinal element, which will support and keep the spine upright.

The bone screws will come in contact with the longitudinal element and will stick to it (Fig. 10). Once stuck together the patient's spine has been straightened out (Fig. 11). The longitudinal element will now need to bend and deform in relation to the spine (Fig. 12).

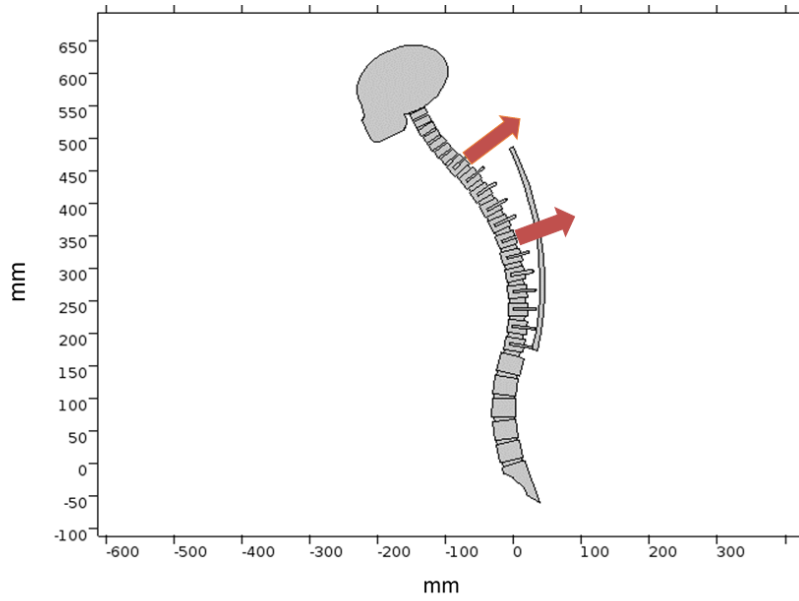


Figure 9: Spine Start

A prescribed displacement is the driving force that moves that elongates the spine from a kyphosis 60° to that of 40° , and is also the model analog to the surgeon moving the spine back in place. The prescribed displacements move the bone screw on the

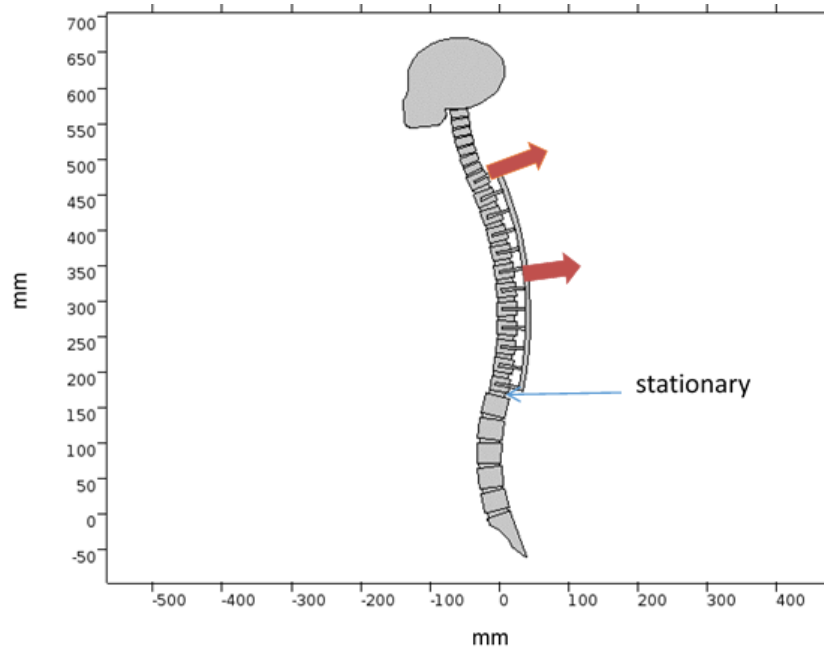


Figure 10: Spine Movement

The spine will continue to migrate until the prescribed displacement makes it even with the longitudinal element. Notice that the longitudinal element is fixed at the T12 bone screw, so the entire spine pivots at this area.

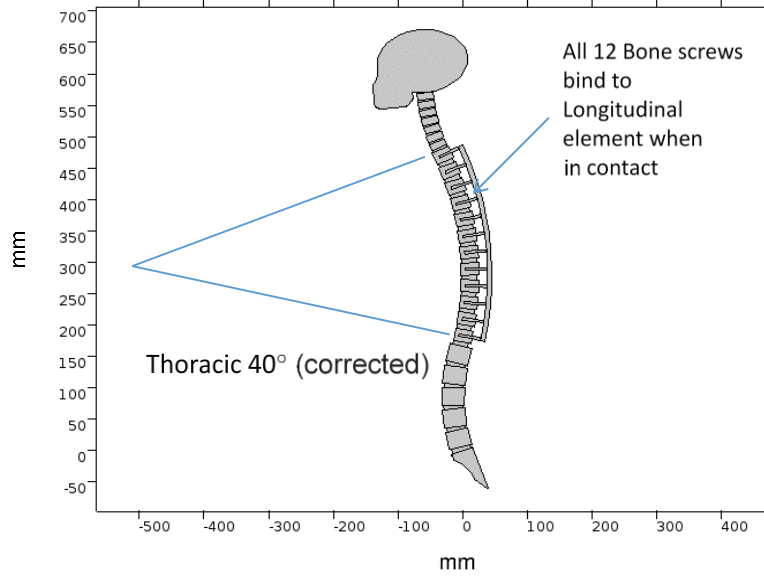


Figure 11: Spine Attachment

The thoracic spine has now migrated from 60° to 40° . The bone screws interlock with the longitudinal element, so now the longitudinal element will move and bend however the spine changes next.

After sticking to the longitudinal element, the spine will move slightly passed where the longitudinal element is, moving it backwards. This represents the surgeon over correcting the spine. The spine will then travel in the reverse direction. By moving the spine backwards, this simulate the surgeon under correcting the spine. This will provide unique stress profiles, and can give better insight as to fracturing inside the cervical region.

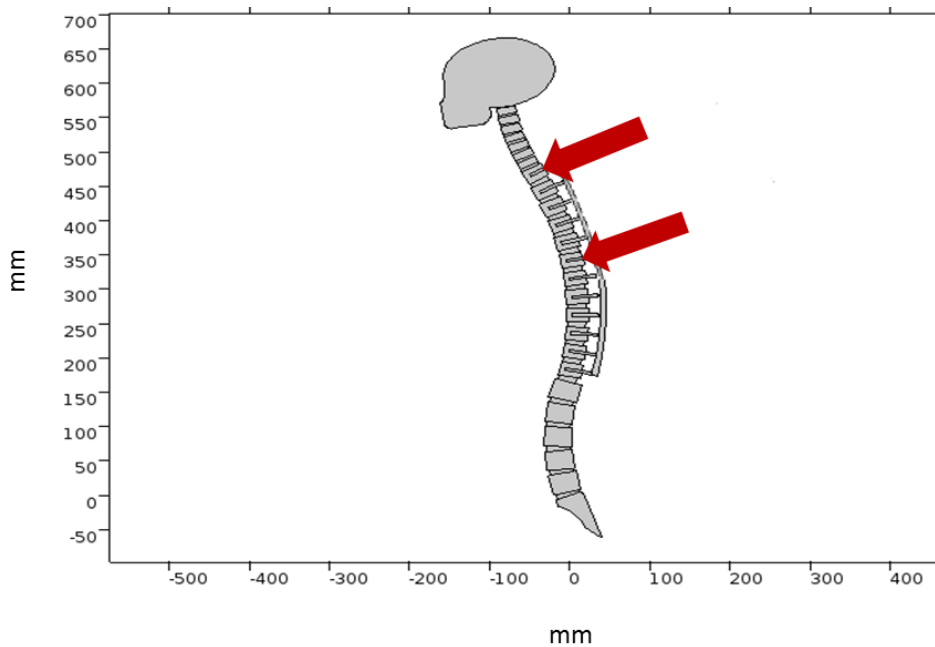


Figure 12: Spine Reverse

The thoracic spine is now migrating in the reverse direction pulling the longitudinal element along with it. This will put stress on the longitudinal element as well as on the vertebrae. In the simulation the spine will not move all the way back to 60°

2.2.2 Triangle function

The prescribed displacement is the driving force that incrementally moves the spine. The spine's prescribed displacement is programmed to move it all the way upward (connecting to the longitudinal element and then back down again, but with the longitudinal element attached to the bone screws. The x, y vectors of the prescribed displacement are multiplied by Z, which marginally increases from 0 to 1 and then back down again from 1 to 0. So at a Z value of 1, the prescribed displacement has moved its entire length or 1 times the entire length x and y whereas at a Z value of .5, the prescribed displacement has moves half its total length or .5 times the entire length x and y.

The value of Z is dependent on another variable, par. The value of par is defined by what is called a parametric sweep. In COMSOL a parametric sweep allows a variable to increase automatically from a starting value to an ending value in increments set by the user. In this case the range would start at 0 and progress to 2 in increments of 1. When par and Z are plotted together, a triangular function is formed (Fig. 13).

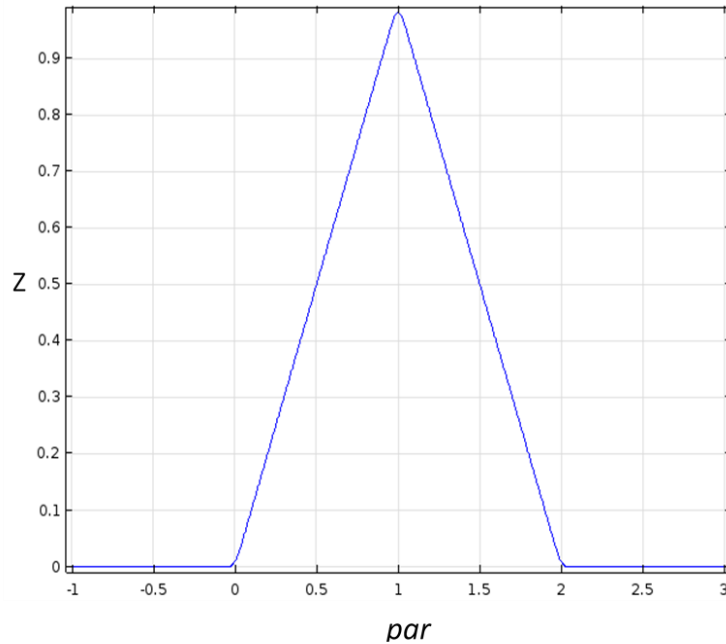


Figure 13: The Triangle Function

Formed by plotting *par* against *Z*. *Z* rises and falls in value between 0 and 1 and is multiplied to the prescribed displacement. *Par* is a parametric sweep that increases from 0 to 2 in increments of .5 and dictates the value of *Z*.

The simulation calculates stresses at every *par* step, so at *par* (0) for example COMSOL generates stress values for the model when thoracic kyphosis is at 60°. Identically at *par* (1.0) thoracic kyphosis is at 40°. The *par* value dictates *Z* which sets how far back or forward the spine is angled by manipulating its described displacement. So each *p*value will represent the angle of the thoracic spine at each step (Fig. 14&15).

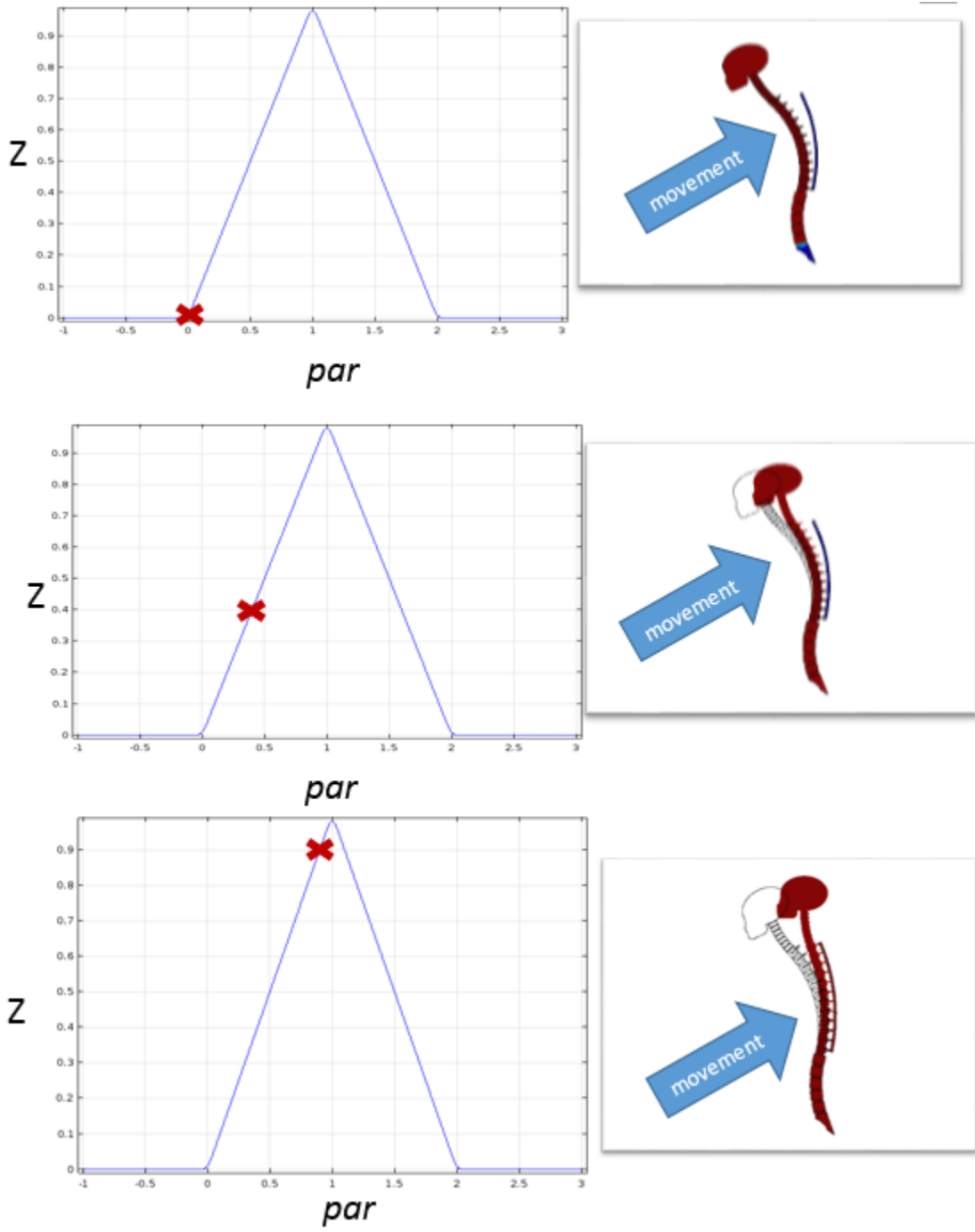


Figure 14: Upward Movement of Triangle Function

$par(0)$, $par(0.5)$ and $par(0.94)$ are shown in their respective order. The spine gradually moves from its hunched position (thoracic kyphosis of 60°) to its corrected position (thoracic kyphosis of 40°). At $par(0.94)$ the spine is firmly connected to the

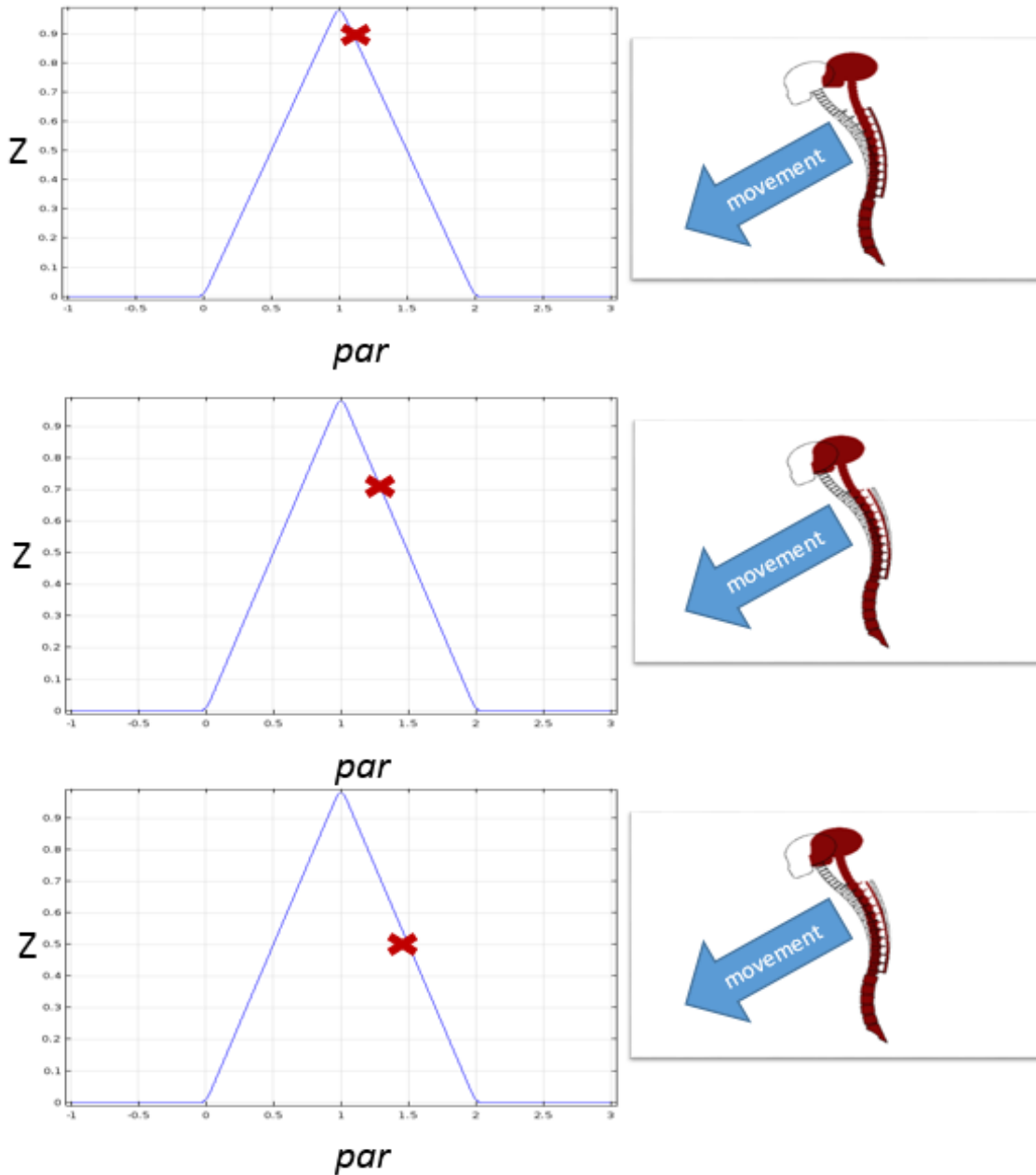


Figure 15: Downward Movement of Triangle Function

$par(1.06)$, $par(1.24)$ and $par(1.5)$ are shown in their respective orders. On the other side of the triangle function, the spine is now migrating in the reverse direction and tugs on the longitudinal element.

2.2.3 Over Correction, Under Correction, and Exact Angle

The prescribed displacement of the spine ends at a point several centimeters behind where the longitudinal element is actually located. So when the spine reaches its full displacement or par (1.0) it is actually pushing on the longitudinal element. The spine has already connected to the spine at par (.94) so it will now move however the spine migrates. This is the position that surgeons strive for during a procedure. This will be the “perfectly aligned” case (Fig. 16). Par (1.0) will represent the “over corrected” case because the spine has moved slightly passed its normal positioning (Fig. 17).

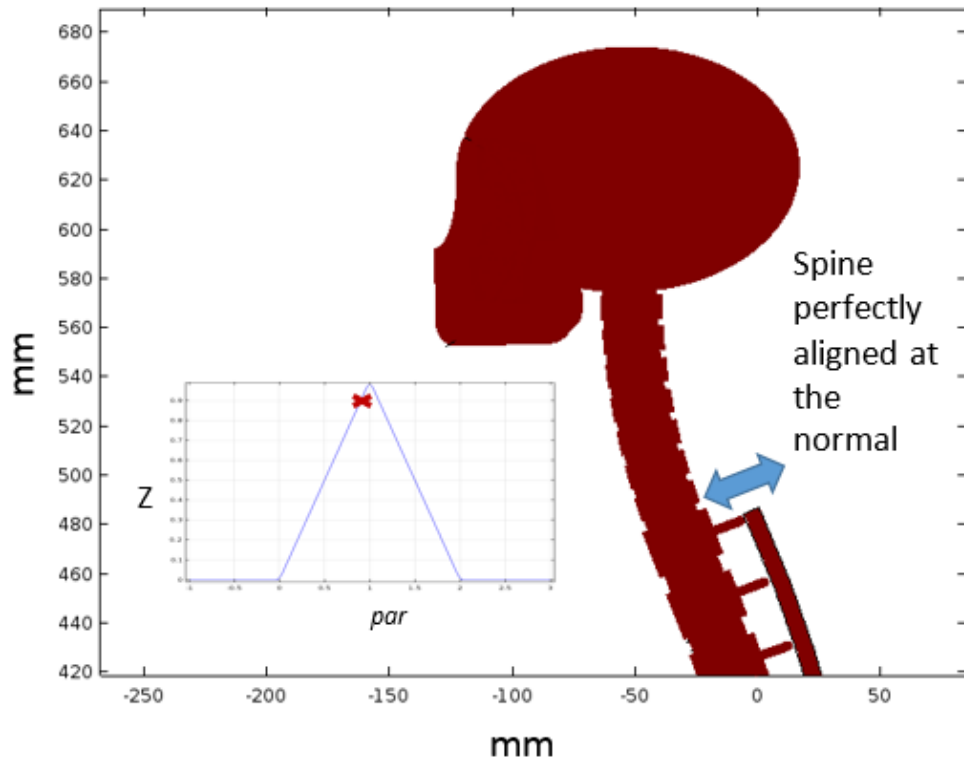


Figure 16: Exact Alignment

At *par* (.94) the spine is perfectly aligned and connected with the longitudinal element. So there is no pull or push on it. This exactly aligned positioning is the ideal alignment surgeons strive for in a procedure

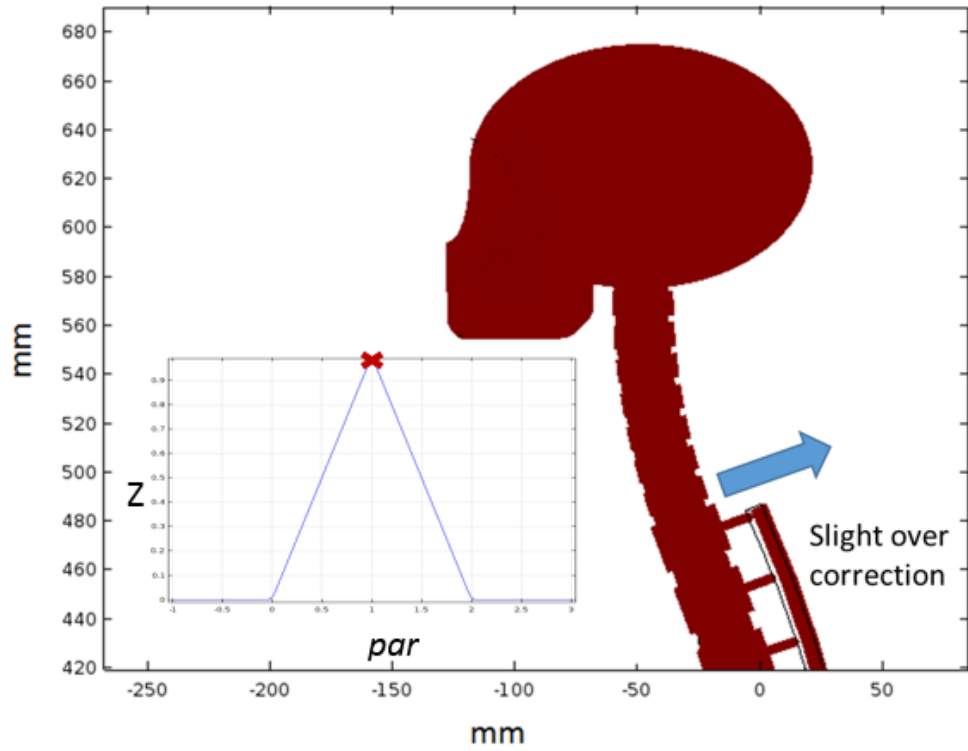


Figure 17: Over Correction

At $par (1.0)$ the thoracic spine is slightly over the 40° mark making the spine over corrected. Because the full pre-described displacement is behind the longitudinal element the spine will have to move the longitudinal element backward.

When the spine makes its decent back down the triangle function, the spine will bend the longitudinal element ventrally creating the “under corrected” scenario . The par (1.24) will represent the under corrected positioning (Fig. 18).

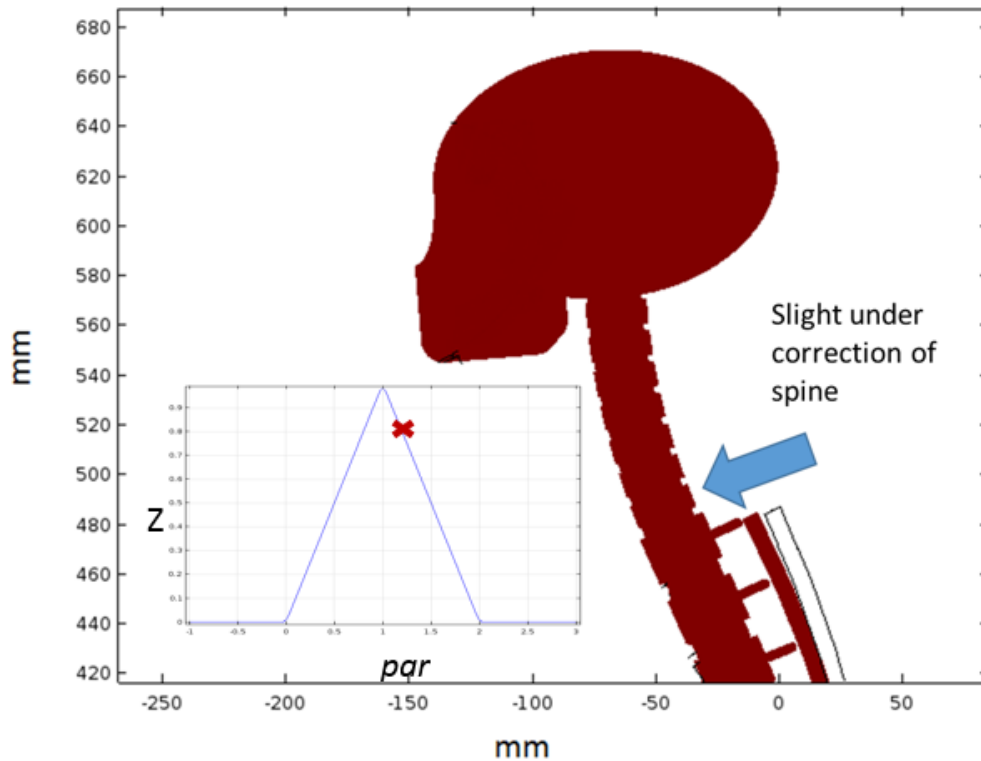


Figure 18: Under Correction

At par (1.24) the spine is descending back down the triangle function. The prescribed displacement is therefore in front of the longitudinal elements normal position, causing it to lean forward creating the under-corrected position.

2.2.4 Cervical Spine parametric sweep

Layered on top of the thoracic spines movement along the triangle function is another parametric sweep function that controls the different angles of the cervical spine. The purpose of this experiment is to determine how stresses in the cervical spine are influenced not only by the over correction, under correction and perfect alignment of the thoracic spine but also the angle that the cervical spine assumes when at each of the three scenarios.

In this experiment, the cervical spines angle will be a parametric sweep starting at 0° and ending at 50° and increases by 10° increments (Fig. 19). The sweep will be running at each of the three correction scenarios and the stress profiles at each angle will be recorded making for a total of 18 individual stress profiles (six different cervical angles multiplied by three different correction scenarios)(Fig. 20).

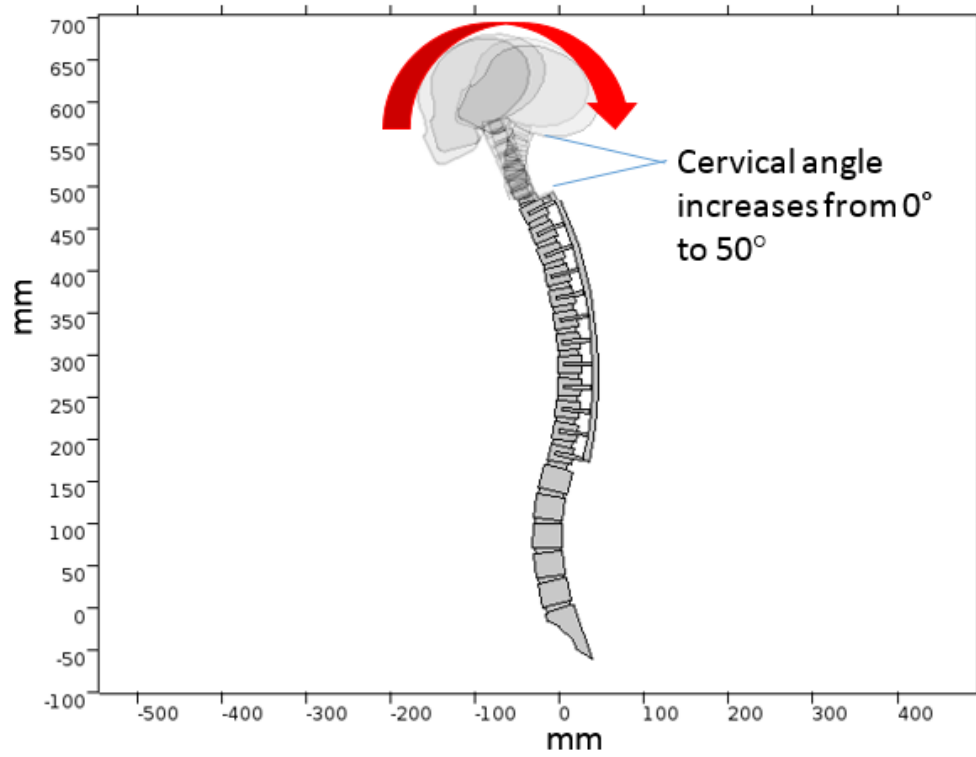


Figure 19: Cervical Spine Movement

The cervical spine will need to perform a parametric sweep ranging from 0° to 50° in increments of 10°

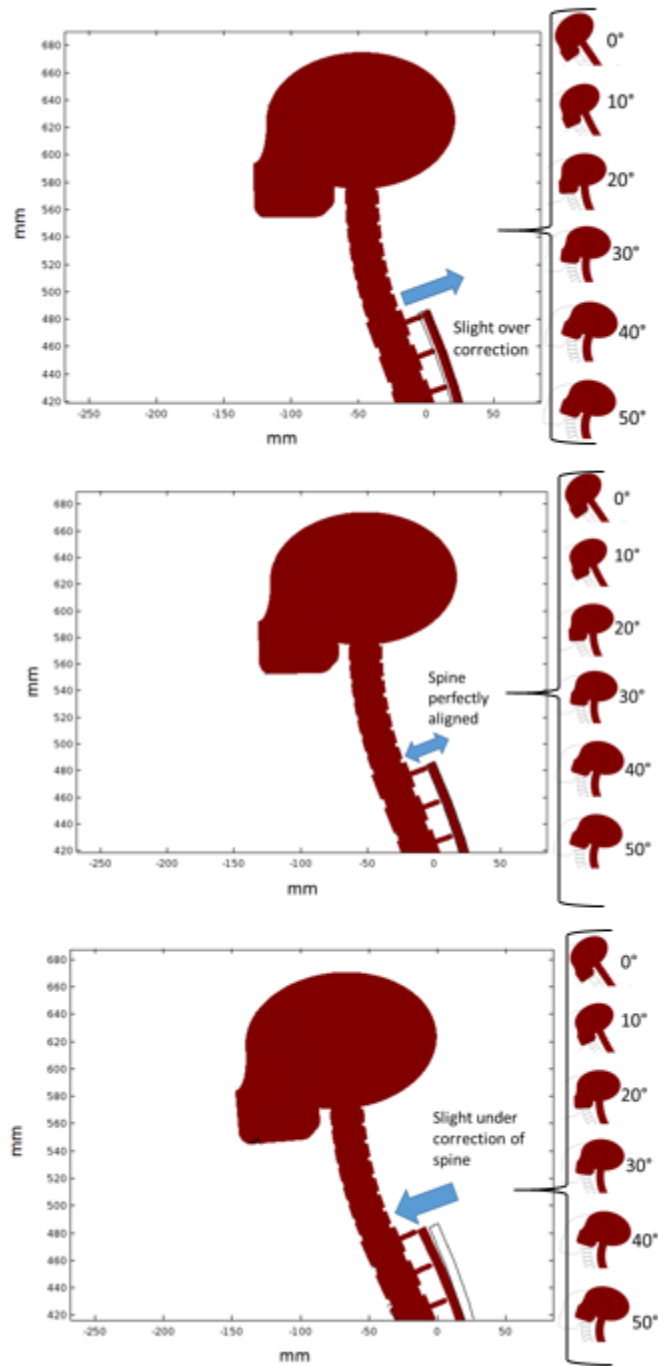


Figure 20: Cervical Sweep at Each Spine Correction Case

The cervical spine will need to perform a parametric sweep ranging from 0° to 50° in increments of 10°. Each sweep will be calculated at the over correction, under correction, and perfect alignment scenarios

2.2.5 Material properties

The perimeter boundary of the vertebrae is given material properties of cortical bone while the bounded internal area is given properties of trabecular bone. The hardware implants are composed of titanium and are given the appropriate parameters. Both bone and metal are designated as linear elastic. The vertebral disc will include only material properties of the annulus in order to keep the geometry simple.

The material properties for the different aspects of the model are found below (Table 2).

Table 2: Material Property Layout for Model

Material type	Material Property	Value
Titanium	Density	4940[kg/m ³]
	Young's Modulus	25e9[Pa]
	Poisson's Ratio	0.33
Bone-cortical	Density	2[g/cm ³]
	Young's Modulus	10000[MPa]
	Poisson's Ratio	0.3
Bone-cancellous	Density	1[g/cm ³]
	Young's Modulus	100[MPa]
	Poisson's Ratio	0.3
Vertebral Disc	Density	5000[kg/m ³]
	Young's Modulus	4[MPa]
	Poisson's Ratio	.45

2.2.6 Physics and Study

COMSOL features a structural mechanics module capable of running structural analysis on various 1, 2, and 3 dimensional models. In the design tree, all domains are programmed to follow linear elastic properties and critical variables relating to elastic moduli, Poisson's ratio and density are all linked to the values already programmed in the materials section of the design tree. A body load is then linked to the domain 1 or the skull. The force directed by the skull is 40N assuming that the average weight of the human head is nearly 6kg. A gravity module is added to measure the gravitational effects of the skull on to the rest of the model. In when the patient is rehabilitated and walking upright, the weight of the skull will have an effect on the spine's curvature as it attempts to retain its normal posture. The contact module contains an important feature that allows bone screws to adhere to the longitudinal element. So wherever the bone screw moves, so will the rod. This adhesion function will allow the longitudinal element to bend with the spine as it gradually loses posture.

Prescribed displacements are the driving force of the model. They allow the model to curl forward in order to meet up with the longitudinal element behind it. By setting prescribed displacements to the T1, T6 and T12 so as to make sure that all bone screws cleanly adhere to their respective regions on the longitudinal element. Issues in the past usually include difficulties in the bone screws reaching the longitudinal element at the same time. Each bone screw would have a prescribed displacement. This served

to be a problem because some bone screws would attach to the longitudinal element before the others pushing it backwards before the others could reach. So only the first middle and last screws were allowed to move. This is similar in surgery. Not all bone screws actually pulled upwards to the longitudinal element but are usually pulled up by the other screws surrounding it. This prevents excessive stress on the spine during surgery. It is worth noting that the prescribed displacement at the bottom T12 is zero because this bone screw is already attached to the longitudinal element. The fixed constraints for the model are the final aspect. The bottom of the T12 cartilage is fixed for that it is the pivotal point the spine moves. By fixing the spine here, no stresses are translated into the lumbar.

2.2.7 Mesh Analysis and Convergence

Meshing of the computational model is crucial for calculating accurate stress profiles based off of the pre-assigned variables and global equations. Tradeoffs in mesh refinement go as follows. The finer an element's mesh the more likely the model will converge and present accurate results. However, processing resources go up exponentially in areas including computational time and computer hardware. A coarser mesh will in turn be less likely to converge and display meaningful results. Coarse meshes are still very efficient as long as the geometry is not overly complicated, so they are faster in solve time and are less taxing on the machine they are ran on.

The 2D feature of this model as stated before is crucial for computational simplicity. By eliminating an entire dimension of the model, run time is drops

significantly. But meshing is still crucial when creating a stress profile emulating real life parameters. In the model the adhesion function allows the simulated bone screws to properly attach to the longitudinal element. The posterior ends of the bone screws are curved and have a higher mesh density than in any other location in the spine. Because the screw and longitudinal element are contact pairs, the higher meshing density allows for more solid contact between them during the simulation. The model as whole is set to extra fine, however the default settings for an extra fine mesh would not suffice for the longitudinal element. Since the longitudinal element is both thin and has a large deformation the mesh size needs to especially modify. By adjusting the free triangular element size to a maximum of 2mm (Fig. 21, 22, 24) instead of the preset 7mm (Fig. 23) a finer mesh can be created to suit the computational needs of this type of hardware.

For FEA, convergence assures the user that changing the size of the mesh will not alter the solutions provided by the model. The mesh must accurately define the model without altering data due to its density (Fig. 25). As a mesh is changes from a coarse mesh to an increasingly finer one, stresses evaluated at a particular point on that model will begin to converge to one value (Fig. 26).

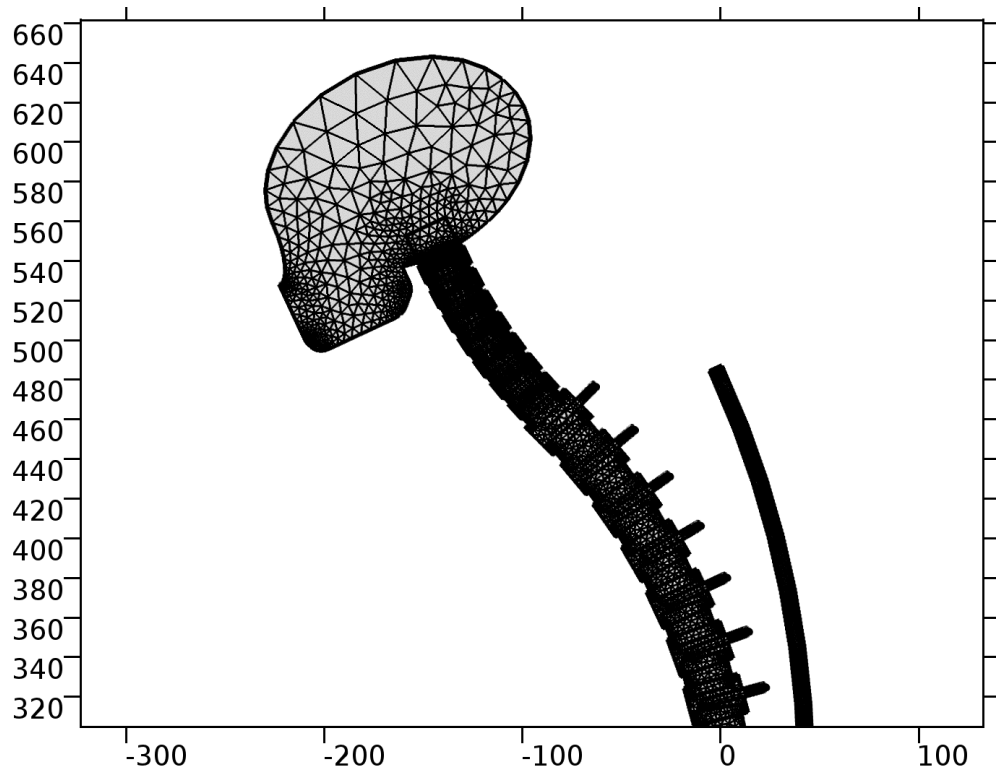


Figure 21: Meshing in Spine

Meshing density in majority of spine (spine-extra fine, longitudinal element-custom with max element size 2mm)

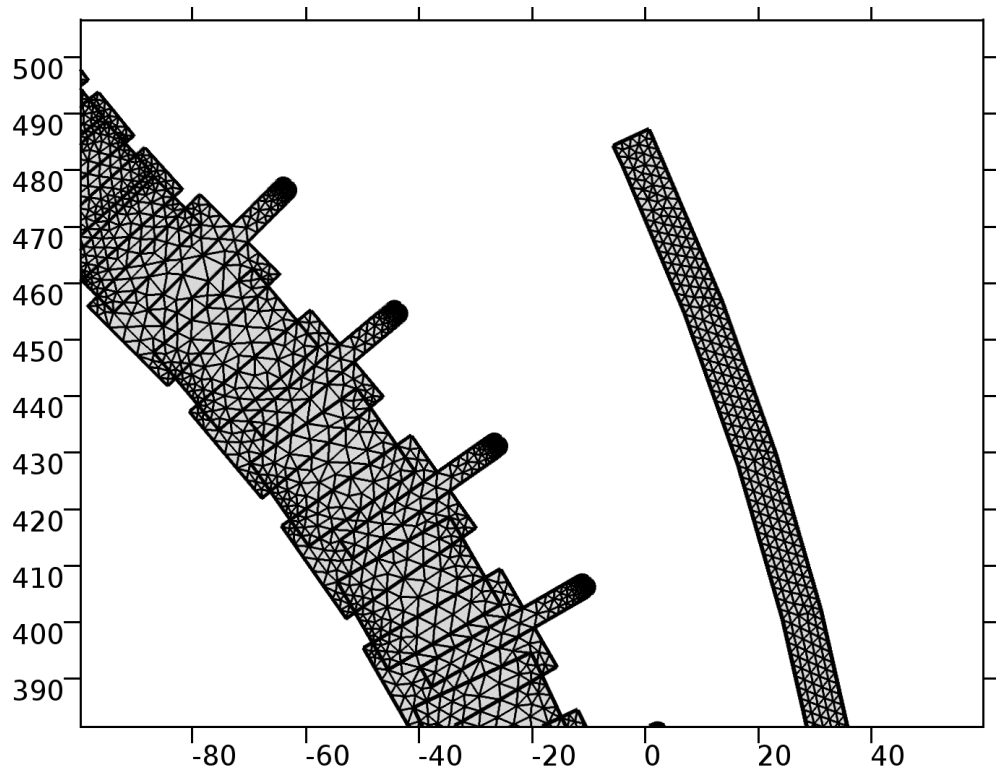


Figure 22: Increased Meshing Density of Bone Screw.

High density in the filleted portion of the screw allows for sound adhesion.

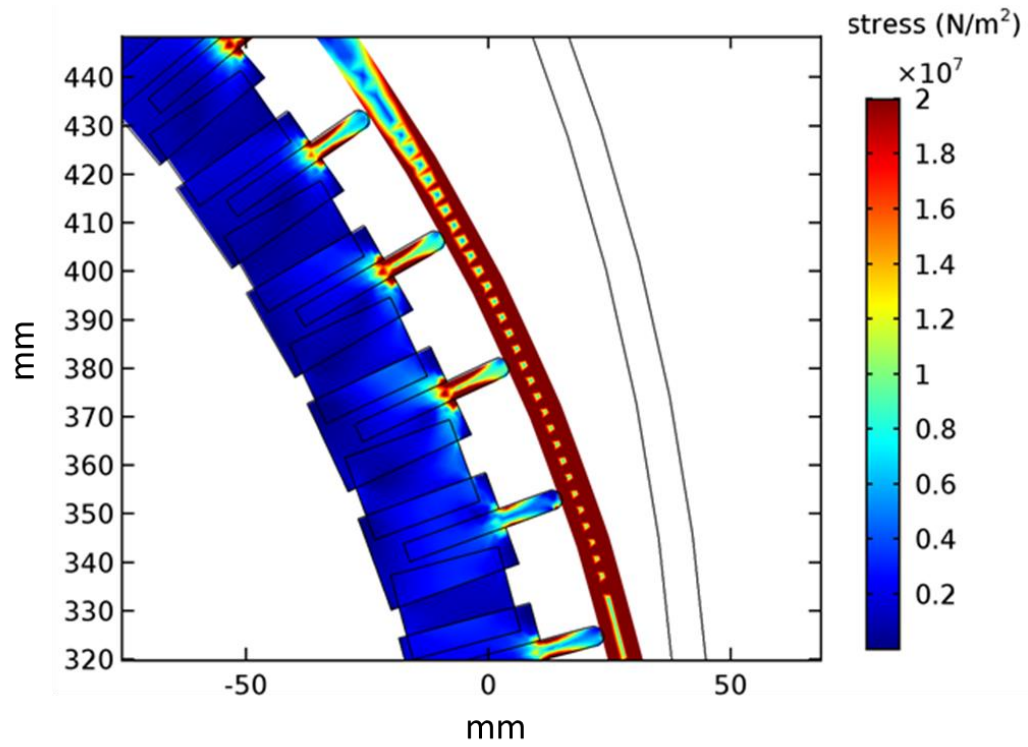


Figure 23: Lack of Proper Meshing Density.

The high stress in the longitudinal element forms a grid pattern and instead of the expected gradient evident in other points in the mode (max element size 7mm)

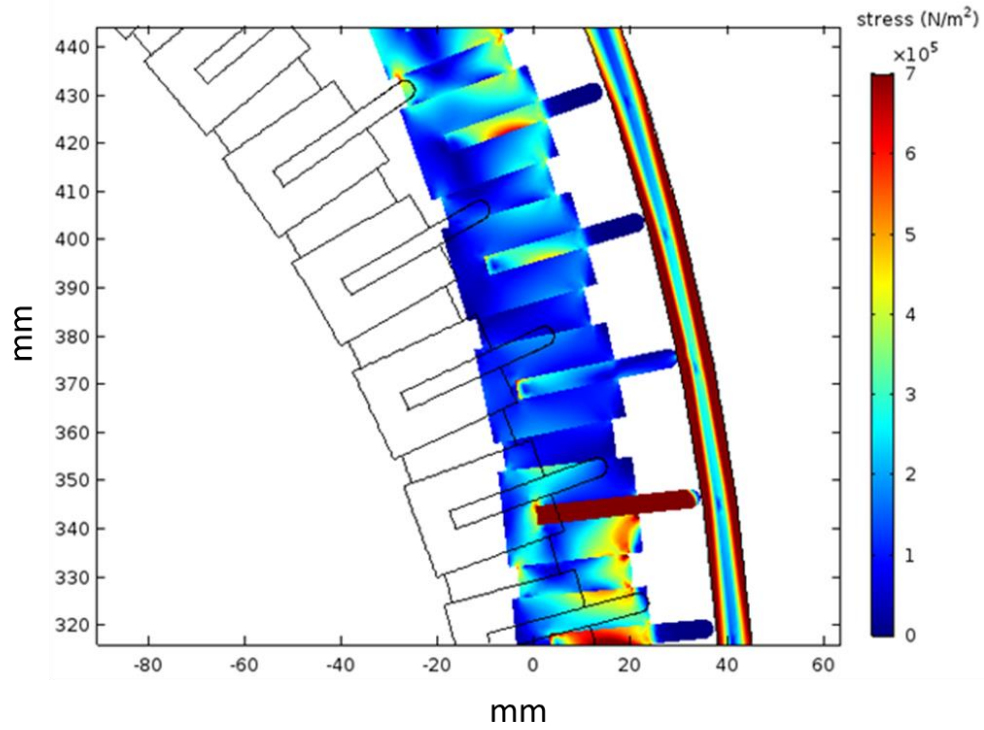


Figure 24: Proper Mesh Density

The high stress in the longitudinal element forms the expected gradient evident in other points in the model

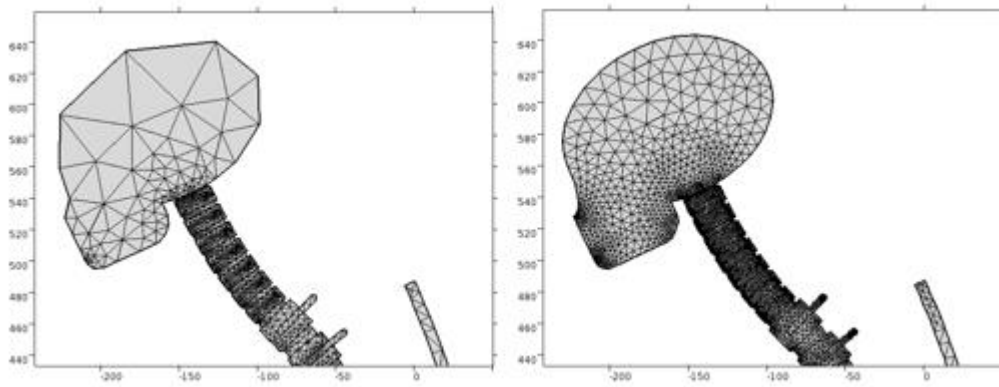


Figure 25: Convergence and Mesh Density

Values will need to converge as the mesh gets finer and finer. The mesh is coarsest on the left and finer on the right. Notice the density on the meshes of both.

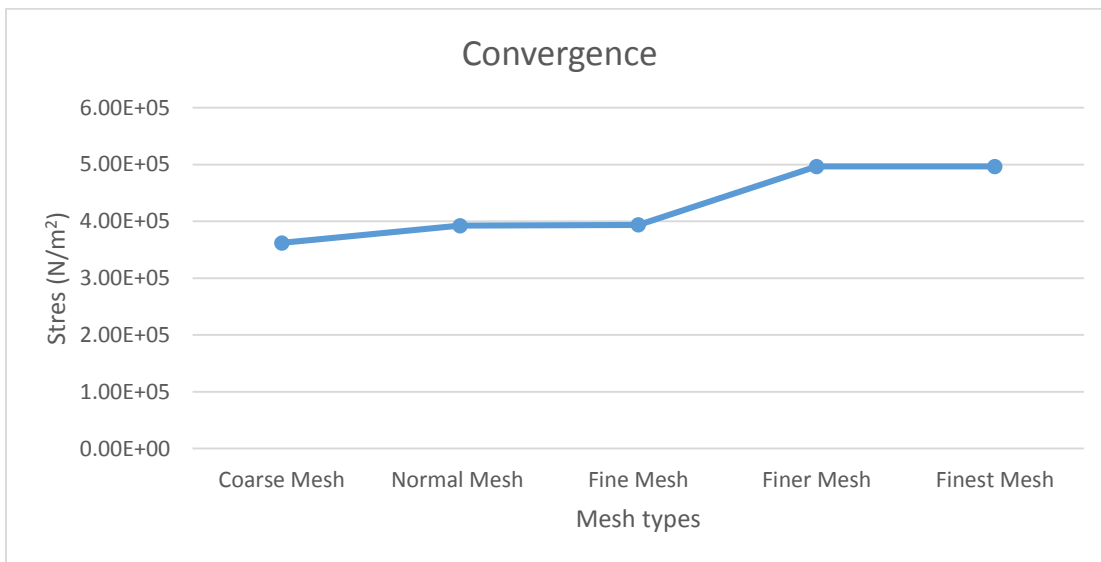


Figure 26: Stress Convergence

Simulation reaches convergence as we decrease our mesh size from coarse all the way to finest.

CHAPTER 3

EXPERIMENTS AND RESULTS

The Experiments

In order to prove or disprove the hypothesis, two questions must be answered dealing with the relationship between the correction types, FHP and spinal stress. First, does change in cervical angle have any significant effect on thoracic spine after being corrected? And second, does under-correction, over correction, or exact angle have influence over the biomechanical favorability of the cervical spine? Biomechanical favorability is the point where the bone has the least amount of stress.

3.1.1 Experiment 1 (Exp. 1)

Experiment one will address the first question, and will help determine if particular angles of the cervical spine actually help prevent stress build up in the thoracic spine. The method goes as follows:

1. Thoracic spine will be surgically corrected and will be set at three cases (over corrected, under-corrected, and exactly aligned)
2. At each case a parametric sweep is ran over the cervical spine from 0° to 50° in increments of 10°.
3. Averages Stresses will be evaluated in the thoracic bone (table 3). A two tailed T-test will be determining if there is a significant difference in stress in the thoracic spine when we are at an over corrected, under corrected or exact alignment. A standard deviation of thoracic stresses will be compared at each of the correction cases at cervical angles between 10° and 50°.

3.1.2 Experiment 2 (Exp.2)

Experiment two will determine if under-correction, over correction, or exact angle have influence over the biomechanical favorability of the cervical spine. The procedure is similar to that of experiment 2 but it is now a matter of what stresses are being analyzed:

1. Thoracic spine will be surgically corrected and will be set at three cases (over corrected, under-corrected, and exactly aligned)
2. At each case a parametric sweep is ran over the cervical spine from 0° to 50° in increments of 10°.
3. Average Stresses will be evaluated in the cervical bone. A line plot showing the maximum stresses at each cervical angle will be made and compared between the 3 correctional cases (figure 27).

3.2 Results

3.2.1 Experiment 1 Results

Average stress in the thoracic vertebrae is calculated at every cervical angle and is used to determine if there is a significant difference between stresses in the thoracic spine due to correction type (over, under, exact). Stress profiles for each angle iteration and at each correctional case can be found in appendix A. At first glance it was very apparent that there were no changes in stress throughout the experiment. A 2-tailed T test between all three of the correctional cases fails to reject null hypothesis with confidence of 95%. This confirms that the change whether the spine was under corrected, over corrected or at an exact angle had little to no effect on the thoracic

spine itself. Standard deviation of spinal stress within each correctional case is comparatively low to the average of stresses at each cervical angle. We can assume that any angle of the cervical spine will have little to no effect on stresses in the thoracic spine (Table 3). FHP is represented at cervical angle 0°.

Table 3: Average Force Experienced by Thoracic Spine (N/m²)

Cervical angle (°)	Over corrected (N/m ²)	Exact Angle (N/m ²)	P-value	Difference statistically significant?
0	1.94E+05	1.95E+05	3.36E-02	No
10	1.93E+05	1.94E+05		
20	1.93E+05	1.94E+05		
30	1.92E+05	1.93E+05		
40	1.92E+05	1.93E+05		
50	1.92E+05	1.93E+05		
SD	816.5	816.5		
Cervical angle (°)	Over corrected (N/m ²)	Under corrected (N/m ²)	P-value	Difference statistically significant?
0	1.94E+05	1.98E+05	1.69E-03	No
10	1.93E+05	1.97E+05		
20	1.93E+05	1.96E+05		
30	1.92E+05	1.95E+05		
40	1.92E+05	1.95E+05		
50	1.92E+05	1.94E+05		
SD	816.5	1472.0		
Cervical angle (°)	Exact Angle (N/m ²)	Under corrected (N/m ²)	P-value	Difference statistically significant?
0	1.95E+05	1.98E+05	1.46E-02	No
10	1.94E+05	1.97E+05		
20	1.94E+05	1.96E+05		
30	1.93E+05	1.95E+05		
40	1.93E+05	1.95E+05		
50	1.93E+05	1.94E+05		
SD	816.5	1472.0		

3.2.2 Experiment 2 Results

Stresses in the cervical are now evaluated and used to help determine the level of stress apparent in the spine at each cervical angle and each of the correctional cases (Fig. 27). Stress profiles of the cervical spine for each step of the experiment are located in appendix B. FHP is represented at the 0° cervical angle.

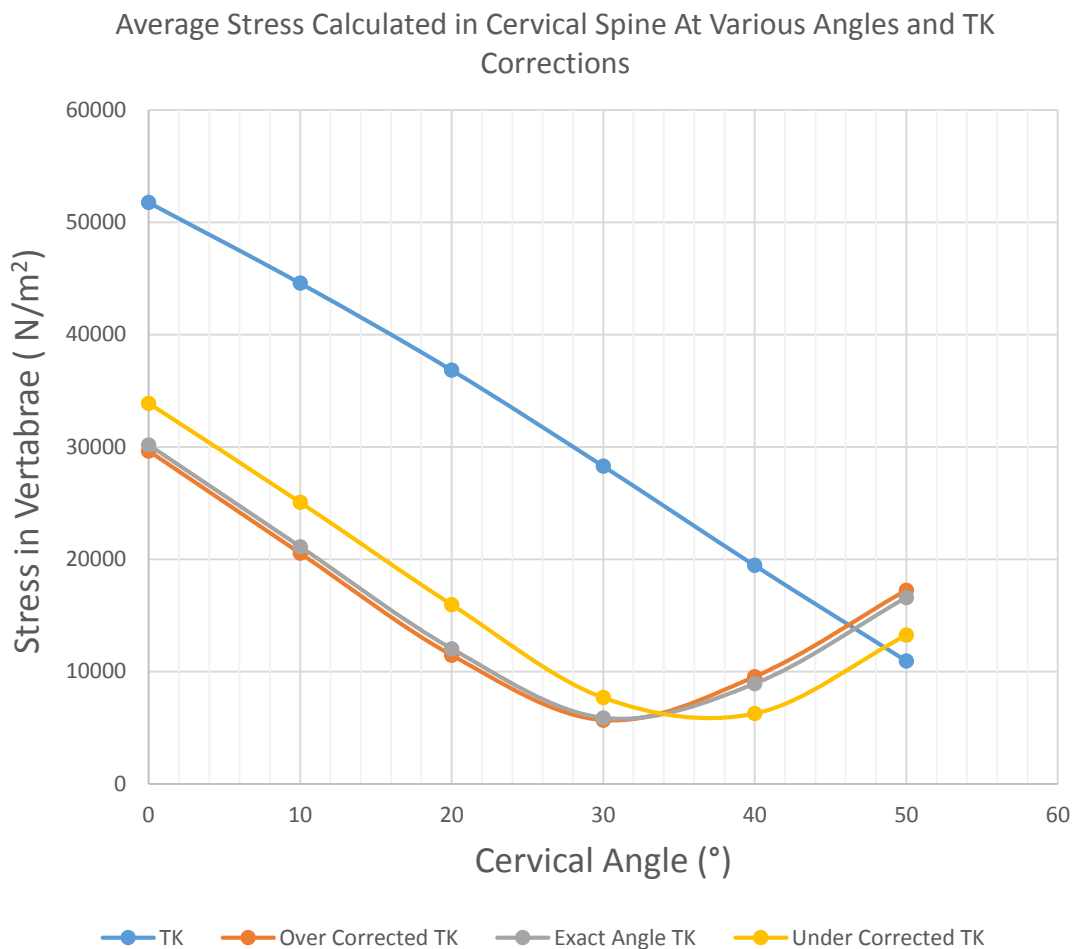


Figure 27: Average Stress Calculated in Cervical Spine At Various Angles and TK Corrections. The orientations of the thoracic spine set by the surgeon will influence the stress profiles of the spine.

CHAPTER 4

DISCUSSION

The first observation relates to the differences in stresses between cervical and thoracic regions. Stresses in thoracic and cervical regions differ by a factor of 10 making it necessary to display data on two different spine stress profiles (refer to appendix A and B).

In the first experiment, we investigated whether or not forward head posture of 0° (and several other angles) affects the stress profile inside the thoracic spine after being corrected. If it does, there is room to speculate that FHP may be a mechanism that can displace stresses in the thoracic spine in attempts to reach a more biomechanically favorable position. According to table III however, there is no observable relationship between changes in cervical angle or the corrected thoracic spine whether it has been over corrected, under corrected, or exactly aligned (refer to appendix A). The rigidity of the thoracic spine due to its new implant could make it unable to move even if the cervical spine is acting as an external force.

In the second experiment it's seen that stress distribution in the cervical spine is highly dependent on its angle. The location of stress concentration varies at different angles but always remains constant towards the base of the neck. The interface between the stiffer portion of the thoracic spine and the free moving cervical spine creates a high stress of 2.00×10^4 N/m² in the C7 vertebrae. High stress at the base of the cervical spine is expected due to PJK which relates to the buildup of stress at the

base of the neck due to a contrast in spinal stiffness at the interface between thoracic and cervical regions. Surgeons have tried combating PJK by using a series of hooks and wires connected between vertebrae and the longitudinal element as a way of blending the stiff areas together as opposed to having an abrupt change.

Localization of stress is correlated with the angle at which the cervical spine is oriented. Between all three thoracic spine configurations (under, over and exact angle) the general location of stress remains similar. At 0°, highest stress forms at the C7 and near equally radiates on the anterior and posterior portions of the vertebral body. At 10°, stress continues to diminish along the anterior and posterior ends of the spine. This continues until a minimum point is reached. It is here where stresses on the anterior and posterior sides of the vertebral body are at its lowest. This minimum however is different for all three TK correction types. When the thoracic spine is over corrected, the minimum point is at 30°. 40° when under-corrected. Exact angle's minimum is slightly lower than the over corrected case. It is at this minimum where the patient should theoretically experience the least stress in the spine making it the most biomechanically favorable (Fig. 27), this makes sense because the minimum also correlates with the C7 plumb line being directly over the sacrum. Beyond the minimum (40° or more) stress will continue to radiate up the spine on the posterior end and diminish on the anterior. At 50° and beyond, all correction types of the thoracic spine post-surgery exude stresses beyond even when the spine was misaligned state.

The minimum for an exact angle thoracic kyphosis is closely related to that of the over correction. The gap between the two minimums is expected to increase the more the surgeon over corrects the spine. The surgeon could go so far as to set the minimum to a value below 30° forcing the patient's biomechanically favorable position to a more stable 20°. This can potentially be very important during patient rehabilitation after surgery. Currently, rehabilitating the cervical spine involves a one size fits all approach where the patients head is forced backwards to retrain their cervical lordosis with no special regard to what angle it corrects to. But after reviewing data in Fig. 277, a surgeon could design a rehabilitation regimen that re-corrects a patients cervical spine to an angle best suited for their level of over or under correctness.

CHAPTER 5

CONCLUSION AND FUTURE WORK

5.1 Conclusion

1. In determining whether FHP post-surgery is habitual or whether that curvature has biomechanically favorability, we found that it is more likely to be the former. A 0 degree FHP has the highest stress of all correction types, to the point where that stress is causing fractures in the cervical spine. It is worthy to mention that FHP is seen in older patients who underwent TK correction surgery. It would appear that older patients are not as plastic as children and have a harder time retraining to a new cervical angle. Older patients will therefore revert more often to the same downward head angle as they did before surgery.
2. A patient's most biomechanically favorable cervical angle will also change depending on the correctness of their spine. If a patient is over corrected, a 30° curve may be more comfortable for that patient than one who is under corrected. This becomes important when rehabilitating because not every patients biomechanically favorable angle will be the same. Surgeons will need to recognize this and develop methods of retraining the cervical spine more accurately.

5.2 Future Work

Finite element analysis in the field of biomechanics and orthopedics has a very bright future as more complex and sophisticated models come into existence. By using better geometric capture with 3D scanners and CT images there will soon come a day where a personalized FEA model can be generated for a specific patient to run biomechanical analysis on before surgery is even performed. Personalized diagnosis that uses FEA models as surgical consultants could be an exciting new step in the world of personalized medication. The accuracy of the model will be strongly dependent on the realism of the model and its ability to produce converged results over that complex model. Future work concerning this spine model is mainly focused on the addition of more features to the models' authenticity.

Most anatomical features on the model are a series of blocks and other miscellaneous geometric shapes. Bones in actual patients have softer edges and contours that can potentially alter the output of a model. By constructing anatomical parts that are not strictly straight lines or edges, users can see the minor nuances in bone fracturing with a more detailed model shape. Future work will involve using CT slices of the spine to create geometry domains in COMSOL. Software like MIMICS can be used to create 2D geometry for the midsagittal plane.

MIMICS has the ability to capture CT slices and transform them into a geometric domain, and then stitch multiple images together to a single cohesive 3D rendering. A 3D rendering of the spine would be useful in studying how kyphosis and scoliosis can occur

in a patient at the same time. This next potential step in the project would use this 3D rendering to explore how implants built for the correction of kyphosis biomechanically affects the correction of scoliosis and what alternate spinal fixation methods could be used in the future for its correction.

Even before a 3D rendering is created there is still work to be done in adding more anatomical feature to the current 2D model. Ligaments and muscles are significant features that add to the overall stabilization and movement of the spine. These elements unfortunately were not included due to the narrower scope of this research. Studies that focus on how the chin brow angle could be affected by learned muscle memory is an example of the questions added muscles and ligaments can cause the model. Adding muscles can be used to determine the fatigue level felt by muscles at different kyphosis angles as well. Muscles and ligaments would be programmed differently in the model. Muscles for example would use point loads with force vectors in the direction of their anatomical counterparts and then calculating what force load are required to displace the spine in its appropriate orientation. Similarly, ligaments would be responsible for upholding the spine but would instead use a “cable” or spring like feature whose insertion and origin is located at different regions along the spine.

In order for various muscles and ligaments to attach, the vertebrae must include more attachment point sites. Features like pedicles and spinal processes become a requirement when attaching these added anatomical features. As mentioned before, it was opted to exclude their parts in order to maintain the simplicity of the model, and muscles and ligaments were not a mandatory design criterion. The shape profiles of

these remaining processes would be constructed with simple geometric shapes, but to make a more realistic contour, programs like MIMICS would be used.

Another feature that could be added are ribs and the organs contained with the rib cage. By adding ribs and organ surrogates to the model another moment force could be generated and studied, similar to that of weight of the skull and brain. The total mass of the rib cage could carry more insight as to the stress factors related to thoracic kyphosis as a patient tries to support not only the weight of their head but the weight of rib cage as well. It would also be worthwhile to study how the effects of poor spinal curvature could compress the organs in the rib cage which is just one of many ailments cause by thoracic hyper kyphosis.

The use of different fixation methods can also be used to compare its efficacy with that of a spinal rod system. Surgeons use a plethora of techniques to straighten the spine including back braces, wire and hooks and spinal fusion just to name a few. By creating geometry that represents these different correctional methods, result can be compared to the gold standard spinal rod to validate its true proficiency in advanced spinal correction. Different spinal rod configuration could also be tested. In some cases, the longitudinal element does not include the entire length of the spine but may only cover significant portions. Rearrangement of these different segments could be tested to optimize spinal rod placements for maximized comfort for the patient.

Lastly, more work will need to go into the internal structure of the intervertebral disc. Addition of a true annulus would be better than just adding the fibrous regions. This

model could potentially be used to study how a patient walking gate affects the vertebral column. The annulus becomes very important when the spine begins to compress from the walking movements.

The main goal for the future is to essentially add more features to the model. Added features provided added realism and opens the opportunities for more mechanical spinal testing which can answer more questions related to the onset, progression, and finally treatment for thoracic kyphosis.

APPENDICES

Appendix A. Stress in Thoracic Vertebrae at Under Corrected, Over Corrected, and Exact Angle

Simulation of the surgical procedure occurs in steps dependent on the paravalues and triangular function. Shown here are the stress profiles related to the spine during the surgical procedure. After the spine has been corrected an investigation between the movement on the head and the corrected thoracic spine is performed and the nature of their relationship is analyzed (Fig. A1, A2, A3).

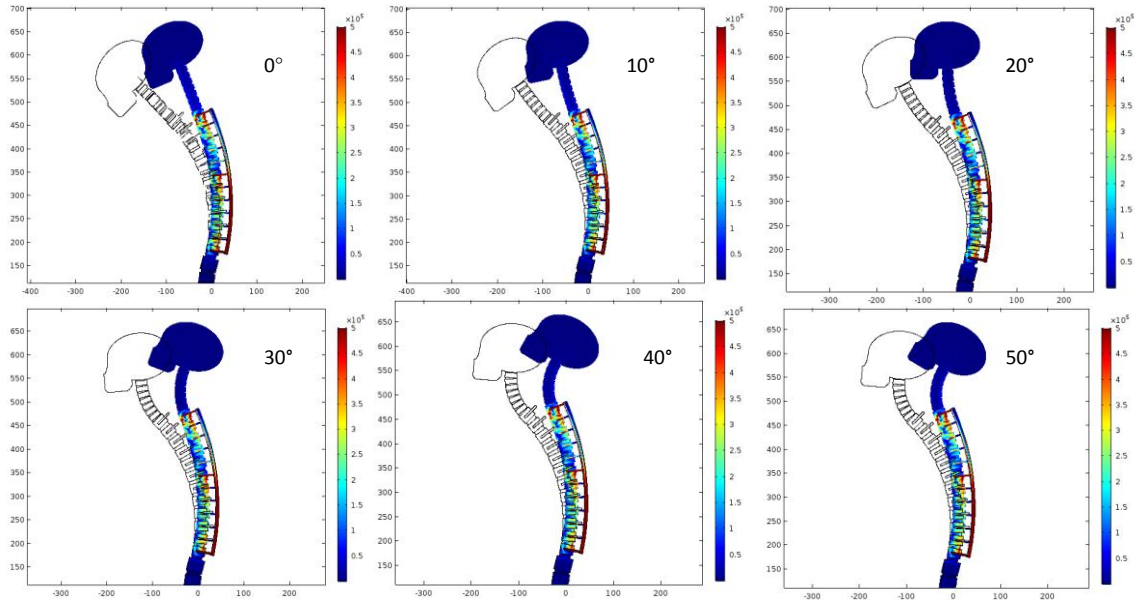


Figure A 1: Over Correction with Cervical Sweep (Exp. 1)

Thoracic spine at an over corrected orientation at cervical angles between 0° and 50°. Stress bar is in units of (N/m²). Ordinate and Abscissa in millimeters. Notice little change in stress and displacement in the thoracic spine.

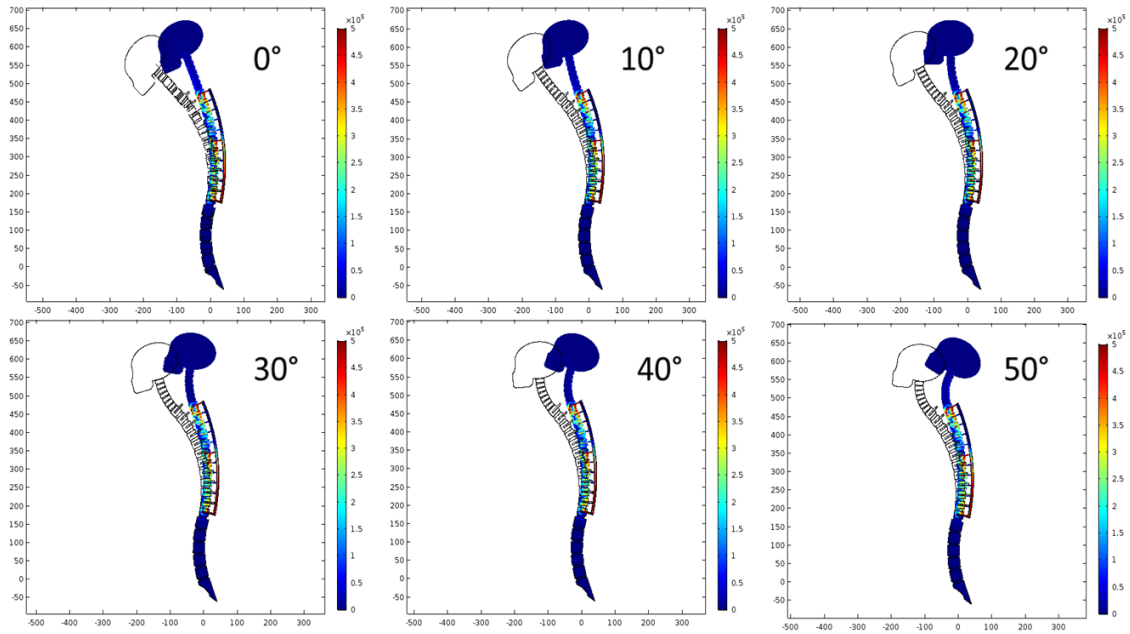


Figure A 2: Exact Alignment with Cervical Sweep (Exp.1)

Thoracic spine at exact alignment at cervical angles between 0° and 50° . Stress bar is in units of (N/m^2) . Ordinate and Abscissa in millimeters. Similar to the over corrected case, there is little change in stress and displacement in the thoracic spine.

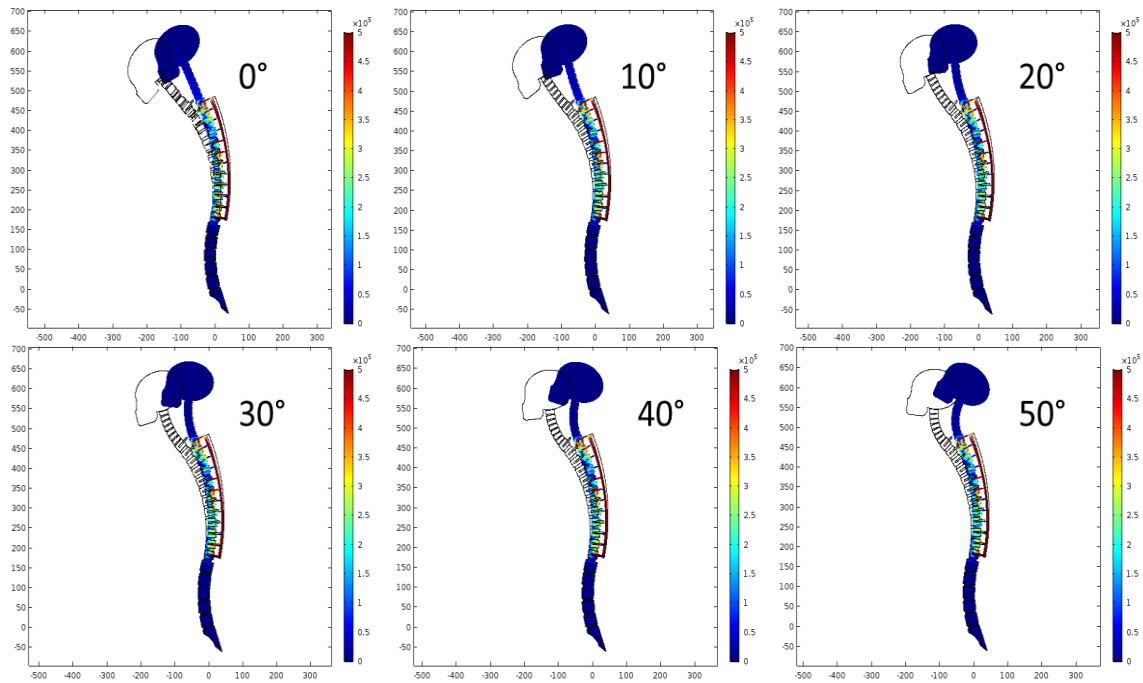


Figure A 3: Under Correction with Cervical Sweep (Exp. 1)

Thoracic spine at an under corrected orientation at cervical angles between 0° and 50°. Stress bar is in units of (N/m²). Ordinate and Abscissa in millimeters. Similar to the over corrected case, there is little change in stress and displacement in thoracic spine.

Appendix B. Stress in Cervical Vertebrae at Under corrected, Over corrected, and Exact Angle

Forward head posturing is caused by the lack of curvature in the cervical spine. This results after surgery and can be linked to the patient's natural habitual tendency to position their head as it was before surgery. The relationship between stresses in the cervical spine is related to the positioning of the thoracic spine (over corrected (Fig. B1), under corrected (Fig. B2) or exact alignment (Fig. B3)).

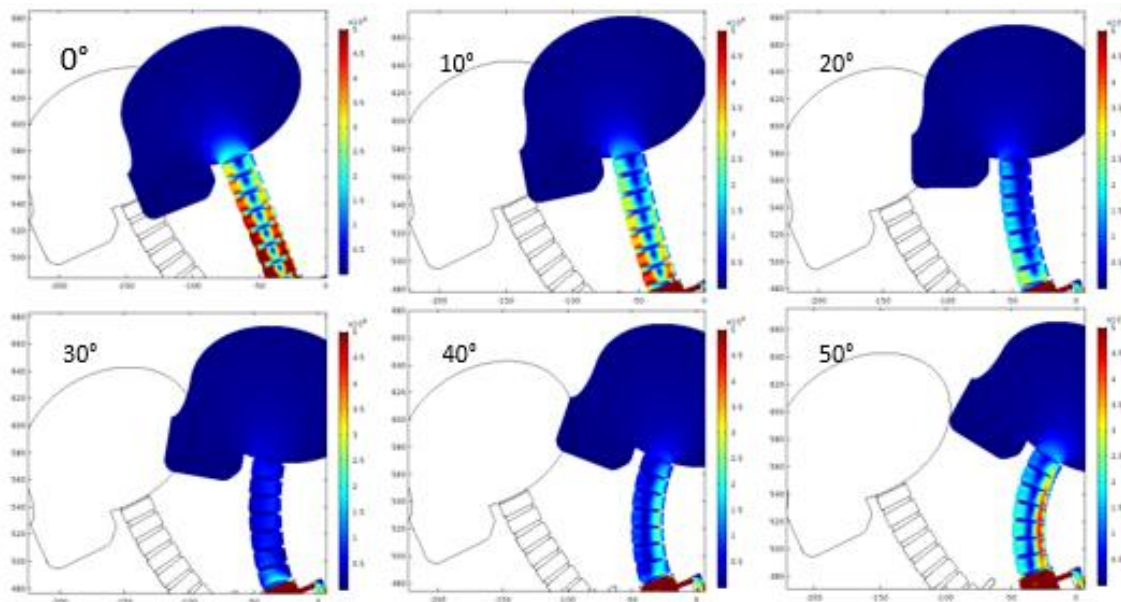


Figure B 1: Over Corrected with Cervical Spine Sweep (Exp. 2)

Cervical spine at an over corrected orientation at cervical angles between 0° and 50°. Stress bar is in units of (N/m²). Ordinate and Abscissa in millimeters. Notice high stress radiating from the base of the cervical spine.

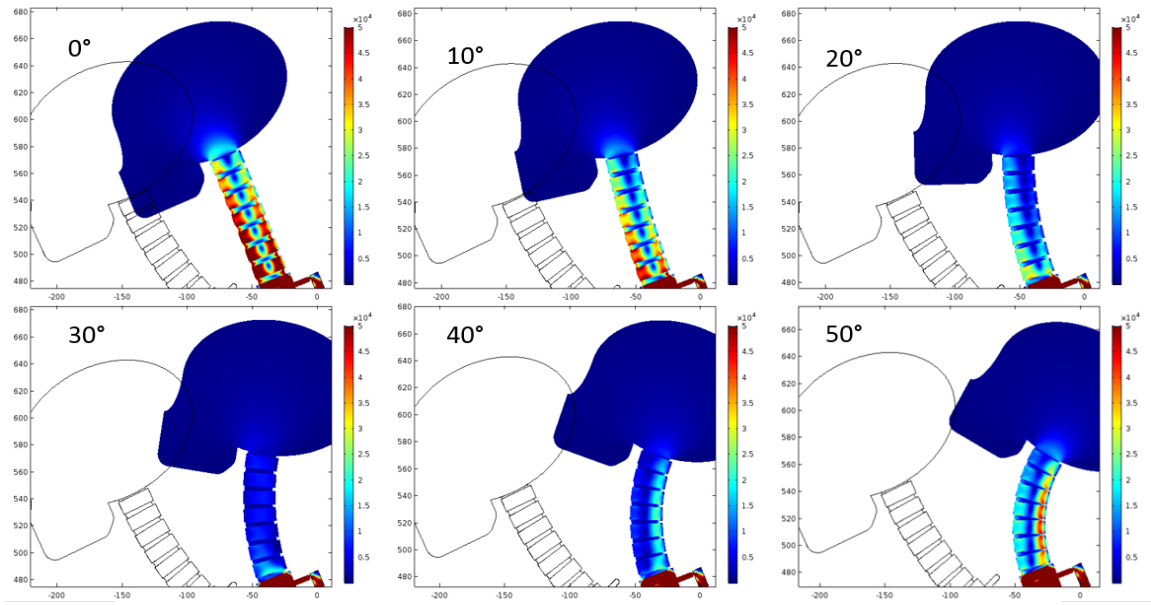


Figure B 2: Exact Alignment Cervical Sweep (Exp. 2)

Cervical spine at an exactly aligned orientation at cervical angles between 0° and 50°. Stress bar is in units of (N/m²). Ordinate and Abscissa in millimeters. Overall stresses are lower than in over corrected case up until 30° but the pattern distribution is more or less the same.

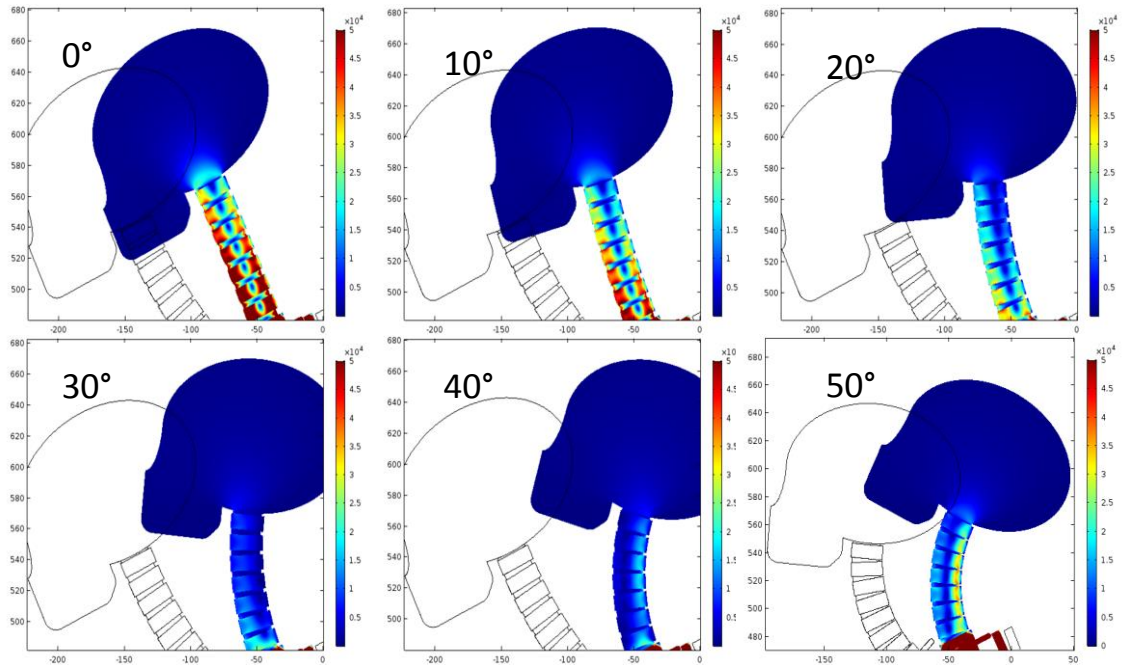


Figure B 3: Under Correction with Cervical Sweep (Exp.2)

Cervical spine at an under corrected orientation at cervical angles between 0° and 50°. Stress bar is in units of (N/m²). Ordinate and Abscissa in millimeters. Stresses are higher between 0° and 30° than when spine is under corrected or at an exact angle. However cervical stresses are lower at 40° and 50° in an under corrected spine than in the over corrected.

REFERENCES

1. Nicholson, P.h.f., X.g. Cheng, G. Lowet, S. Boonen, M.w.j. Davie, J. Dequeker, and G. Van Der Perre. "Structural and Material Mechanical Properties of Human Vertebral Cancellous Bone." *Medical Engineering & Physics* 19.8 (1997): 729-37. Web. 15 Feb. 2017.
2. Virgin, W. J. "Experimental Investigations into the Physical Properties of the Intervertebral Disc." *The Journal of Bone and Joint Surgery. British Volume. U.S. National Library of Medicine*, Nov. 1951. Web. 01 Jan. 2017.
3. Hartman, William F. "Deformation and Failure of Spinal Materials." *SpringerLink*. Kluwer Academic Publishers, n.d. Web. 21 Feb. 2017.
4. Bot, S. "Biomechanical Analysis of Posture in Patients with Spinal Kyphosis Due to Ankylosing Spondylitis: A Pilot Study." *Rheumatology* 38.5 (1999): 441-43. Web. 20 Jan. 2017.
5. Narayan, Yoganandan, Kumaresan, Srirangam and Frank A. Pintar. "Geometric and Mechanical Properties of Human Cervical Spine Ligaments." *Journal of Biomechanical Engineering* 122.6 (2000): 623. Web. 20 Feb. 2017.
6. Cortes, Daniel H., Nathan T. Jacobs, John F. Delucca, and Dawn M. Elliott. "Elastic, Permeability and Swelling Properties of Human Intervertebral Disc Tissues: A Benchmark for Tissue Engineering." *Journal of Biomechanics* 47.9 (2014): 2088-094. Web. 21 Feb. 2017.

7. Roussouly, Pierre, and Colin Nnadi. "Sagittal Plane Deformity: An Overview of Interpretation and Management." *European Spine Journal* 19.11 (2010): 1824-836. Web.
8. Joseph, Samuel A., Anthony P. Moreno, Jared Brandoff, Andrew C. Casden, Paul Kuflik, and Michael G. Neuwirth. "Sagittal Plane Deformity in the Adult Patient." *Journal of the American Academy of Orthopaedic Surgeons* 17.6 (2009): 378-88. Web. 9 Oct. 2016.
9. Chen, Chen-Sheng, Wen-Jer Chen, Cheng-Kung Cheng, Shyh-Hua Eric Jao, Shan-Chang Chueh, and Chang-Chih Wang. "Failure Analysis of Broken Pedicle Screws on Spinal Instrumentation." *Medical Engineering & Physics* 27.6 (2005): 487-96. Web. 18 Feb. 2017.
10. Smith, Micah W., Prokopis Annis, Brandon D. Lawrence, Michael D. Daubs, and Darrel S. Brodke. "Acute Proximal Junctional Failure in Patients with Preoperative Sagittal Imbalance." *Spine Journal*. Elsevier Inc., 16 Oct. 2015. Web. 28 Nov. 2016.
11. Suk, Se-li, Won-Joong Kim, Sang-Min Lee, Jin-Hyok Kim, and Ewy-Ryong Chung. "Thoracic Pedicle Screw Fixation in Spinal Deformities." *Spine* 26.18 (2001): 2049-057. Web. 22 Feb. 2017.
12. Niinomi, Mitsuo. "Mechanical Properties of Biomedical Titanium Alloys." *Materials Science and Engineering: A* 243.1-2 (1998): 231-36. Web.
13. Saunders, Stephen E., DC, Dennis Woggon, DC, Christian Cohen, BS, DC, and David H. Robinson, Phd. "Improvement of Cervical Lordosis and Reduction of Forward

- Head Posture with Anterior Head Weighting and Proprioceptive Balancing Protocols." (n.d.): n. pag. 27 Apr. 2003. Web.
14. Schroerlucke, Samuel R., Behrooz A. Akbarnia, Jeff B. Pawelek, Pooria Salari, Gregory M. Mundis, Muharrem Yazici, John B. Emans, and Paul D. Sponseller. "How Does Thoracic Kyphosis Affect Patient Outcomes in Growing Rod Surgery?" *Spine* 37.15 (2012): 1303-309. Web. 1 Jan. 2017.
15. June, Quek, Pua Yong-Hao, Clark Ross A., and Bryant Adam L. "Effects of Thoracic Kyphosis and Forward Head Posture on Cervical Range of Motion in Older Adults." *OMICS International*. OMICS International, n.d. Web. 5 Jan. 2017.
16. Hey, H. W., K. A. Tan, C. S. Neo, E. T. Lau, D. A. Choong, L. L. Lau, G. K. Liu, and H. K. Wong. "T9 versus T10 as the Upper Instrumented Vertebra for Correction of Adult Deformity-rationale and Recommendations." *The Spine Journal : Official Journal of the North American Spine Society*. U.S. National Library of Medicine, May 2017. Web. 28 Apr. 2017.
17. Haghpanahi, Mohammad, and Reza Mapar. "Development of a Parametric Finite Element Model of Lower Cervical Spine in Sagittal Plane." *2006 International Conference of the IEEE Engineering in Medicine and Biology Society* (2006): n. pag. Web. 12 Feb. 2017.