A Thesis

entitled

Biomechanical Evaluation of Vertebral Augmentation to Compare Biocure Cement with PMMA

by

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Submitted to the Graduate Faculty as partial fulfillment of the requirements for the Master of Science in Bioengineering

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An Abstract of

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A vertebral compression fracture occurs when the vertebral body of the spine is collapsed due to trauma. Usually, the magnitude of trauma required to cause a vertebral compression fracture is quite large. However, in certain circumstances and population groups such as elderly people or in individuals with osteoporotic disorder and cancer, vertebral compression fracture may occur with a minimal compressive force. The most common location of such fractures is in the anterior column of the thoracolumbar region of the spine forming an anterior wedge fracture. Vertebral augmentation is a minimally invasive surgical procedure of cement injection in a collapsed vertebra used to stabilize and reinforce such vertebral compression fracture and has proven to be effective in relieving pain compared to conservative treatments.

Complications facing vertebral augmentation procedure are cement leakage into the spinal canal or intervertebral disc, cement curing temperatures, and recurrent fractures in the vertebral bodies adjacent to the augmented vertebrae. This study was focused on the
adjacent level fractures caused due to possible mismatch between augmented osteoporotic bone and high modulus cement resulting in inappropriate load sharing. Biocompliant Biocure bone cement with modulus matching trabecular bone was evaluated and compared with commercially available high modulus PMMA cement.

A three dimensional, non-linear osteoporotic T12-L3 finite element model was developed and validated for this study. The finite element mesh of the L1 vertebral body was modified to form oval concentric cylinders in order to simulate a vertebral augmentation in the center of the vertebral body. A compression fracture was simulated in the L1 vertebral body by using a modulus reduction criterion. Vertebral augmentation was then simulated at two distinct cement stiffness values of Biocure and PMMA bone cement to compare with osteoporotic model. The load control protocol was used to evaluate motion of the spine. Parameters like Von Mises stresses, intradiscal pressure, facet loads, and stress contour plots were analyzed for each model. There was no effect of the cement stiffness on the motion of the FE model of the spine when augmented. The load transferred through the facets decreased and the corresponding endplate stresses increased with increase in cement stiffness. The study also indicated that when a particular level was treated, a higher effect on the stresses was observed at an adjacent upper level than an adjacent lower level. Trends were most evident in the compression loading configuration and at the largest cement volume.

Cadaveric kinematic experiments were carried out to investigate the motion in the spinal segment as well as to analyze the increase in intradiscal pressure at adjacent levels when loaded and augmented with Biocure and PMMA cement. Fatigue performance was studied by mechanically loading the augmented vertebral segment (with Biocure and
PMMA) for 1.25 million cycles. This was followed by computer tomography scan evaluation for potential endplate cracks. Additionally, destructive sectioning of augmented vertebrae was undertaken for possible cement cracks due to fatigue loading.

The results of this study provide an insight on the mechanics of vertebral augmentation with respect to variation in cement properties. Load sharing with Biocure bone cement augmentation was similar to intact compared to PMMA. Fatigue analysis helped in evaluating the long term sustainability of Biocure cement when augmented in the osteoporotic bone. This suggests that cement stiffness lower than the commercially available (high modulus) bone cement may be beneficial from biomechanical perspective.
I would like to dedicate this work to my parents — thank you for your love, support and encouragement.
Acknowledgements

This thesis is an outcome of more than two years of research, hard work and perseverance. It has been an excellent learning experience that also inculcated in me a sense of responsibility and discipline.

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Chapter 1

Introduction

1.1 Overview

In this chapter, a brief description of the thoracolumbar spine anatomy of the human spine is discussed, followed by the reason for studying osteoporosis. Vertebroplasty and kyphoplasty surgical procedures are explained in brief. The chapter will conclude with the rationale and scope of this study.

1.2 Thoracolumbar spine anatomy

The spine is the core structure of the body. It is responsible for protecting the spinal cord and nerve roots, a base for attachment for ligaments, tendons and muscles and provides flexibility and mobility as well as supporting much of the weight of the upper body including the head, arms, and torso. The spinal column is made up of 33 bones known as vertebrae. Human spine extends from the skull to the pelvis and is made up of 33 individual bones termed vertebrae (figure 1.1). The vertebrae are stacked on top of each other and they group into four regions: Cervical, Thoracic, Lumbar and Sacrum.
Between each vertebra is an inter-vertebral disc. The inter-vertebral disc acts as a cushion between each vertebra. The intervertebral discs are fibrocartilaginous cushions serving as the spine's shock absorbing system. The vertebrae, discs, and ligaments are all connected to make a functional spinal unit or a motion segment. Each motion segment has the ability to move with six degrees of freedom, three degrees of rotation and three degrees of translation. The natural curves in the spine, kyphotic and lordotic, provide resistance and elasticity in distributing body weight and axial loads sustained during movement.

The thoracolumbar junction is made up of five vertebrae where the thoracic and lumbar regions of the spine meet. This region is considered to include the T10 through L2 vertebrae. The thoracolumbar junction is unique in that this is where the spine’s curvature changes from a kyphotic to lordotic making this region almost straight making it more susceptible to compressive injuries.

The vertebrae and discs are held together by groups of ligaments. Ligaments connect bone to bone, whereas tendons connect muscle to bone. In the spine, tendons connect muscles to the vertebrae. Ligaments and tendons are fibrous connective tissues made up of densely packed collagen fibers. The ligaments and tendons help to stabilize the spine and guard against excessive movement in any one direction. The spinal joints are called facet joint and they help to make the spine flexible. The joint surfaces are coated with cartilage allowing joints to move or glide smoothly (articulate) against each other. These joints allow flexion (bend forward), extension (bend backward), and twisting motion.
Each vertebra is made up of a vertebral body, two pedicles, two laminae, two transverse processes, a spinous process, and two sets of facets. The vertebral body is the main load bearing part of the vertebrae and makes up the anterior portion of the vertebrae. The pedicles and laminae are the bony areas connecting the processes to the vertebral body. The spinous and transverse processes are located on the posterior side of the vertebrae and connect the posterior ligaments to the spinal column. There is a superior set of facets and an inferior set of facets. The facets connect vertebrae together and limit the motion of a motion segment in flexion, extension, lateral bending, and axial rotation. The facets are surrounded by cartilage, allowing them to glide smoothly over each other. The inter-vertebral disc is made up of two main parts, the annulus fibrosus and the nucleus pulposus. The annulus fibrosus makes up the outer portion of the disc is made of concentric sheets of collagen fibers. The nucleus pulposus is soft core of the disc and is more of water based material. Figures 1-1 through 1-2 illustrate the anatomy of the functional spinal unit.
Figure 1.1 (a) Lateral view of thoracolumbar motion segment (b) Transverse view of Typical Thoracolumbar vertebra (c) Ligaments of the thoracolumbar spine
1.3 Osteoporosis:

Osteoporosis is a disorder of progressive bone loss associated with an increased risk of fractures. The term osteoporosis literally means porous bone. The disease often develops unnoticed over many years, with no symptoms or discomfort until a fracture occurs. This disorder of the skeleton weakens the bone, disrupts the bone micro-architecture, and alters the amount and variety of non-collagenous proteins in bone.
In the United States, 44 million people have low bone density which amounts to 55% of the U.S. population 50 years-old and over. More than 10 million people have osteoporosis and almost 34 million more have low bone density. These statistics give credence to the study of osteoporosis and other age-related disorders and the search for easy, quick and cost-effective treatment options for it. Osteoporosis is a major health problem, affecting 28 million Americans and contributing to an estimated 1.5 million bone fractures per year. One in two women and one in five men older than 65 years will sustain bone fractures caused by osteoporosis.

Bone is a living tissue, just like any other tissue in the body. Old bone is replaced by new bone all the time. Every day, cells called osteoclasts break bone down, and another type of cell known as osteoblasts, try to repair the damage by forming new, healthy bone. Throughout our life, the speed of this process changes: Up until our early 30's, the osteoblasts make more bone than is lost. These are the years when the body builds the skeleton up to hold the greatest amount of bone. From our 30's to our 50's, we lose old bone and make new bone at about the same rate. When women reach menopause and men reach their 50's, the process changes again. Then, our bodies always make less bone than we lose. At some point, this bone loss may cause the bone to fracture. Along with aging some factors like nutrition and lifestyle, medications and other illness can be responsible for osteoporosis.

Dual energy X-ray absorptiometry (DXA, formerly DEXA) is considered the gold standard for the diagnosis of osteoporosis. Osteoporosis is diagnosed when the bone
mineral density is less than or equal to 2.5 standard deviations below that of a young adult reference population. This is translated as a T-score. The World Health Organization has established the following diagnostic guidelines:

- **T-score** -1.0 or greater is "normal"
- T-score between -1.0 and -2.5 is "low bone mass" (or "osteopenia")
- T-score -2.5 or below is osteoporosis.

### 1.4 Vertebral Compression Fractures:

Low bone mass and bone deterioration caused by osteoporosis and leading to increased susceptibility to fracture is an issue that is seen in escalating numbers as the US population ages. These vertebral fractures will primarily lead to pain, collapse of the vertebral column, and loss of physiologic posture. Collapse of the vertebral column and loss of posture can lead to cardiovascular, pulmonary, and neurological diseases [1]. A fracture that collapses a spinal vertebra as a result of the compression of bone, much like a sponge collapses under the pressure of one's hand is termed as a vertebral compression fracture (VCF) [1]. The fracture causes an immediate onset of severe localized pain that can radiate around the waist in a band-like fashion and is made intensely worse with body motions. Fractures commonly occur in the thoracolumbar region because a change in the facets provides less resistance to anteroposterior displacement at these levels [2].
Specifically, they occur at the point where the thoracic vertebrae of the midback meet those of the lumbar region of the lower back [3].

When vertebral compression fractures occur, they may cause sudden, sharp pain or no pain. Pain may gradually develop. It may be mild to very severe. Sitting for a long time, standing, bending forward, twisting, carrying heavy objects, and walking usually make the pain worse. Activities like sneezing and coughing may cause pain. People with many large vertebral compression fractures can lose height, and the back may become rounded and bent. This condition is sometimes called a dowager's hump (kyphosis). Standing up straight may be impossible. Some people have difficulty bending, reaching, lifting, climbing steps, and walking. Rarely, vertebral compression fractures damage the spinal cord or spinal nerves. In such cases, symptoms include weakness in the leg, numbness, paralysis, and loss of control of the bowels or bladder (fecal or urinary incontinence).

A vertebral compression fracture occurs when the bones of the spine become broken due to trauma. Usually the trauma necessary to break the bones of the spine is quite large. In certain circumstances, however, such as in elderly people and in people with cancer, these same bones can break with little or no force. The vertebrae most commonly broken are those in the lower back.

Until recently, doctors were limited in how they could treat osteoporosis-related spine fractures. Pain medications, bed rest, bracing or invasive spinal surgery were the only options available. Today there are two promising therapeutic and preventive treatments for compression fractures. They are called vertebroplasty and kyphoplasty.
Limitations in the traditional treatments of vertebral compression fractures have led to the refinement of such procedures as vertebroplasty and kyphoplasty. These procedures provide new options for compression fractures and are designed to relieve pain, reduce and stabilize fractures, reduce spinal deformity, and stop the “downward spiral” of untreated osteoporosis.

Additional benefits of these procedures include:

- Short surgical time
- Only general or local anesthesia required
- Average hospital stay is one day (or less)
- Patients can quickly return to the normal activities of daily living
- No bracing required

Both vertebroplasty and kyphoplasty utilize a cement-like material that is injected directly into the fractured bone. This stabilizes the fracture and provides immediate pain relief, in many cases. Kyphoplasty has the additional advantage of being able to restore height to the spine, thus reducing deformity. After either procedure, most patients quickly return to their normal daily activities [4].
1.5 **Introduction to Vertebral Augmentation**

Low bone mass and bone deterioration caused by osteoporosis and leading to increased susceptibility to fracture is an issue that is seen in escalating numbers as the US population ages. In osteoporotic bones, the rate of Osteoporoticity in cortical bone (and high density region) is remarkably delayed/lesser, compared to cancellous bone[5]. Bone cements are injected in the cancellous bone of the vertebral body when it suffers from compression fractures. The main task of the cement is to transfer the forces of the impact affecting the bone as evenly as possible. The ideal bone cement should be less viscous, should be bioactive, non-toxic, and must have a low exothermal temperature and sufficient mechanical strength.

1.6 **Vertebroplasty**

Vertebroplasty was first introduced in 1987 by Galibert et al.[2]Vertebroplasty is a minimally invasive surgical procedure in which an injured vertebral body is augmented with bone cement, most commonly polymethylmethacrylate (PMMA). It has been further refined and available in the US since 1991.Vertebroplasty is a minimally invasive, non-surgical procedure that is designed to relieve the pain of compression fractures. Vertebroplasty literally means fixing the vertebral body. In addition to relieving pain, those vertebral bodies that are weakened but not yet fractured can be strengthened, thus preventing future problems.
It involves the injection of bone cement into the fractured vertebral body, in order to stabilize the spine, increase mobility and decrease the pain from symptomatic vertebral compression fractures [6]. The bone cement is injected through both the pedicles (Bi-pedicularly) or through a single pedicle (Uni-pedicularly). Vertebroplasty has been successful in the relief of pain caused by compression fractures. However, cement leakage, loss of height and stiffness are a few concerns of this treatment strategy.

1.7 Kyphoplasty

Kyphoplasty is a newer treatment for patients immobilized by the painful vertebral body compression fractures associated with osteoporosis. Like vertebroplasty, kyphoplasty is a minimally invasive procedure that can alleviate up to 90% of the pain caused by compression fractures. In addition to relieving pain, kyphoplasty can also stabilize the fracture, restore height, and reduce deformity. A balloon, referred to as an inflatable bone tamp, is passed though the pedicle into the vertebral body and inflated and cement is injected into the cavity. The creation of the cavity allows for introduction of cement under low pressure and reduces the risk of cement extravasations as seen with vertebroplasty.
1.8 BioCure bone cement

BioCure Inc. is developing a series of more biocompliant bone cements with mechanical properties that match those of different vertebral trabecular or cancellous bones for vertebroplasty and kyphoplasty applications. The elastic modulus and strength of the bone cements are in a range of 5 to 400 MPa, and 5 to 75 MPa, respectively. The bone cement shows excellent thermal and dimensional stability in simulated body fluid of Ringer’s solution at both 37°C and 70°C. The elastic modulus, strength, ultimate strains are constant over time at both temperatures and did not alter dimensions. The bone cement has a much lower exothermic temperature during polymerization compared with commercially available polymethylmethacrylate (PMMA) bone cements, and has high cohesiveness in liquid environments. It is able to be delivered through an 18 gauge needle ranging from 4 to 6 inches in length and reaches final mechanical properties within 30 min post-delivery.

1.9 Scope of this study

The intent of this thesis is to investigate the biomechanical influence of vertebral augmentation on augmented vertebrae and adjacent vertebrae. This study provides comparison of widely accepted polymethylmethacrylate cement with high stiffness to Biocure cement with low stiffness (similar to cancellous bone) thus propose the mechanical efficiency of the Biocure material for the treatment of the vertebral
compression fracture. The mechanical parameters evaluated in this research work are changes in the stiffness, height and motion, intradiscal pressure after augmentation.

The study consisted of two main parts, finite element modeling and experimental testing. The finite element study in this research work indicated the changes in endplate stresses and intradiscal pressure changes in the adjacent levels. Experimental study discusses about the difference in the ease of cement injection, intradiscal pressure change in adjacent levels and osseointegration of the cement. It is hypothesized that the cement stiffness closer to that of cancellous bone will provide the most physiologically normal response and that cements of greater stiffness such as polymethylmethacrylate (PMMA) will lead to increase in the risk of adjacent fractures. In addition to cement stiffness analysis, the effect of cement volume was investigated.

Although long term effect are unknown, fatigue performance of cement augmented spine was performed[7] to look for any introduction of any fatigue cracks developed in cement over the period of time due to continuous compressive and shear forces on cadaveric spine.
Chapter 2

Literature Review

2.1 Overview

This chapter includes a historical review of vertebral augmentation. This review will include osteoporosis, vertebral compression fractures, commonly practiced vertebral augmentation surgical techniques, clinical issues, computational studies, and experimental studies.

2.2 Osteoporosis

Osteoporosis is a bone disease in which the amount of bone is decreased, the structural integrity of trabecular bone is impaired and the cortical bone becomes more porous and thinner. It is one of the major determinants for vertebral compression fracture. Osteoporosis is defined by the World Health Organization (WHO) as either a bone mineral density 2.5 standard deviations below peak bone mass (20-year-old sex-matched healthy person average) as measured by DEXA, or any fragility fracture.
In the USA, about 45% of postmenopausal women have low bone density and about 25% of women above the age of 50 will suffer from vertebral compression fractures (VCF) [8]. An estimated 700,000 osteoporosis-related vertebral compression fractures occur annually, resulting in 150,000 hospitalizations[9]. The economic costs of fragility fractures are immense. According to the National Osteoporosis Foundation, as of 1995, hospital and nursing home expenditures relating to the condition in the United States were greater than $13.8 billion annually, or $38 million a day. The projected cost for 2030 is $60 billion annually. In the UK the annual cost is projected to be £2 billion by 2020, and in Germany the projection is €500 million.
Given the costs, the world should have an interest in this problem. The above figure shows that major impact on osteoporosis is on spine. According to the International Osteoporosis Foundation, 20%–25% of women over 50 have one or more VCFs, and 20% of women who have a VCF are estimated to suffer from another within a year. However, an increasing number of fractures also result from secondary osteoporosis caused by therapeutic drugs such as catabolic steroids, anticonvulsants, cancer chemotherapy and heparin.

### 2.3 Vertebral Compression Fractures:

Compression fractures of the spine occur when a vertebra, the bones of the spinal column cracks, fractures, or collapses. These spinal fractures are unique because they frequently occur without apparent trauma and may cause no symptoms, often going undiagnosed until other complications arise [10]. It is estimated that two-thirds of
Vertebral compression fractures are never diagnosed because many patients dismiss their back pain as a sign of aging and arthritis. Compression fracture of the vertebral body is common, especially in older adults. Osteoporosis accounts for approximately 85% of all vertebral compression fractures (VCFs). Severe fractures can cause significant pain, leading to inability in performing activities of daily living, and life-threatening decline in the elderly patient who already has decreased reserves. Acute fractures occur when the weight of the upper body exceeds the ability of the bone within the vertebral body to support the load. Compression fractures lead to a loss of height of the vertebral segment, and the resulting spinal deformity can lead to a decrease in pulmonary capacity, malnutrition, decreased mobility and depression [11].

Figure 2.3 Bone injuries and its percentage in terms of trauma related and osteoporosis related injuries. Source: PearlDiver Patient Records Database (2004-2006)
In cases of severe osteoporosis, however, the cause of trauma may be simple, such as stepping out of a bathtub. In cases of moderate osteoporosis, more force or trauma is required to create a fracture. A healthy spine can sustain a compression fracture from severe trauma such as an automobile crash or a hard fall. Fig 2.4 shows a typical compression fracture in the spine.

Figure 2.4 (a) Burst fracture creation due to anterior loading (b) x-ray showing burst fracture case.

Without intervention the pain may subside as the fracture heals, the vertebra heals in its deformed, compressed position. This shortened, compressed vertebra alters the normal alignment of the spine, putting the spine at greater risk for subsequent level fractures[12,
When multiple vertebral compression fractures have occurred, there is a significant change in the structure and shape of the spine. This spinal deformation, called kyphosis, gives individuals a hunched-back appearance, often referred to as a “dowager hump.” Moreover, this structural change can affect the internal organs and body functions, negatively impacting the overall health of the individual and their quality of life[14, 15].

A burst fracture is an injury to the spine in which the vertebral body is severely compressed (fig 2.5). They typically occur from severe trauma, such as motor vehicle accident or a fall from a height. With a great deal of force vertically onto the spine, a vertebra may be crushed. With the bony margins spreading out in all directions the spinal

Figure 2.5 (a) Vertebral compression fracture loading mechanism (b) Sagittal reconstruction of a CT scans showing a vertebral compression fracture.
cord is liable to be injured. The bony fragment that is spread out toward the spinal cord can bruise the spinal cord causing paralysis or partial neurological injury. The most susceptible region in the spine for a burst fracture is the thoracolumbar region (T12-L2).

### 2.4 Treatment Options

The treatment for these fractures is primarily non-operative and includes bracing (fig 2.6), external arthrosis (fig 2.6), path medications, physical and medical therapies, psychosocial support, bed rest, pain management, and back school programs.

![Figure 2.6 (a) External Bracing (b) TLSO Spinal Orthoses (c) Kyphotic Deformity](image)

However, elderly people who have chronic pain and develop kyphotic deformities (fig 2.6) need surgical treatment for vertebral compression fractures. Surgical treatments are fraught with complications that are amplified by osteoporosis. Achieving stable
fixation of the spinal implants can be problematic because of the underlying weakened bone. The spinal rods (Fig 2.7-a), hooks (Fig 2.7-b) and screws (Fig 2.7-c) may become dislodged from the bone, resulting in loss of fixation and deformity correction. Also, the fixed levels may cause altered kinematics of the spine and can place adjacent vertebral bodies at higher risk for VCFs.

![Spinal rod, Spinal hook, Spinal screw](image)

**Figure 2.7** (a) Spinal rod (b) Spinal hook (c) Spinal screw

### 2.5 Pain relief Mechanism:

The most important aim for any augmentation procedure is the pain relief. Injection of bone cement is believed to provide pain relief over two decades. However, accrual pain relief mechanism is not clear yet. Injection of cement mechanically reinforces the vertebral body and may eliminate pain by halting further compression or micro motion of fracture site. Alternative mechanism could be palliative effect of heat generated by the
exothermic polymerization of PMMA by damaging pain-sensitive nerve endings within the vertebral body.

2.6 Surgical procedure of treatment

Since its introduction in the mid 1980’s vertebroplasty has evolved in many aspects. Advances have come in the way of different materials (cements) to inject into the vertebral body, preparations of the vertebral body for injection, and cement containment devices. This section will discuss the original vertebroplasty procedure and then some of the major changes that have taken shape since.

The initial percutaneous, or through the skin, technique of treating osteoporotic vertebral compression fractures (VCF) has evolved from the use of polymethylmethacrylate (PMMA). Galibert’s initial goal of the techniques was to treat pain in patients with spinal tumors and hemangiomas[2]. Basically, PMMA was injected into the vertebral body under radiological guidance using a percutaneous approach.

2.6.1 Vertebroplasty

The technique of vertebroplasty begins with the placement of an 11 to 15-gauge needle into the involved vertebral body. The patient is typically in the prone position on an appropriate table for CT scan and/or fluoroscopic guidance. Usually local anesthesia and IV sedation are used, although general anesthesia may be needed if the patient is unable to lie in the prone position. The typical procedure is performed through a 1-cm
paramedian incision leading to a transpedicular approach to the vertebral body. A special mix of PMMA is used with increased radiopacity so that the cement can be seen more easily with radiological guidance during and after the procedure. The PMMA is injected into the vertebral body under pressure, but usually not enough pressure to restore or change vertebral height. The posterior vertebral cortex must be intact during vertebroplasty in order to avoid cement leakage into the spinal canal.

Figure 2.8 Vertebroplasty procedure and x-ray showing vertebral augmentation using vertebroplasty.

2.6.2 Kyphoplasty

Kyphoplasty grew out of vertebroplasty and follows a similar procedure with an additional step illustrated in figure 2-9. A tract for a larger gauge needle is drilled into
the vertebral body through the pedicle followed by an inflatable balloon (bone tamp). The balloon serves to create a cavity when inflated. The balloon inflation step can re-expand the vertebral body and thus, increase the height of the vertebral body. The balloon is then deflated and withdrawn from the vertebral body prior to the injection of the PMMA cement. The use of a large gauge needle and the creation of the cavity in the vertebral body allows for a more viscous PMMA mixture to be injected under relatively low pressure. This increased viscosity cement and low-pressure injection has the advantage of reducing cement leakage. The average volume of the injected PMMA is about 7 cc. Kyphoplasty is typically done bilaterally for each vertebral body fracture. As many as 6 levels have been done at once, although usually just 1 or 2 vertebral bodies undergo the procedure during a single setting.

Figure 2.9 kyphoplasty procedure and x-ray showing vertebral augmentation using kyphoplasty.
2.6.3 Optimesh

Optimesh is another step in the evolution of vertebral augmentation. Optimesh implants are three-dimensional mesh pouches constructed of polyethylene terephthalate thread. Each mesh includes a main body, which is filled with a bone slurry rather than PMMA, and a neck through which the bone passes. The mesh provides containment of the granular bone graft allowing the formation of a graft pack. Nominal mesh pore size is 1500 microns to permit graft incorporation through the mesh pores [16]. The procedure involves creating a cavity in the vertebral body with a drill and shaper, rather than a balloon as in kyphoplasty. The mesh is then inserted into the void created by the surgeon and is filled with the bone slurry. This system aims to solve many of the clinical problems facing vertebroplasty with the use of a containment device to stop leakage and a bone graft rather than PMMA to promote new bone growth, maintain bone material
properties to reduce unwanted changes to the biomechanics, and avoid temperature increases due to cement curing.

2.7 New Devices for Vertebral augmentation

![Implant insertion](image1.png) ![Implant Expansion](image2.png)

Figure 2.11 Vexim’s spine jack implant and x-ray showing its implantation.

Vexim's SpineJack implant for treating vertebral compression fractures looks and works much like a car jack. Once inserted into a patient's vertebra through a minimally invasive procedure, the device ratchets up to restore an anatomically correct profile. SpineJack maintains height restoration until the load bearing is transferred to bone cement. So rather than merely treating back pain, the implant also restores the anatomical shape of the vertebra. Considering anatomical restoration, height is not the only factor that should be taken into consideration. Cranio-caudal expansion to restore sagittal local angulations, adoption of implant’s expansion to restore coronal angulations along with adoption of implant’s positioning for endplate need to be considered. Pre and Post-op CT scans showed height restoration after implantation of such devices. Once the height
restoration is achieved cement injection need to be performed to maintain height and strengthen the bone. In May 2008 the company received the CE mark for the SpineJack system. Vexim is pursuing the 510k path to FDA approval in US. In July 2008, the company began enrolling in a multicentre post-market clinical study. In June 2006, Vexim received an investment of €3 million from Truffle Venture (Paris) in Series A financing which shows a promising technology.

![Figure 2.12 Alphatec’sosseofix implants and x-ray showing its implantation.](image)

Consider the OsseoFix Spinal Fracture Reduction System for use in the T1-L5 region, which is used to treat VCFs. Benefits of the technology include Creation of bony channels during implant deployment and retention of bone within the implant promotes cement interdigitation. Implant construct has potential to provide enhanced ability for vertebral height maintenance. Controlled implant deployment delivers reproducible results. Deployed implant is designed to maintain fracture reduction throughout cement delivery. This last point highlights an important feature of the product. In standard vertebroplasty, vertebral height restoration can decline as the cement cures. However,
with Alphatec’s technology, the implant allows for the vertebral height to be maintained during the procedure.

Additionally, Alphatec offers OsseoFix+ Radiopaque Bone Cement for use with the OsseoFix technology. According to the company, OsseoFix is a self-hardening, ready to use, medium-viscosity bone cement. OsseoFix+ contains a high percentage of radiopaque agents (45%) for maximum visibility and radiological control during a percutaneous vertebroplasty procedure. In 2008, the company announced it had received a European CE mark for the device, allowing it to market the device in the European Union to treat vertebral compression fractures. OsseoFix is currently not available for sale in the U.S. The company is beginning a clinical study to support its 510k application.

Figure 2.13 Staxx FX device implantation for vertebral height restoration. (Source: http://www.spinewave.com/products/fx_us.html)

The Spine Wave, Inc. Staxx system was designed to treat VCFs in a minimally invasive fashion as an alternative to traditional vertebroplasty and kyphoplasty. The
device is designed to allow for controlled fracture reduction in 1mm increments. The company cites the following benefits:

- No intra-operative reduction loss
- Controlled vertical expansion in 1mm increments
- Barrier to posterior extravasation
- Reduced bone cement volume

The device is composed of stackable, 1mm PEEK wafers, which are inserted into the vertebral body one at a time in order to provide for the appropriate height restoration. The system is designed to be used with PMMA cement. Conceptually the treatment method looks promising.

![Image of Kivaplasty with vertebral height restoration.](image)

The Kivaplasty technique utilizes a unilateral (singlestick), transpedicular approach to access the vertebral body, which is expected to reduce access related risks. It also creates an internal scaffold designed to reduce the fracture and maintain the restored height of the vertebral body and provides a structural repair with a material that is mechanically
similar to natural bone, thereby offering the potential to reduce adjacent level fractures. This system is designed to facilitate cement direction and containment, reduce the risk of extravasations, and decrease the amount of bone cement required to stabilize the fracture.

2.8 VCF Treatment Moving Forward

We hear about potentially disruptive technologies and how they are going to be game changers. Often, they either take a decade longer than anticipated to make an impact, or make no impact at all. What makes the technologies above so interesting is that they are going to hit the market soon or are already approved. So the question is not if, but when. Here are potential trends in these devices.

- A movement toward devices that allow for controlled/adjustable height restoration
- A focus on devices that decrease the potential for cement leakage
- A trend toward adopting technologies that can be offered at a discount to kyphoplasty
- Devices that allow for increased control over cement viscosity and lower-pressure injection
- Devices that maintain vertebral height while cement cures preventing further damage to the vertebral body.

In future, orthopedic companies may collaborate with pharmaceutical companies that have prophylactic treatments, decreasing the risk of subsequent or adjacent level
fractures. Products such as Amgen’s denosumab could literally decrease the incidence of osteoporosis related fractures. A focus on preventative medicine and prophylactic treatments could shift former fracture repair related revenues to pharma/biotech companies. Companies developing innovative treatments to treat compression fractures s know that VCF still stands for Very Compelling Future!

2.9 Clinical Issues

Considering the effectiveness and widespread use of vertebral augmentation there are some serious clinical issues. The exact source of pain in patients with vertebral compression is still not known. This makes it difficult to say what aspect of vertebral augmentation provides the pain relief. In all likelihood it is a combination of things such as stabilization of macro and micro level fractures, destruction of nerve endings, and restoration of the anatomical boney structure of the vertebral body. In addition to not knowing the exact source of pain or the pain relieving element of the procedure there have issues with procedural error, cement leakage, cement curing temperatures, and adjacent level fractures. Procedural errors include infection, fracture of the transverse processes, fracture of the ribs, fracture of the pedicle, and respiratory distress due to anesthetics [17-23]. These errors are typically attributed to the experience of the surgeon.
2.9.1 Cement Leakage:

Cement leakage can be a serious complication. Leakage can occur through cracks or punctures in the vertebral cortex or through the veins in the vertebral body. Theoretically, the presence of cement in the disk would be expected to increase the apparent stiffness of the disk and thereby alter the normal spine kinematics at the injected level. A review of clinical data including 2283 vertebrae augmented by vertebroplasty and 1487 vertebrae augmented by kyphoplasty found that leakage occurred in 41% and 9% for vertebroplasty and kyphoplasty, respectively. Most leakage cases were asymptomatic, clinical complications occurred in 2.6% and 1.3% of the vertebrae augmented by vertebroplasty and kyphoplasty respectively [24]. Although the majority of leakage is asymptomatic the long term effects of them are unknown. Cement leakage is a function of cement viscosity too. So all these results are based on pressure of cement injection and higher rate of leakage might be due to high cement injection pressure. Study by Lin et al [25] reported that 58% of vertebral bodies adjacent to an intradiscal leak fractured during follow-up compared with only 12% of vertebral bodies adjacent to augmented vertebrae in which no intradiscal leakage occurred.

Cement leak into the spinal canal can induce nerve pressure and damage, typically requiring immediate medical attention. The symptomatic neural canal leaks have dramatic consequences, such as paraplegia and death. Neurologic complications described as radiculopathy, a worsening of pain or spinal compression or injury, have
been reported in seven vertebroplasty cases [18, 26-31] and four kyphoplasty cases [32-35]. Other major neurologic complications following vertebroplasty have been documented, including paraparesis, spinal claudication, and paraplegia [36, 37].

Cement leakage can be caused by many different variables such as injection pressure; volumes of cement injected, and cement viscosity. It is believed that creating a cavity and using higher viscosity cement in kyphoplasty reduces the risk of cement leakage. In a study by Berlemann et al it was noted that when the attempt was made to fill the bone around the cavity made during kyphoplasty rather than just filling the cavity leakage occurred 33% of the time, similar to that of vertebroplasty leakage rates [38]. When only the cavity was filled, leakage occurred 10% of the time [39]. The Optimesh system has resolved the leakage issue with its cement containing pouch, but it has not yet gained widespread use.

2.9.2 Temperature elevation:

The curing temperature of PMMA is believed to cause thermal necrosis, or the killing of neurons due to excessive heat. Temperatures above 50°C are known to cause thermal necrosis. Experimental studies have been done to measure the temperatures reached during curing. Belkoff et al [40] recorded peak temperatures from 70°C to 114°C in the anterior and center regions of vertebral bodies using two brands of PMMA at 6 ml and 10 ml. The posterior region never exceeded 50°C. A study by Deramond et al [41] found the mean peak temperature in the center of the vertebral body to 61.8°C and 51.2°C for 10 cc
injections of Simplex and Orthocomp respectively. Once again the temperatures in the neural canal did not reach 50°C. This evidence shows that intervertebral nerve tissue is at risk for damage and it was believed that this may be one aspect of pain relief. Since then, augmentation materials that do not have a thermal reaction have been used with similar pain relieving success indicating that the thermal necrosis caused by the curing temperature of PMMA is not the main pain relieving attribute of vertebral augmentation. Though curing temperatures measured in the lab did not exceed 50°C in the neural canal in a laboratory setting, it would be assumed that in the cases of cement leakage into or near the neural canal would have very different results. There should be a push for use of cements without the thermal reaction prevalent in the use of PMMA.

2.9.3 Adjacent level fractures:

A much debated issue with vertebroplasty has been recurrent vertebral compression fractures adjacent to augmented vertebrae. Vertebral augmentation has been established as an effective technique for pain relief, stabilization, and strength restoration of vertebral compression fractures. Pain relief has been attributed to the stabilization of micro and macro fractures as well as the destruction of nerve endings by the heat given off while the bone cement cures. Even with the success vertebral augmentation has realized some questions still remain unanswered. The effects of vertebral augmentation on segments adjacent to the augmented vertebrae are not well documented/understood [42-44]. Many patients have experienced recurrent fractures in adjacent vertebrae following vertebral
augmentation, and there has not been a clinical control trial to prove or disprove that vertebroplasty increases the risk of these adjacent fractures. The incidence of recurrent fracture after kyphoplasty is substantial at 10% within the first 90 days [45]. The Kaplan–Meier curve showed that 7.8% of the patients would experience new symptomatic osteoporotic vertebral compression fractures within 1 year after initial vertebroplasty[43]. Grados et al. [26] reported that the relative risk of VCF was 1.44 in vertebrae adjacent to VCFs not treated with PVP and 2.27 in vertebrae adjacent to VCFs treated with PVP. Lee et al [43] reported that During a follow up period, 38 (15.6%) of the 244 patients returned for subsequent intervention due to the development of newly developed symptomatic osteoporotic vertebral compression fractures. Furthermore, Trout et al [46] have shown that fractures occurring adjacent to augmented vertebrae are more like to occur in the superior endplate when the incident fracture occurs cephalad to the augmented vertebrae and in the inferior endplate when the incident fracture occurs caudal to the augmented vertebrae. In other words, when recurrent fractures occur in the adjacent vertebrae they are originating at the endplates adjacent to the augmented vertebrae. The original compression fractures are more commonly seen in the superior endplate, making the increased number of fractures in the inferior endplate caudal to the augmented vertebrae a reason to believe that vertebral augmentation may lead to an increased risk of adjacent fractures. This point makes it difficult to place blame of these recurrent fractures purely on natural progression of osteoporosis. It needs to be determined if vertebral augmentation has an effect on load distribution, leading to an
increase in risk of adjacent fractures. More data on load and stress transfer following vertebroplasty may assist in understanding the causes of recurrent vertebral compression fractures adjacent to augmented vertebrae.

2.9.4 Injection Pressure:

An important limitation of vertebroplasty is the excessive pressure required to inject sufficient cement [47]. The pressure applied on the plunger of the syringe during vertebroplasty can often exceed 1500 kPa which is approximately 170 N of force, or 17 kgs. Sometimes the pressure requirements are so excessive that the procedure needs to be aborted. This early termination may result in insufficient filling, which may impact the success of the procedure. Two methods have emerged as a method to overcome this problem: either the cement viscosity should be lowered or the pressure applied should be increased. Increase in the cement viscosity leads to complications such as toxicity and reduced cement strength and the risk of leakage. The viscosity of PMMA cement is comparatively less than that of mineral cements such as calcium phosphate cements [48]. This is one of the reasons for the popular use of PMMA to treat fractures.

2.9.5 Effect of volume fill on mechanical properties:

There is a correlation between cement volume and the resulting mechanical properties such as strength and stiffness [49]. The volume of cement needed to restore the strength and stiffness of fractured vertebral bodies as a function of level is unknown. A study
carried out by Belkoff et al[49] studied the association between cement volume and mechanical properties. Their results indicated that the strength was restored for all regions when 2 ml of cement was injected into the vertebral bodies. Stiffness restoration was weakly associated with cement volume. In the thoracic and thoracolumbar region, 4 ml of cement restored stiffness but 6 ml was needed to restore stiffness in the lumbar region. Restoring the strength would cause the adjacent vertebral bodies to fracture before the augmented vertebral body if loaded to the force that caused the earlier fracture. Post-treatment stiffness is probably responsible for pain relief, as it internally stabilizes the body preventing micromotion and subsequent pain. Very high stiffness will prevent the natural healing process by bearing the entire load and preventing any mechanical stimulus.

On average fills of 16% and 29.8% of the vertebral body volume was needed for strength and stiffness restoration respectively. Thus it is unlikely that routine estimation or measurement of vertebral body volume before PVP to calculate the amount of cement needed for injection is clinically indicated. There is no consensus yet on the correct fill volume, though a volume between 2-8 ml is generally acceptable depending on the VB volume. Biomechanical restoration of a vertebral body depends on the volume of cement injected, the fracture type being repaired (crush, wedge or endplate fracture) and the loading environment the vertebral body is subjected to (loading type, magnitude, direction and rate).
2.10 Bone Cements

Various types of bone cements are commercially available for the use of vertebroplasty or kyphoplasty procedures. There has been continuous effort towards development of bone cement towards optimizing mechanical as well as biological properties. Ideal bone cement needs to be less viscous for ease of injection, non-toxic, less exothermic temperature, bioactive and ability to osseointegrate, sufficient mechanical strength and radiopaque properties.

The first clinical use of PMMA was attempted in 1938 and dough was produced by mixing ground polymethylmethacrylate (PMMA) powder and a liquid monomer. Even today PMMA is widely used bone cement with good clinical data supporting its short and long term results. Even though PMMA is choice for many surgeons, every year new bone cements are introduced to overcome disadvantages of PMMA. PMMA is not bioactive bone cement and hence it does not contribute to osseointegration and might interfere with natural remodeling process of the bone inserted. Also there is a large mismatch in the compressive strength of PMMA compared to osteoporotic bone. The monomer in PMMA is toxic so possibility of foreign body reaction, necrosis.

Hydroxyapatite-reinforced PMMA: Hydroxyapatite is a ceramic which has been shown to induce bone formation and has been used in a number of successful implant applications. Addition of HA to PMMA do increase its flexural modulus and fracture
toughness and reduce potentially harmful heat generated during the polymerization of PMMA based cements.

Calcium phosphate cements: Calcium phosphate based cements are resorbable. The composition of this cement is very similar to composition of bone which could enhance local bone regeneration and remodeling through osteoblastic activity. It can be suggested as a good alternative to PMMA cement though study by Lim et al [48] found that the stiffness decreases after cement injection compared to PMMA.

2.11 Computational Studies

Past computational studies have investigated the effects of vertebral augmentation in single motion segments and have not provided information on effects in the posterior column. The major focal point in past works has been intradiscal pressure and endplate stress and strain [50-55]. Most computational studies exploring the effects of vertebral augmentation on spine biomechanics have used single motion segment or single vertebrae models. Models have been altered to mimic osteoporotic and fractured spines. The consensus among these studies seems to be that vertebral augmentation does alter the load transfer. Polikeit et al [52] predicted increases in intradiscal pressure and adjacent endplate bulge in an L2-L3 osteoporotic finite element model. Wilcox et al [51] had similar results, concluding that rigid cement augmentation leads to increased deformation in the adjacent endplate in L2-L3 osteoporotic vertebral compression fracture finite
element model. Baroud et al [54] also saw increases in the intra-discal pressure and adjacent endplate deformation in a L4-L5 (no posterior bone) finite element model. All of these results indicate that increasing the stiffness of the vertebral body leads to an increase stress in the adjacent vertebral endplate. Increased stresses in the endplate means there must be greater load transfer through the disc and endplate. None of these studies have tried to explain why there is an increase in load transfer through the vertebral body and disc beyond the fact that the endplate of the augmented vertebrae will not bulge as much after augmentation. The total load transfer through the entire spine should not change, as the same loads are being applied. This means that an increase in load in one region of the spine should cause a decrease in load in another region of the spine.

2.12 Experimental Studies

The experimental work that has been done in vertebroplasty has includes stiffness comparisons of single vertebrae and motion segments, as well intradiscal pressure and vertebral cortex strain measurements [53, 56-62]. As discussed in the previous section, computational work had concentrated on endplate stress and intradiscal pressure. These parameters have proven to be difficult to obtain in a laboratory setting. Intradiscal pressure measurements have been performed in some studies [57, 63], but it is difficult to say whether or not they provide an accurate picture of physiological disc. A number of things could affect the results of intradiscal pressure measurements. The intervertebral
disc is designed to hold fluid to use as a load distributor. A cadaveric specimen is not likely to have the same fluid or nutrition levels of an in vivo intervertebral disc. The damage that can be done while inserting a pressure measuring device and placement of the device makes it difficult to put a lot of faith in any such measurements. No studies have attempted to measure adjacent endplate stress in vitro, as this would be even more difficult to measure. Putting these factors aside, these studies did show significant increases in intradiscal pressure following vertebral augmentation.

Figure 2.15: Dynamic compression fatigue testing of vertebral segment after cement augmentation.

The evaluation of the stiffness of osteoporotic vertebral bodies before and after vertebral augmentation has shown that the stiffness and ultimate strength of augmented vertebral bodies are twelve and thirty-six times greater than osteoporotic vertebral bodies respectively [56]. The consequences of this change in stiffness and strength are believed to be the increase in recurrent adjacent vertebral compression fractures. Berlemann et al
[58] found that a treated vertebral motion segment when compared to untreated segments had a 17% decrease in ultimate strength due to adjacent vertebral failure. All of this experimental work has supported the notion that increase in vertebrae stiffness by vertebroplasty can lead to an increased risk of adjacent fractures.
Chapter 3

Materials and Methods

Chapter 3 starts with an introduction to the computational portion of this study including the advantages of using the finite element method. It will be followed by a detailed description of the intact T12-L3 model, validation of the model, and the alterations made to simulate vertebral augmentation. This study also includes one vertebra model and then we will discuss about each type of the models implemented and the respective methods those were used to analyze the results will be discussed.

In the later part, this chapter describes the experimental set up along with the material and methodology used for a cadaveric study to evaluate the kinematic and kinetic (intradiscal pressures) behavior of two types of cement augmentation into cadaveric lumbar spines using the follower load test protocol will also be explained.

Cyclic complex fatigue testing was performed additionally followed by non-destructive as well as destructive analysis of augmented vertebra for any fatigue cracks in cement and endplate.
3.1 Introduction

The aim of this study was to quantify the loading at a macro level, by identifying the changes in loads through the facets and stresses in the endplates at different cement stiffness and fills volumes. This information helped to determining if vertebral augmentation increases the risk of adjacent fracture and also in identifying optimal cement stiffness and fill volume for the procedure. Our null hypothesis states that increasing stiffness of cement as well as an increase in augmentation volume will cause higher stress levels in the adjacent endplates as well as unnatural load sharing in the augmented vertebral body.

Experimental studies helped us in studying the anatomical variability with more realistic approach by applying the physiological loads and other simulating conditions. Osteoporotic vertebral segments were considered in experimental studies and aim was to look at the altered motion due to inappropriate interdigitation of cement with cancellous bone while correcting fracture and study the percentage increase in disc pressure with two different stiffness cement used for augmentation at various volumes. Fatigue study helped in simulating human motion in an year after fracture fixation using augmentation method and study the endplate damage due to stiffer cement. Also fatigue loading helped to study the fatigue performance of cement inside the vertebra experiencing continuous loading.
3.2 Finite Element Analysis

To accomplish this, the finite element method was utilized. The finite element method has been used to model biological systems as early as 1969 [64]. Belytschko et al [65] first used the finite element method for spine biomechanics modeling in 1974. The irregular geometry and composite nature of the vertebrae and intervertebral discs make the spinal column an ideal FE modeling candidate. The FE method provides a means to quantitatively assess the role of the individual spinal components in load distribution or stress distribution, which is difficult to measure experimentally. There are still many outstanding questions of whether vertebral augmentation increases the risk of recurrent fractures. The finite element method allows us to evaluate how vertebral augmentation alters the load and stress distribution throughout the functional spinal unit. This provides a cost effective tool to test different augmentation methods and materials before proceeding with other experimental tests.

3.3 Thoracolumbar Finite Element Model

A finite element model of the intact ligamentous T12-L3 spine segment was developed, Figure 3-1. The commercial software ABAQUS 6.8 was used to construct and analyze the model. The geometric data for the T12 – L3 motion segment was obtained from computed tomography (CT) scans of a cadaveric ligamentous spine specimen using transverse slices of 1.5 mm thickness. The cadaver spine was free of deformities and
abnormalities, including severe degeneration. Material properties were selected from the literature. Sequentially stacked, digitized cross-sectional data provided the means to generate the model. The model consists of 39,764 nodes and 31,768 elements, Table 3-1. The material properties assigned to the various components in the model are shown in Table 1. The intact models motion response was validated against the literature. Elements in this model were classified as C3D8 or hexagonal elements meaning each element having 8 nodes and each node possesses 3 degrees of freedom.

Figure 3.1 T12-L3 Finite Element Model
3.3.1 Vertebral body and posterior bone modeling:

The vertebral body and posterior bony regions were defined using three-dimensional hexagonal elements. The vertebral bodies have been modeled as a cancellous (porous) bone core surrounded by a 0.5 mm thick cortical (dense) bone shell. The appropriate isotropic material (refer Table 3.1) properties were defined for the respective regions. The vertebral body and posterior elements were made with three-dimensional solid continuum elements with eight nodes, with each node possessing six degrees of freedom.

3.3.2 Intervertebral Disc

The intervertebral disc is made up of two sections, the annulus fibrosis and nucleus pulposus. The annulus fibrosis is modeled as a composite, a series of fiber bands or lamellae embedded within a ground substance. Each circumferential layer of ground substance has two fiber layers in alternating angles of ± 30°. The annulus fibrosis was made of eight layers, consisting of a collagenous fiber content of 16%. The fibers were modeled with REBAR or reinforcing elements only acting in tension. The ground substance was modeled as a continuum. The material properties of these elements can be found in table 3-1. The nucleus pulposus is modeled as the core of the annulus fibrosis. To give the nucleus pulposus hydrostatic characteristics it was given an elastic modulus of 1 MPa and a Poisson’s ratio close 0.5.
3.3.3 Apophyseal (Facet) Joint

The apophyseal joints were modeled with GAPUNI elements that support uni-directional loads. Each joint was modeled with 20 gap elements simulating the physiologic nature of the cartilaginous layer lining the articular surface. Based on CT images the initial gap distance was modeled as 0.5 mm. As the load transferred through the facets increases the gap closes. A softened interaction allows the elastic modulus to reach that of the posterior bone as gap closes. The facets are oriented at approximately 72° to the horizontal plane.

3.3.4 Ligaments

When modeling the ligaments of the functional spinal unit emphasis must be given to the fact that ligament strain predictions are dependent on original lengths, direction, point of attachment, initial laxity, and realistic modeling of the posterior bone. Ligaments become progressively stiffer as they are stretched. This property was modeled with two noded truss elements assigned hypoelastic material properties allowing for the non-linear behavior. The cross-sectional areas of the various ligaments were assigned based on literature and are listed along with material properties in table 3-1. With exception to the Ligamentum flavum and longitudinal ligaments all ligaments were assumed to be initially unstressed.
3.3.5 Assignment of the material properties

Table 3.1 Material properties assignment and element type of intact T12-L3 Finite element model.

<table>
<thead>
<tr>
<th>Element Set</th>
<th>Number of Elements</th>
<th>ABAQUS Element Library Type</th>
<th>Modulus of Elasticity (MPa)</th>
<th>Poisson’s Ratio; η</th>
</tr>
</thead>
<tbody>
<tr>
<td>Cortical Bone</td>
<td>2976</td>
<td>C3D8</td>
<td>12000</td>
<td>0.30</td>
</tr>
<tr>
<td>Cancellous Bone</td>
<td>9984</td>
<td>C3D8</td>
<td>100</td>
<td>0.20</td>
</tr>
<tr>
<td>Posterior Bone</td>
<td>4934</td>
<td>C3D8</td>
<td>3500</td>
<td>0.25</td>
</tr>
<tr>
<td>Annulus (Ground Substance)</td>
<td>3384</td>
<td>C3D8</td>
<td>1.2</td>
<td>0.45</td>
</tr>
<tr>
<td>Annulus Fibers</td>
<td></td>
<td>REBAR</td>
<td>550</td>
<td>0.30</td>
</tr>
<tr>
<td>Nucleus Pulposus</td>
<td>1792</td>
<td>C3D8</td>
<td>1.0</td>
<td>0.4999</td>
</tr>
<tr>
<td>Apophyseal Joints</td>
<td>52</td>
<td>GAPUNI</td>
<td>Softened, 3500</td>
<td></td>
</tr>
<tr>
<td>Anterior Longitudinal</td>
<td>180</td>
<td>T3D2</td>
<td>15.6, 17.8, 20.0</td>
<td>0.30</td>
</tr>
<tr>
<td>Posterior Longitudinal</td>
<td>96</td>
<td>T3D2</td>
<td>5.8, 10.0, 20.0</td>
<td>0.30</td>
</tr>
<tr>
<td>Transverse</td>
<td>20</td>
<td>T3D2</td>
<td>6.0, 12.0, 59.0</td>
<td>0.30</td>
</tr>
<tr>
<td>Ligamentum Flavum</td>
<td>16</td>
<td>T3D2</td>
<td>8.5, 13.0, 19.5</td>
<td>0.30</td>
</tr>
<tr>
<td>Interspinous</td>
<td>28</td>
<td>T3D2</td>
<td>7.29, 12.0</td>
<td>0.30</td>
</tr>
<tr>
<td>Supraspinous</td>
<td>8</td>
<td>T3D2</td>
<td>4.2, 8.8, 15.0</td>
<td>0.30</td>
</tr>
<tr>
<td>Capsular</td>
<td>2</td>
<td>T3D2</td>
<td>4.9, 8.48, 32.9</td>
<td>0.30</td>
</tr>
</tbody>
</table>

The material properties of the intact T12-L3 model are summarized in table 3.1. All materials were modeled as homogenous and isotropic material with exception of the annulus fibrosis as discussed earlier. The ligaments were assigned hypoelastic properties by defining elastic moduli for corresponding strain values resulting in a nonlinear material behavior.
3.3.6 Boundary and Loading Conditions

The inferior most surface or the L3 inferior endplate was fully constrained. Loads were applied at the superior most surface or the T12 superior endplate. This allowed for the upper three vertebrae to move without constraint. Different loads were applied to the model and compared to the literature for validation of the motion response. All loads were distributed over the entire T12 superior endplate to ensure proper distribution. White and Panjabi demonstrated that under the action of a load in one direction, the relative displacement between the joint members in the direction of the load is coupled with displacements in the other directions [66]. For the most part, displacement analyses were restricted to the direction of the applied loading modality.

3.4 Model Validation

Validation of a finite element model is important part of using such analysis techniques. Validation of the model shows that the robustness and assumptions made while making the model were appropriate and provide realistic results and justify the conclusions made from the modeling results. The results obtained when modeling the spine rely not only on appropriate geometrical and material definitions, but also on a range of physiologic motions. The load-displacement data generated from the model was compared to past experimental and computational data [67]. These comparisons can be seen in figures 3-2 through 3-4. Evaluation of these comparisons show that the model
predicted load displacement values similar to past experimental and computational studies. At this point we proceeded to develop osteoporotic, fracture, and vertebral augmentation models. Once these modifications were complete, further comparisons were made to other vertebral augmentation studies. These comparisons will be discussed later.

Figure 3.2 Finite element segmental motion validation with existed data in flexion/Extension motion at the applied moments
Figure 3.3 Finite element segmental motion validation with existed data in lateral bending motion at the applied moments

Figure 3.4 Finite element segmental motion validation with existed data in axial rotation motion at the applied moments
3.5 Osteoporotic T12-L3 Model

Vertebral compression fractures commonly occur in people with osteoporosis. To add this factor into the model the Young’s modulus of the cortical, cancellous, and posterior bone was reduced. Reduction of the Young’s modulus reduces the stiffness of the bone and alters the load transfer. Polikeit et al [52] established this modulus reduction technique in a L1-L2 finite element model. Table 3-2 lists the reductions made to the intact model.

Table 3.2 Material properties assigned for osteoporotic ligamentous thoracolumbar spine model.

<table>
<thead>
<tr>
<th>Spine</th>
<th>Young's Modulus (MPa)</th>
<th>Poisson's Ratio</th>
</tr>
</thead>
<tbody>
<tr>
<td>Cortical Bone</td>
<td>8040</td>
<td>0.3</td>
</tr>
<tr>
<td>Cancellous Bone</td>
<td>34</td>
<td>0.2</td>
</tr>
<tr>
<td>Posterior Bone</td>
<td>2345</td>
<td>0.25</td>
</tr>
<tr>
<td>Annulus (ground)</td>
<td>4.2</td>
<td>0.45</td>
</tr>
<tr>
<td>Annulus (fiber)</td>
<td>175</td>
<td>×</td>
</tr>
<tr>
<td>Nucleus Pulposus</td>
<td>1</td>
<td>0.499</td>
</tr>
<tr>
<td>Ligamentum Flavum</td>
<td>10.0(&lt;11%), 20.0(&gt;11%)</td>
<td>0.3</td>
</tr>
<tr>
<td>Transverse Ligament</td>
<td>15.0(&lt;6.2%), 19.5(&gt;6.2%)</td>
<td>0.3</td>
</tr>
<tr>
<td>Capsular Ligament</td>
<td>10.0(&lt;18%), 58.7(&gt;18%)</td>
<td>0.3</td>
</tr>
<tr>
<td>Interspinous Ligament</td>
<td>7.5(&lt;25%), 32.9(&gt;25%)</td>
<td>0.3</td>
</tr>
<tr>
<td>Supraspinous Ligament</td>
<td>8.0(&lt;20%), 15.0(&gt;20%)</td>
<td>0.3</td>
</tr>
</tbody>
</table>
3.6  Mesh Edit for Posteo-lateral Vertebral Augmentation

The mesh of the L1 vertebrae was altered to simulate a cylinder. The cylinder consisted of six circumferential rings and ten layers of elements laterally. To accomplish this modification thousands of nodes would have to be moved within the existing mesh. A mathematical algorithm was developed in Microsoft Excel to move the nodes along a predetermined radius to create each ring. The nodes were moved in each layer separately. The new node positions were then put into the model. This method of creating a vertebral augmentation provides a cylindrical shape for the cement while maintaining a continuous material. Each ring of the cylinder was assigned an element set. This allowed for simulations of differing cement volumes with relative ease.

3.7  Fracture Simulation

Vertebral compression fractures are defined as a collapse of the anterior column of the vertebral body. The fractures are typically attributed to the onset of osteoporosis. To simulate a fracture in the finite element model a couple of methods were attempted. In the first method some elements were removed from the anterior region of the vertebral body to reduce the stiffness of the vertebra but this method did not result in a satisfactory change to the model. The endplate stresses did not show any increase when model was compressed with 1 KN compression load. The second method was attempted by reducing the Young’s modulus of two rows of elements in anterior region of the vertebral body.
and in addition some elements were removed from the anterior region of the vertebral body which resulted in wedge shape fracture effect. The fracture created by Wilcox et al was similar to our model.

![Fracture Simulation](image)

**Figure 3.5** Fracture simulation in validated finite element T12-L3 Model

### 3.8 Simulation of Vertebral Augmentation

To simulate a L1 vertebral augmentation the nodes of the elements of vertebral body were edited to for concentric oval cylinder. Oval cylindrical shape was selected to mimic the cement placement in the vertebra when augmented. Material properties of cylinders were selected according to cement selection. Appropriate numbers of cylinders were
chosen for 6cc of cement injection and properties assigned for Biocure and PMMA cement as per table 3.3. If elements with reduced (fractured) properties were not within the cylinder there were left as it is. Each volume / material property combination was subjected to five loadings. All loads were applied at the T12 superior endplate and the model was fixed at L3 inferior endplate and posterior bone. Single vertebra model was generated from the T12-L3 model keeping L1 vertebra intact. All other disc and adjacent levels were removed.

Figure 3.6 Vertebral augmentation simulation in validated finite element T12-L3 Model


Table 3.3 Material properties assigned to Bone cement for vertebral augmentation simulation in single vertebra model as well as thoracolumbar model.

<table>
<thead>
<tr>
<th>Bone Cement</th>
<th>Young’s Modulus (MPa)</th>
<th>Poisson’s Ration</th>
</tr>
</thead>
<tbody>
<tr>
<td>Biocure Bone Cement</td>
<td>100</td>
<td>0.41</td>
</tr>
<tr>
<td>PMMA Bone Cement</td>
<td>2200</td>
<td>0.41</td>
</tr>
</tbody>
</table>

3.9 Data Analysis

In this study facet loads, endplate stress and deformation, intra-discal pressure, and load distribution throughout the augmented vertebral body were analyzed to try and explain what is happening to the load transfer. A better understanding of the load transfer that occurs may provide valuable insight into how vertebral augmentation is so effective in providing pain relief and whether or not vertebral augmentation is responsible for adjacent level fractures.

To evaluate load transfer the maximum Von Mises stress, total Von Mises stress, minimum principal strain, total principal strain, and facet loads were analyzed. The stress and strain contour plots were also evaluated. Single vertebra model was also generated using the thoracolumbar model by taking out all the ligaments, disc and facet which then simulated with and without fracture.
3.10 Experimental Testing

The in-vitro study consisted of investigation of the Biocure Bone Cement system with comparison to PMMA Bone cement after wedge fracture simulation to quantify motion, intradiscal pressure at augmented level as well as adjacent level [68].

Figure 3.7 Experimental set up for mechanical loading of spine with 400N Preload
The study used six fresh, ligamentous bisegmental human cadaveric spine specimens. Radiographs and DEXA scans were taken for each specimen to ensure usability in the study, eliminating those had fused levels. The specimens were potted in hardened resin superiorly at the cephalad level and inferiorly at the caudal level, as shown in Figure 3-7. The specimens were fixed to the frame at the caudal end and free to move in any plane at the proximal end. The top frame of the potted specimens contained threaded rods on each of the four sides. Weights were hung on these rods using a system of pulleys to apply pure moments to the specimen in extension (Ext), flexion (Flex), lateral bending (LB), and axial rotation (AR). The injury consisted of wedge shape fracture in anterior column followed by kyphoplasty procedure. Fluoroscopes were taken at every step to study the cement augmentation. The following conditions were evaluated:

- Intact
- Injury to create wedge shape fracture
- Cement augmentation using Kyphoplasty technique (BioCure/ PMMA)
- After Fatigue Cyclic loading for crack opening displacement (COD)

Table 3.4 Specimen data and information used for biomechanical testing.

<table>
<thead>
<tr>
<th>Group</th>
<th>Age</th>
<th>Sex</th>
<th>T-Score</th>
<th>Specimen Segment</th>
<th>Augmented Level</th>
<th>Bone Cement</th>
<th>Cement Injection Volume</th>
</tr>
</thead>
<tbody>
<tr>
<td>Group I</td>
<td>62</td>
<td>Male</td>
<td>-3.2</td>
<td>1 L3-L5</td>
<td>L4</td>
<td>PMMA</td>
<td>6 cc</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td></td>
<td>2 T12-L2</td>
<td>L1</td>
<td>Biocure</td>
<td></td>
</tr>
<tr>
<td>Group II</td>
<td>71</td>
<td>Male</td>
<td>-2.7</td>
<td>3 L3-L5</td>
<td>L4</td>
<td>PMMA</td>
<td>5 cc</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td></td>
<td>4 T12-L2</td>
<td>L1</td>
<td>Biocure</td>
<td></td>
</tr>
<tr>
<td>Group III</td>
<td>58</td>
<td>Female</td>
<td>-3.1</td>
<td>5 L3-L5</td>
<td>L4</td>
<td>PMMA</td>
<td>4 cc</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td></td>
<td>6 T12-L2</td>
<td>L1</td>
<td>Biocure</td>
<td></td>
</tr>
</tbody>
</table>
3.10.1 Motion Measurement

Figure 3.8 LED markers attached to each vertebra and motion capture system to capture its motion.

Range of motion data was collected during all loading sequences for each instrumented case. An Optotrak® (NDI, Waterloo, ON, CAN) motion tracking system was used to record ROM data for each loading step. This system uses infrared emitting diode (IRED) markers to record X, Y, and Z-coordinates. The positions of the markers were recorded at 0, 1.5, 3, 4.5, 6 and 7.5 Nm for extension, lateral bending, and axial rotation. The data collected from the Optotrak system was used to calculate the Euler
angles of these local coordinate systems and subsequently to calculate the relative angular motion between the vertebra by means of specially developed software and MS excel macros. From this data, the percentage change in relative angular motions of various test conditions was calculated at different load levels. The p-values were then compiled to determine any significant differences between the cases for a given loading condition. The student’s paired T-test was performed using Minitab statistical software (Minitab Inc., State College, PA).

3.10.2 Intradiscal Pressure Measurement (IDP)

Figure 3.9 Placement on Intradiscal strain gauge pressure sensor placed in anterior one third for pressure measurement comparison in different loading modes.
In order to map the IDP, several pressure transducers (Model 060, Precision Measurement Company) were used. Pressure sensors were placed inside the disc above and below augmented vertebra Figure 3.9. For placing the transducers in correct positions, anteroposterior (AP) and lateral dimensions of both the discs were measured. A suture at each level was used to place the sensors in the correct position. On the suture, a pressure transducers were glued firmly that will be placed in anterior one third of vertebral column. A mark was made on this suture at a distance one-half of the total AP dimension from the middle transducer.

3.11 Cyclic complex fatigue testing for cement fatigue

Cyclical testing under complex coupled motions provides a loading paradigm that simulates and may exceed the mechanics of human motion during daily activities. Each augmented specimen was loaded to a maximum bending moment of 7.5Nm in flexion/extension coupled with $5^0$ axial rotations ($90^0$ phase shift from flexion) for the cement injection segments. Following preconditioning, each specimen were loaded up to 125,000 cycles, which has been cited as the maximum bending cycles performed within two year. The testing was conducted on augmented specimens only [7, 69].
Figure 3.10  Cyclic loading testing on MTS machine for application of bending moment and rotational torque to individual spine segment

The experimental set up for the cyclic complex loading is shown in Figure 3.8. MTS 858 Mini Bionix Servo-hydraulic was used to provide complex motion to the motion segment. A follower preload of 400N was also applied to the specimens. Biaxial MTS machine allows linear and rotational displacement. Testing jig was utilized to apply flexion/extension moments along with axial rotation moments to mimic physiological loading experienced by augmented vertebra in first year after surgery. Six lumbar motion segments (T12-L2 or L3-L5) were tested under cyclic complex loading. The L1 or the L4 were augmented vertebrae with Biocure and PMMA bone cement respectively. Radiographic pictures were taken before and after dynamic loading.
Figure 3.11 Lateral radiographic image of cement augmented spine segment
3.12 Post Fatigue Analysis

Figure 3.12 CT images of fatigue tested vertebral body for non destructive analysis for endplate crack/ damage detection after fatigue testing

CT images were taken of individual augmented vertebra to look for any potential crack developed in a vertebra after complex cyclic fatigue loading. This is done for comparison of both augmented cemented vertebrae [7].
Figure 3.13  Destructive sectioning of vertebral bodies after fatigue loading to identify cracks in bone cement

Destructive Analysis (Sectioning) was carried out in the sagittal plane to obtain two halves of the vertebral bodies. After obtaining the microscopic pictures of the sagittal sections, each half was further sectioned through its center, to make four equal parts of the vertebral body. Microscopic pictures were taken for each of the four sections.
Chapter 4

Results

4.1 Overview

This chapter will first outline the finite element analysis of augmentation done at the thoracolumbar spine. Since single vertebra model was considered initially the endplate stresses will be the important parameter that will be addressed initially. Then the T12-L3 thoracolumbar finite element model was considered and that will investigate the effect of cement stiffness on the motion, facet forces and intradiscal pressure in addition to endplate stresses at constant volume.

It will be followed by cadaveric testing results and other image analysis. From the in vitro experiments performed, height, kinematics and intradiscal pressure in anterior column will be addressed based on different cement stiffness injection. Fatigue performance will be described after cyclic fatigue loading of the augmented vertebral segment for 1,250,000 cycles. Sectioning and CT-scans results will help to get more insides of cement fatigue and possibility of endplate crack with different stiffness cement injection will conclude the chapter.
4.2 Result of finite element analysis of Biocure Cement compared to PMMA in single vertebra model.

Finite model of a single vertebra with cement injection 3-5 cc was developed. All bone and cement properties were same as previous studies. Compression force was applied on superior surface of a vertebra L1. Endplate Stresses were computed. Results were calculated for fractured as well as for non fractured model.

![Single vertebra model under axial compressive load.](image)

Figure 4.1 Single vertebra model under axial compressive load.
4.2.1 No Fracture model

In this model only cement augmentation of 5 cc of cement was done with BioCure cement and PMMA at different loads.

Table 4.1 Endplate stresses comparison and percentage increase after cement augmentation with different stiffness

<table>
<thead>
<tr>
<th>Loads</th>
<th>Intact</th>
<th>Biocure</th>
<th>PMMA</th>
<th>% Increase Biocure</th>
<th>% Increase PMMA</th>
<th>Intact</th>
<th>Biocure</th>
<th>PMMA</th>
<th>% Increase Biocure</th>
<th>% Increase PMMA</th>
</tr>
</thead>
<tbody>
<tr>
<td>800N</td>
<td>0.09</td>
<td>0.11</td>
<td>0.17</td>
<td>22.2%</td>
<td>88.9%</td>
<td>0.04</td>
<td>0.05</td>
<td>0.11</td>
<td>25.0%</td>
<td>175.0%</td>
</tr>
<tr>
<td>800N</td>
<td>0.13</td>
<td>0.15</td>
<td>0.22</td>
<td>13.6%</td>
<td>66.7%</td>
<td>0.09</td>
<td>0.06</td>
<td>0.15</td>
<td>1.7%</td>
<td>154.2%</td>
</tr>
<tr>
<td>1000N</td>
<td>0.16</td>
<td>0.19</td>
<td>0.28</td>
<td>10.8%</td>
<td>75.0%</td>
<td>0.07</td>
<td>0.08</td>
<td>0.23</td>
<td>14.3%</td>
<td>300.0%</td>
</tr>
<tr>
<td>1200N</td>
<td>0.19</td>
<td>0.22</td>
<td>0.34</td>
<td>15.8%</td>
<td>81.1%</td>
<td>0.08</td>
<td>0.102</td>
<td>0.34</td>
<td>27.5%</td>
<td>330.0%</td>
</tr>
<tr>
<td>1600N</td>
<td>0.24</td>
<td>0.28</td>
<td>0.492</td>
<td>18.7%</td>
<td>80.0%</td>
<td>0.11</td>
<td>0.127</td>
<td>0.432</td>
<td>33.4%</td>
<td>285.7%</td>
</tr>
<tr>
<td>2000N</td>
<td>0.33</td>
<td>0.392</td>
<td>0.576</td>
<td>15.0%</td>
<td>74.5%</td>
<td>0.16</td>
<td>0.17</td>
<td>0.576</td>
<td>13.3%</td>
<td>284.0%</td>
</tr>
</tbody>
</table>
Figure 4.3: Endplate stresses observed at superior endplate after loading of the vertebra at various loads.

Figure 4.4: Endplate stresses observed at inferior endplate after loading of the vertebra at various loads.
4.2.2 Fracture Model

Table 4.2 Endplate stresses and percentage increase in endplate stresses with cement augmentation at different stiffness.

<table>
<thead>
<tr>
<th>Loads</th>
<th>Intact</th>
<th>Biosecure</th>
<th>PMMA</th>
<th>% Increase Biosecure</th>
<th>% Increase PMMA</th>
<th>Intact</th>
<th>Biosecure</th>
<th>PMMA</th>
<th>% Increase Biosecure</th>
<th>% Increase PMMA</th>
</tr>
</thead>
<tbody>
<tr>
<td>800N</td>
<td>0.2523</td>
<td>0.301</td>
<td>0.7225</td>
<td>19.3%</td>
<td>186.4%</td>
<td>0.0648</td>
<td>0.0673</td>
<td>0.1819</td>
<td>23.3%</td>
<td>233.2%</td>
</tr>
<tr>
<td>900N</td>
<td>0.3086</td>
<td>0.4015</td>
<td>0.9035</td>
<td>19.3%</td>
<td>186.2%</td>
<td>0.0729</td>
<td>0.0869</td>
<td>0.2426</td>
<td>23.3%</td>
<td>232.8%</td>
</tr>
<tr>
<td>1000N</td>
<td>0.4208</td>
<td>0.5021</td>
<td>1.2045</td>
<td>19.3%</td>
<td>186.2%</td>
<td>0.0911</td>
<td>0.1124</td>
<td>0.3036</td>
<td>23.4%</td>
<td>233.2%</td>
</tr>
<tr>
<td>1200N</td>
<td>0.5053</td>
<td>0.8028</td>
<td>1.4458</td>
<td>19.3%</td>
<td>186.1%</td>
<td>0.1094</td>
<td>0.135</td>
<td>0.3643</td>
<td>23.4%</td>
<td>233.0%</td>
</tr>
<tr>
<td>1500N</td>
<td>0.632</td>
<td>0.7558</td>
<td>1.8074</td>
<td>19.3%</td>
<td>186.0%</td>
<td>0.1389</td>
<td>0.1889</td>
<td>0.4688</td>
<td>23.4%</td>
<td>232.9%</td>
</tr>
<tr>
<td>2000N</td>
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<td>1.0829</td>
<td>2.4105</td>
<td>19.3%</td>
<td>185.8%</td>
<td>0.1928</td>
<td>0.2033</td>
<td>0.6086</td>
<td>11.2%</td>
<td>232.9%</td>
</tr>
</tbody>
</table>

Figure 4.5 Endplate stresses observed at superior endplate in fractured vertebra model after loading it at various loads.
Figure 4.6  Endplate stresses observed at inferior endplate in fractured vertebra model after loading it at various loads.

Since the Biocure cement has properties similar to cancellous bone properties, results showed that the endplate stresses did not increase drastically with Biocure cement at higher loads. The effect of cement stiffness did get amplified when fracture was introduced in osteoporotic model. In all cases it was seen that the superior endplate stresses were higher than inferior endplate. At higher loads percentage increase in endplate stresses with PMMA bone cement increases by almost three times and that would increases a chance of fracture in dynamic loading conditions.
4.3 Finite Element Analysis of Thoracolumbar No-Fracture model with cement augmentation

4.3.1 Compression load – 600 N

Figure 4.7 Intradiscal pressure measurement at augmented and adjacent levels in intact model after cement augmentation.
Figure 4.8  Endplate stresses measured at the augmented and adjacent levels in intact model after cement augmentation.

Figure 4.9  Endplate stresses at superior endplate of augmented vertebra after in the intact model after application of 600N compression force.
At 600N of compression loads which is experienced by spine in daily activities our model showed slight increase in intradiscal pressure and increase in adjacent level endplate stresses with the augmentation of cement of higher stiffness. Endplate stress variation could be seen in adjacent endplate to the augmented vertebra without affecting extreme caudal and cephalad endplates.
4.3.2 No fracture only cement injection – 1500 N

![INTRADISCAL PRESSURE AT 1500N](image)

**Figure 4.11** Intradiscal pressure measurement at augmented and adjacent levels in intact model after cement augmentation.
Figure 4.12 Endplate stresses measured at the augmented and adjacent levels in intact model after cement augmentation.
Figure 4.13 Endplate stresses at superior endplate of augmented vertebra after in the intact model after application of 1500N compression force.

Figure 4.14 Endplate stresses at inferior endplate of augmented vertebra after in the intact model after application of 1500N compression force.

Increase in intradiscal pressure with augmentation of cement was seen at disc levels adjacent to augmented vertebra. Adjacent level endplate stresses increases as stiffness of the cement increases. Adjacent level stresses gets almost double when application of compressive load increase from 600N to 1500 N.
4.4 Results of Finite element analysis on Osteoporotic thoracolumbar model with cement augmentation.

This section discusses the result and data acquired by application of various moments and compressive forces on thoracolumbar validated model.

4.4.1 Bending moments application with preload on T12-L3 Model comparing BioCure and PMMA Cement

Figure 4.15 Load and moment application on the finite element model of thoracolumbar spine.
Three cases of model were prepared:

- Intact Osteoporotic Model
- Biocure cement Injection Model
- PMMA cement Injection Model
4.4.1.1 *Kinematic data*

![Motion at T12-L1](image1)

![Motion at L1-L2](image2)

![Motion at L2-L3](image3)

Figure 4.16 Relative motion at all levels in various loading modes.
### 4.4.1.2 Intradiscal pressure measurement

![Graphs showing intradiscal pressure at T12-L1, L1-L2, and L2-L3](image)

**Figure 4.17** Relative intradiscal pressure at all levels in various loading modes.
4.4.1.3 *Facet loads*

Figure 4.18 Facet force comparison at all levels in various loading modes.
4.4.1.4 *Endplate Stresses*

![Graph showing endplate stresses in extension at various levels with different cement injection.](image)

Figure 4.19 Endplate stresses observed in extension at various levels with different cement injection.

![Biocure and PMMA stress distribution graphs.](image)
Figure 4.20 Superior endplate stresses in extension after injection of Biocure and PMMA bone cement.

Figure 4.21 Inferior endplate stresses in extension after injection of Biocure and PMMA bone cement.
Figure 4.22 Endplate stresses observed in flexion at various levels with different cement injection.

Figure 4.23 Superior endplate stresses in flexion after injection of Biocure and PMMA bone cement.
Figure 4.24 Inferior endplate stresses in extension after injection of Biocure and PMMA bone cement.
Figure 4.25 Endplate stresses observed in left bending at various levels with different cement injection.
Figure 4.26 Superior endplate stresses in extension after injection of Biocure and PMMA bone cement.

![Diagram showing superior endplate stresses in extension after injection of Biocure and PMMA bone cement.]

Figure 4.27 Inferior endplate stresses in left bending after injection of Biocure and PMMA bone cement.

![Diagram showing inferior endplate stresses in left bending after injection of Biocure and PMMA bone cement.]

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Figure 4.28 Endplate stresses observed in left rotation at various levels with different cement injection.
Figure 4.29 Superior endplate stresses in left rotation after injection of Biocure and PMMA bone cement.

Figure 4.30 Inferior endplate stresses in left rotation after injection of Biocure and PMMA bone cement.

Osteoporotic thoracolumbar model was loaded with 400N preload and moment of 10Nm and various parameters were studied. There was no change in motion seen at augmented level or adjacent level due to change in stiffness of cement in augmented vertebra compared to intact. Increase in intradiscal pressure at disc level superior to augmented vertebra was seen in bending and axial rotation but no significant change were observed in adjacent disc levels. Facet loads were unaltered in all loading modes with injection of different cement stiffness. PMMA cement injection shows 15% increase
in endplate stresses near augmented compared to Biocure in flexion. Endplate stresses with PMMA cement injection were slightly higher than Biocure in other loading modes.

4.4.2 Compression Test Results

Figure 4.31 Application of compressive force using preload wires in thoracolumbar T12-L3 spine model.
4.4.2.1 600N Compressive Loading

Figure 4.32 Cross sectional view of stress distribution over the thoracolumbar finite element model after application of compressive load.
Figure 4.33 Maximum stresses experienced at superior and inferior endplate of augmented vertebra compared to normal.
Figure 4.34  Vertical component of force measured at the center of superior endplate for various models.

Figure 4.35  Shear force calculated at center superior endplate for various model simulations
Figure 4.36 Endplate stresses compared at augmented as well as adjacent levels after cement injection

Figure 4.37 Endplate stresses at superior endplate of augmented vertebra after application of 600N compression force.
Figure 4.38  Endplate stresses at inferior endplate of augmented vertebra after application of 600N compression force.
Figure 4.39  Intradiscal pressure measurement at augmented and adjacent levels after cement augmentation.
Figure 4.40 Facet force comparison at augmented and adjacent level after different stiffness cement injection.
4.4.2.2 1500N Compression Loading

![Figure 4.41](image)

Figure 4.41 Endplate stress comparison at augmented and adjacent level after different stiffness cement injection.
Figure 4.42 Endplate stresses at superior endplate of augmented vertebra after application of 1500N compression force.

Figure 4.43 Endplate stresses at inferior endplate of augmented vertebra after application of 1500N compression force.
Figure 4.44 Intradiscal pressure measurement at augmented and adjacent levels after cement augmentation.
Compressive loads were applied to osteoporotic fractured model and effect of Biocure and PMMA bone cement augmentation were compared. Figure 4.33 shows maximum stresses experienced by endplate with various conditions. Result showed that endplate stresses increased by more than double when fracture is introduced in osteoporotic model and then stabilized using cement augmentation. Significant increase in endplate stresses at augmented levels was observed with PMMA augmentation in comparison to Biocure. With 600N compression and 1500 N compression loading there is increase in endplate stresses and intradiscal pressure in PMMA cement augmentation is comparison to Biocure. Facet forces were reduced at augmented level with PMMA with respect to Biocure augmentation which shows the change in load distribution in spinal column at augmented levels.

4.5 Result of In-vitro biomechanical evaluation of Biocure Cement compared to PMMA.

The average values of the motion achieved after applying 7.5 Nm is shown in Figure 4.13. The values of this resultant motion are tabulated in Table 4.4. Flexion and extension demonstrated the greatest motions as compared to lateral bending and rotation. There was
excessive motion in fractured segment which gets stabilized after cement augmentation. Percentage increase in intradiscal pressure after cement augmentation was seen but it was comparatively less in Biocure in comparison to PMMA. There were minimal changes in moments for the augmented spines in flexion, bending, and rotation when compared to the intact situation. The values of the resultant moments are tabulated in table 4.5. All six specimens could be divided into three groups based on cement volume in augmentation and grouped based on same specimen with similar bone properties and different levels.

Table 4.3 Specimen information selected specimen samples for biomechanical testing.

<table>
<thead>
<tr>
<th>Group</th>
<th>Age</th>
<th>Sex</th>
<th>T-Score</th>
<th>segment</th>
<th>Augmented Level</th>
<th>Bone Cement</th>
<th>Cement Injection Volume</th>
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</thead>
<tbody>
<tr>
<td>Group I</td>
<td>62</td>
<td>Male</td>
<td>-3.2</td>
<td>L3-L5</td>
<td>L4</td>
<td>PMMA</td>
<td>6 cc</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td></td>
<td>T12-L2</td>
<td>L1</td>
<td>Biocure</td>
<td></td>
</tr>
<tr>
<td>Group II</td>
<td>71</td>
<td>Male</td>
<td>-2.7</td>
<td>L3-L5</td>
<td>L4</td>
<td>PMMA</td>
<td>5 cc</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td></td>
<td>T12-L2</td>
<td>L1</td>
<td>Biocure</td>
<td></td>
</tr>
<tr>
<td>Group III</td>
<td>58</td>
<td>Female</td>
<td>-3.1</td>
<td>L3-L5</td>
<td>L4</td>
<td>PMMA</td>
<td>4 cc</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td></td>
<td>T12-L2</td>
<td>L1</td>
<td>Biocure</td>
<td></td>
</tr>
</tbody>
</table>
4.5.1 Group I: Kinematic and Intradiscal pressure measurement.

4.5.1.1 Motion with and without preload

Figure 4.46 Motion at L3-L4 augmented and adjacent levels at various cases in six loading modes.
Figure 4.47 Motion at L3-L4 with preload in flexion/extension.
Figure 4.48  Motion at T12-L1 augmented and adjacent levels at various cases in six loading modes.

Figure 4.49  Motion at T12-L1 with preload in flexion/extension.
4.5.1.2  **IDP Comparison: with and without preload**

**Figure 4.50** Percentage increase in intradiscal pressure at disc superior and inferior to augmented vertebra compared to Intact.
Figure 4.51 Percentage increase in intradiscal pressure at disc superior and inferior to augmented vertebra compared to Intact in preload case.

4.5.2 Group II: Kinematic and Intradiscal pressure measurement

4.5.2.1 Motion with and without preload

![L3-L4 Motion at 7.5Nm without Preload](image)

Figure 4.52 Motion at L1-L2 with different loading models.
Figure 4.53  Motion at L3-L4 levels with preload in flexion/extension mode.

L3-L4 Motion at 7.5Nm with Preload

L1-L2 Motion at 7.5Nm without Preload
Figure 4.54  Motion at L1-L2 levels with different loading modes.

Figure 4.55  Motion at L1-L2 levels with preload in flexion/extension.
4.5.2.2 *IDP Comparison: with and without preload:*

**Figure 4.56** Percentage increase in intradiscal pressure at disc superior and inferior to augmented vertebra compared to Intact.
Figure 4.57 Percentage increase in intradiscal pressure at disc superior and inferior to augmented vertebra compared to Intact in preload.

4.5.3 Group III: Kinematic and Intradiscal pressure measurement

4.5.3.1 Motion with and without preload

Figure 4.58 Motion at L1-L2 levels with different loading modes.
Figure 4.59  Motion at L1-L2 levels with different loading modes in preload cases.
Figure 4.60 Motion at T9- T10 levels with different loading modes.

Figure 4.61 Motion at L1-L2 levels with different loading modes in preload cases.
4.5.3.2 *IDP Comparison: with and without preload:*

Figure 4.62 Percentage increase in intradiscal pressure at disc superior and inferior to augmented vertebra compared to Intact.
Figure 4.63  Percentage increase in intradiscal pressure at disc superior and inferior to augmented vertebra compared to Intact in preload.

Figure 4.64  In vitro range of motion normalized data with PMMA bone cement injection.
Figure 4.65 In vitro range of motion normalized data with Biocure bone cement injection.

Figure 4.66 In vitro range of motion preload case normalized data with PMMA bone cement injection.
Figure 4.67 In vitro range of motion preload case normalized data with Biocure bone cement injection.

The resultant motions were evaluated with fracture introduction, cement augmentation in the spinal segment. Motion was measured in different loading configurations and compared with intact. Loading was done with and without preload application. Intradosal pressure were measured at anterior column and compared. In all the loading configurations and specimens tested, PMMA induced much higher intradosal pressure in comparison to Biocure cement injection. The difference in intradosal pressure was prominent in disc below augmented vertebra and flexion, bending mode. Though PMMA showed a trend to induce higher intradosal pressure in comparison to Biocure, due to less sample size at specified volume we could not normalize the data. Post fatigue motion data do show some crack opening in extension is around 7% more than Biocure.
augmented cement. Although statistically this may sound comparable, it has no clinical importance. Large standard deviation values were seen due to small sample size.

Quantification of Disc Pressure after fatigue step were not much reliable and were out of consideration since the sensor was removed before dynamic loading and were inserted again for post fatigue step. Sensor location and disc property after fatigue can contribute to change in IDP.

4.6 Fatigue testing

![Axial Displacement](image)

Figure 4.68 Displacement experience by vertebral segment in 1.25 million cycles is shown in flexion and extension.
Figure 4.69 Axial force maintained or required for $7.5^\circ$ flexion/extension moments over the period of 1.25 million cycles.

Figure 4.70 Rotation torque experience by specimen over the period of 1.25 million cycles.
Cyclic Loading for Fatigue testing significantly changes laxity of ligaments and disc properties so motion data was affected similarly. All specimens completed 1.25 million cycles of dynamic loading where rotation torque were reduced over the time at the angle controlled protocol. Fluoroscopic images of specimens (lateral view) were compared with post fatigue fluoroscopic images and did not show any cement migration in any specimen.
4.7 CT images evaluation

Figure 4.72 CT-scan images showing the superior endplate of the vertebra with no cracks and cancellous bone adjacent to with cement touching the endplate.
Figure 4.73 CT images showing Bipedicular vertebral augmentation in osteoporotic vertebra.
Figure 4.74 CT image of Inferior endplate and cancellous bone near interior endplate
4.8 Vertebral body sectioning for visible and microscopic cracks

Figure 4.75 Vertebral sectioning using minitome machine.
Figure 4.76 Vertbral body sections of vertebrae augmented with PMMA bone cement.

Figure 4.77 Vertbral body sections of vertebrae augmented with Biocure bone cement.
Figure 4.78  Sectioned vertebra augmented using Biocure bone cement.
Chapter 5

Discussion

5.1 Overview

This section discusses the protocols that were undertaken to conduct the biomechanical evaluation of bone cement with different stiffness. It then discusses the results obtained from various studies performed. This section includes the conclusion and summary of this research study. Finally the chapter concludes with the limitations of the present study along with its future scope.

5.2 Experimental and computational methods

Vertebral augmentation has proven to be an effective method of vertebral compression fracture stabilization and pain relief. Even with its many successes, there are still questions about the procedure and its clinical complication. Since there is learning associated with clinical investigation, it inspires inventors to come up with innovative solutions. This study was concentrated on the biomechanics of vertebral augmentation. This study addresses the important question whether stiffness of the bone
cement have a biomechanical long term impact? If we cannot avoid the problem based on different variables, can we decrease the probability of compression fracture?

Several clinical studies have shown that recurrent fractures have a tendency to occur in the endplates adjacent to the augmented vertebral bodies [46, 70, 71] indicating that abnormal loading. There has been concern whether or not these recurrent fractures are due to the progression of osteoporosis in patients, but then fracture pattern was not typical of vertebral compression fractures in vertebrae not adjacent to augmented vertebrae. The results of this study have shown that using cements with stiffness greater than that of cancellous bone causes an increase in stresses in the endplates adjacent to augmented vertebral bodies which could lead to adjacent level fractures. Commonly used bone cements such as PMMA fall into this category with a Young’s modulus of about 2000-3000MPa[51]. Previous studied models also indicated that there was no further increase in endplate stresses and deformation of endplate when cements with a Young’s modulus beyond 2500 MPa were simulated. Berlemann et al [58] had shown in their experimental study that augmented cadaveric functional spinal units have lower load to failure than intact cadaveric functional spinal units. These results were counter intuitive to some extent since the functional spinal unit is expected to become stronger after augmentation. Failure was occurring in the adjacent vertebrae. Increase in the stiffness of a vertebral body due to augmentation must be altering the load transfer of the spine.

Our results have shown that as the stiffness of the simulated cement was increased to a value greater than cancellous bone stiffness the adjacent endplate stresses increased
with a high amount especially during compression. These results could explain as to why the augmented spines failed before the intact spines in the Berlemann et al study. Our studies showed that as the stiffness of a vertebral body increases, more of the load was transferred through the vertebral body and comparatively less load through the facets. This means that when the spines were loaded, the augmented spines would transfer maximum load through the first and second columns of the functional spinal unit, whereas the intact spine would distribute the load more evenly (physiologically) through all three columns of the functional spinal unit. This leads to more of the load being transferred through the inter-vertebral disc and to the adjacent vertebral body. When the pressure sensor was placed in the anterior column, a significant increase in the intradiscal pressure with stiffer cement in augmentation was observed. Therefore when an augmented spine is submitted to the same loading as an intact spine, an augmented spine will have most of the load transferred through the vertebral body leading to an increased risk of recurrent fracture. The effects of this change in loading balance can be reduced by keeping the stiffness of the augmented vertebrae closer to the physiological stiffness.

Recently, reversing the effects of osteoporosis with osteoconductive cements has become an area of interest. With the push for osteoconductive cements, it is important to understand what happens within the augmented vertebral body. Bone growth can be affected greatly by the loads applied, and stiffer cements may lead to unbalanced loading within the vertebral body. The stiffness of the augmentation material can play a major role in the load distribution and was evaluated within this study. The stress contour plots
in chapter five illustrated this point. Once again cements closer to that of physiological stiffness would provide better distribution of the load throughout the vertebral body.

Bone is 60–70% mineral with the principal component of calcium phosphate containing carbonate. Hydroxapatite bone replacement material contains large crystalline components that make them difficult for the body to recognize and remodel. These materials have lower fatigue properties than cancellous bone and need to be studied after augmentation. The MTS machine allows applying loads and moments at different rates and cycles while recording the load deformation data. This study utilized MTS machine to apply controlled flexion/extension torque on spinal segment with addition to angle controlled axial rotation for more than one million cycles. This loading method could be applied to many spinal procedures, and could prove to be very useful in fatigue studies. Our study did not show any fatigue crack development when Biocure and PMMA cement were dynamically loaded in a fractured augmented spinal segment. Biocure cement is a combination of PVA (polyvinyl alcohol) and DAA material. This new material was injected in human vertebra in dynamic loading and it passed the testing without any cracks in cement.

Physiological loads and muscle action is considered in the follower load method used while augmentation procedure in motion analysis experiment and finite element analysis. It will also serve as a good tool while evaluate the effectiveness of a surgical method and to simulate more physiologically relevant set up. Such follower load was not considered
in past cement augmentation studies which could result in additional height restoration of collapsed vertebrae in vitro.

In finite element analysis various parameters could be accessed and studied. Finite element model used in this study is a validated model created by Goel et al. All the anatomical as well as material factors were taken into consideration since material is a large determinant of the outcome. Since two bone cements with different stiffness were compared considerations like simulation of osteoporosity and fracture were needed. Finite element results showed increase in endplate stresses and intradiscal pressure while minimal reduction of facet loads in compression or flexion loading. Endplate thickness, its concavity and cancellous bone, posterior bone properties were considered in the finite element modeling, the increase in endplate stress values in the results could be similar to physiological stresses. This effect could be exaggerated in their fatigue performance which could lead to adjacent level fractures. Although, the mechanical properties applied to the model were obtained from literature, large variance of bone properties among the same age group and even the degree of osteoporosis might vary. The large variance in the mechanical values may lead to inaccurate model.

In the experimental study the specimens were potted and moments were applied with and without follower load to simulate muscle action. These conditions do simulate the natural loading of the spine. Cement viscosity is responsible for osseointegration of the cement and it can help in reducing the fracture crack opening displacement. Motion analysis serves as a tool to compare effectiveness of Biocure cement with viscosity lower
than PMMA. Our result showed that the motion in extension loading was lesser when vertebra was augmented with Biocure in comparison to extension values generated with PMMA cement. This shows that Biocure bone cement allows sufficient osseointegration similar to PMMA which had been used for years. In the motion experiment study, pressure sensors were placed in anterior column of disc above and below the augmented vertebrae. Higher intradiscal pressure with PMMA cement augmentation in comparison to Biocure bone cement in all loading modes especially in flexion confirmed our finding of finite element analysis. These findings confirm the changes in load distribution with addition of stiffer cement in fractured vertebra.

Dynamic loading on augmented segments helped in evaluating the fatigue properties of the vertebral segment. All augmented spinal segments were dynamically loaded to simulate flexion/extension with axial rotation for more than one million cycles. In all the specimens who completed the loading cycles, there was a decrease in axial load and rotation torque was observed over the period of time. Since the machine was set in a displacement/angle controlled mode, the reduction in axial load/rotational torque could be responsible for higher motion seen in all post fatigue specimens. For an in situ polymerizing or setting device such as PMMA and calcium salt bone cement, components are prone to migrate or be washed out by surrounding body fluids before the polymerization or setting of the device has occurred. Cohesiveness is defined as the resistance to washout of the formulation for in situ polymerizing or setting of the device. This is a more critical issue for calcium salts than PMMA bone cements because of their
intrinsic setting mechanism. In kyphoplasty procedure where cavity is already created, there are changes of cement migration after it is cured. Fluoroscopic images were taken before and after dynamic loading and there was no migration of cement seen in Biocure as well as PMMA cement.

Non destructive analysis using CT-scans could be a viable tool for encountering the damage to the endplate due to stiffer cement or augmentation method like Kyphoplasty. Various softwares are [72] commercially available which could be used to study the osseointegration, cement volume, voids and crack filling. Different volume distribution could be seen based on surgical approach and cement viscosity. Tomography images could be imported to create a finite element mesh model with matching bone properties and then endplate stresses could be predicted.

Destructive sectioning helped in studying interdigitation of the cement especially in the vertebroplasty. In kyphoplasty procedure the balloon is responsible for cavity creation. In case of vertebroplasty effect of cement viscosity plays a major role in interdigitation and height restoration. In this study, microscopic examination of sectioned vertebrae helped to understand the effect of dynamic loading on different cement material when augmented in an osteoporotic bone. After sectioning the augmented vertebrae some hair-thin fractures were observed only in one Biocure augmented vertebra. Clinical significance of such cracks were not known but studies by Wilke[68] showed similar results. Sectioned images could be studied using image analysis software [73] to compare various cement parameters relating to interdigitation.
5.3 Conclusion

The current study consisted of evaluation of biocompliant material with low modulus using finite element analysis and experimental methods to compare with currently accepted and used PMMA bone cement. Finite element was focused on the load sharing in the vertebral column due to mismatch of injected material in the center of the vertebral body. Results showed that higher increase in stresses on the endplate due to stiffer cement injection could have amplified effect in the long term. Osteoporotic model was consider as a control and models generated with different material augmentation were compared with osteoporotic bone as to achieve similar load sharing to the osteoporotic bone and not a healthy intact bone. Results of our experimental study results revealed that cement augmentation does help in fracture fixation and restricts excessive flexion due to vertebral body collapse. Increase in intradiscal pressure is indicative of higher load transfer through the anterior column due to augmentation of cement with a material stiffer than bone. Biocure cement did not show any fracture in the endplate or fracture in the cement after dynamic loading. Lower viscosity of Biocure cement in comparison to PMMA helped in better osseointegration. Motion experiment results also showed less extension in Biocure augmented vertebral segment than PMMA augmented vertebral segment after dynamic loading.
This study has shown that in the case of vertebral augmentation, stronger is not always better. One must consider how altering such properties of the spine will affect other untreated areas. Additional research could be done to assess the effect of volume of vertebral cement leading to endplate stresses and fractures in long term.

5.4 Limitations

Finite element analysis and experimental studies helped in evaluating the performance of new material low modulus biocompliant Biocure bone cement in comparison to PMMA cement. Sample size in the experimental studies was not sufficient to validate the finite element results. Statistical analysis did not show any significant difference due to small sample size. Though motion analysis showed similar performance of Biocure cement with PMMA using kyphoplasty technique, performance in vertebroplasty procedure with high pressure cement injecting is not known.

Finite element osteoporotic spine model was used and consistent value of cancellous and cortical bone was used with 60% reduction in cancellous bone and 30% reduction in stiffness of cortical bone[52]. In older patient the cancellous bone stiffness could be much lesser than our assumption and lower modulus of cancellous bone could magnify the effect of stiffness and volume of cement injection. In finite element modeling, complete height restoration is assumed for kinematic evaluation whereas maximum surgical height restoration using kyphoplasty procedure could be only up to 90%.
5.5 Future work

It would be interesting to study the efficacy of low stiffness cements in vertebroplasty procedure in terms of their fatigue performance. Development of different cavity creation devices would be helpful for better understanding. There is a need of more sophisticated algorithm to predict the volume of cement needed to restore strength and stiffness based on a material property of bone cement and patients bone quality. Prophylactic vertebroplasty could be studied using lower stiffness cement injection in adjacent vertebrae than cement stiffness used in fractured augmented vertebra.
References


34. Wilhelm, K., et al., [Preliminary experience with balloon kyphoplasty for the


