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Engineering Analysis Of Custom Foot Orthotics

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ENGINEERING ANALYSIS OF CUSTOM FOOT ORTHOTICS

A Thesis Presented

by

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ABSTRACT

ENGINEERING ANALYSIS OF CUSTOM FOOT ORTHOTICS

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This thesis presents an engineering approach to the modeling and analysis of custom foot orthotics. Although orthotics are widely used and accepted as devices for the prevention of and recovery from injuries, the design process continues to be based on empirical means. There have been many clinical studies investigating the various effects that the orthotics can have on the kinematics and kinetics of human locomotion. The results from these studies are not always consistent, primarily due to subject variability and experimental nature of the design. Alternatively, a better understanding of the therapeutic effects of custom foot orthotics, as well as designing for optimal performance, can be achieved through simulation-based engineering modeling and analysis studies. Such an approach will pave the way to clarify some of the ambiguous findings found in the clinical studies-based literature. Towards this goal, this research presents a methodical process for the replication of the orthotics' complex three-dimensional geometry and for the construction of finite element analysis models using estimated nonlinear material properties.

As part of this research, laser scanning techniques are used to capture the objects' details and geometry through generation of point cloud surface images by taking multiple scans from all angles. Material testing and Mooney-Rivlin equations were used to construct the hyperelastic nonlinear material properties. Using the mid-stance phase of gait for loading conditions, the ANSYS finite element package was utilized to run analyses on three different load classifications and the corresponding maximum stresses and deflection results were generated.

The results indicate that the simulated models can augment and validate the use of empirical tables for designing custom foot orthotics. They can also provide the basis for the optimal design thicknesses of custom foot orthotics based on an end-user's weight and activities. From a practical perspective, they can also be useful in further exploring different orthotics, loading conditions, and material properties, as well as the effectiveness of orthotics for different foot and lower extremity deformities.

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CHAPTER 1

ENGINEERING ANALYSIS OF A CUSTOM FOOT ORTHOTIC

1.1. Introduction

Biomechanical-based engineering design is where biomechanical knowledge is applied to design products through engineering packages, software and concepts. It involves a multidisciplinary approach to the solution of problems focusing on improving people's quality of life. Biomechanics can be stated as the research and analysis of the mechanics of living organisms. By applying the laws and concepts of mechanics and physics, biomechanical mechanisms and structures can be simulated and studied. When applied to human performance, a greater understanding of safety and performance can be gained through engineering modeling, simulation, analysis and measurement. Significant advancements in the understanding of kinematics and dynamics of human motion, as well as in the design and development of medical devices to enhance human performance can thus offer new paradigms for the holistic solutions to the challenges faced when quality of life is compromised. On the basis of these considerations, this research aims to design and develop a rigorous engineering modeling and analysis procedure for a widely used performance enhancement device, namely the Custom Foot Orthotic (CFO).

With over 50 CFO manufacturers and over half of North Americans in need of orthotic intervention, there is still little scientific evidence in the literature to support the effects of stress redistribution and positive results seen by patients throughout the years. Proper knowledge of how the forces are applied and the mechanics of the interaction between the body and orthotic can facilitate the development of optimally designed orthotics. Accordingly, this research intends to further our understanding of the effects that CFOs have on human movement and performance.

People from varying communities use orthotics, from frail elders to athletes and everyone in between. In the sports world many athletes use foot orthotics to allow them to continue to participate in their sports even after experiencing some injury or ailment. Many times an orthotic is used as not only an injury healing device, but also as an injury prevention device. Orthotics are typically accepted as a method for resolving symptoms by altering the position of the foot, which in turn alters the lower extremities and one's alignment all the way up the body. These adjustments may also change the applied tissue stresses in the foot.

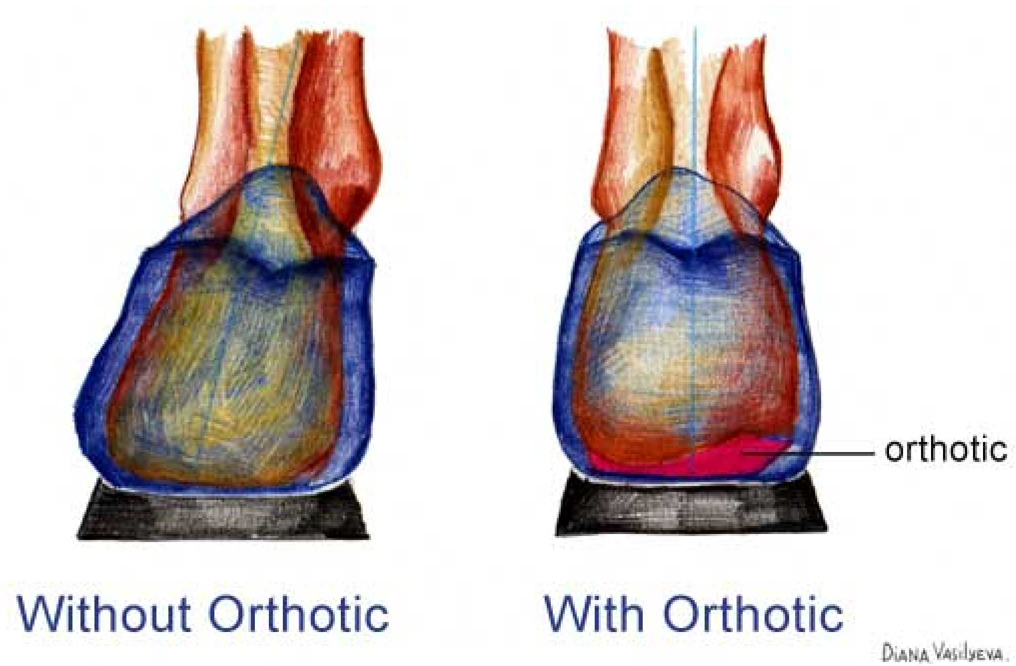


Figure 1. View of rear foot with and without orthotic intervention
(Drawing credit to: www.Podiatrychannel.com)

Despite the fact that engineering modeling and analysis through CAD representations and Finite Element Analysis (FEA) are common in many product designs, tools related to these practices have not yet been fully utilized in the design of custom prescription foot orthotic design. In recent times, the use of FEA to gain insight into the effectiveness of other types of orthotics, as well as provide scientific data and guidelines on orthotic design, has increased tremendously. Although these tools have been used in other orthotic design it has not yet been used in custom prescription foot orthotic design. The combination of complicated geometry, layering of multiple nonlinear materials and computing limitations has made this type of orthotic difficult and time consuming to model. With the computing advances and development of algorithms within FEA as of late, this has become a more attainable research task and is the purpose of this proposed research.

This research stipulates that biomechanics research can be improved when clinical studies are coupled with rigorous engineering methodologies. This research expands on the understanding of human movement and performance through modeling, simulation and analysis. Accordingly, it is the goal of this research to apply FEA based engineering modeling and analysis to better understand the therapeutic effects of orthotics and to validate the FEA results through experimental studies.

The specific steps applied in this research were to acquire specific geometry of custom foot orthotic, acquire specific properties of four materials that make up a custom foot orthotic, build an accurate finite element model of a custom foot orthotic using the specific geometry and material properties and the validation of model accuracy.

These steps can be accomplished by 1) using the latest technological laser scanning device to scan a custom foot orthotic geometry; 2) creating a solid accurate geometry of a custom orthotic to be converted and loaded into ANSYS FEA software package; and finally, 3) mathematically modeling the nonlinear stress-strain behaviors of the materials through repeated data measurements from lab studies.

CHAPTER 2

AN ENGINEERING STUDY OF CUSTOM FOOT ORTHOSIS

2.1. Introduction

This thesis is in the area of biomechanics-based design engineering, focusing on the design and development of optimized assistive and/or rehabilitation devices, specifically Custom Foot Orthoses (CFOs). This will be achieved by building on the expert domain knowledge from the fields of engineering mechanics, design optimization, kinesiology and statistics.

Working in biomechanics-based design engineering allows for a unique opportunity for collaboration between the Mechanical Engineering and Kinesiology Departments. This allows for an engineering perspective to be added to the biomechanics of gait and orthotics research previously done. Previous and current work done by the UMASS Kinesiology Department relating to orthotics includes the orthotic intervention on lower extremity in healthy runners and the orthotic intervention in forefoot and rearfoot strike patterns (Stackhouse C.L. et al., 2004; MacLean C. et al., 2006). The additions will be accomplished by building on work done by Dr. Chris MacLean, and to work hand in hand with Ryan Chang, Pedorthist and PhD candidate in Kinesiology.

2.2. Orthotics

The formal definition of an orthotic is "a support, brace, or splint used to support, align, correct or prevent the function of moveable parts of the body. Shoe inserts are orthotics that are intended to correct an abnormal, or irregular walking pattern, by altering slightly the angles at which the foot strikes a walking or running surface." (Medicinenet) Orthotics work by accommodating irregular foot structures and correcting improper foot function.

Although orthotics are accepted as an effective means of treating and preventing injuries, it may take months for results to be seen because the prescription process is primarily one of qualitative means. In addition, patient compliance may be lower than desired due to a lack of scientific basis for the claim to positive results.

2.2.1. History of Orthotics

In the late 1700's, when shoes were constructed without right/left specificity, i.e., both were made identically, innkeepers at the time recognized that a major complaint of worn out travelers was foot pain. This is when the first insoles were made by innkeepers from matted animal hair (later called felt). Eventually, shoemakers began modifying the foot pads made by the innkeepers. They added leather materials to the insides of shoes thereby improving the fit, and thus giving birth to the first arch supports. Early arch supports were made with layers of leather strips laminated together, molding them to shoe lasts (the solid form around which a shoe is molded), and then shaping an arch support inside the shoe by hand. Although these new arch supports introduced a new level of comfort, they were often heavy and bulky. The bulkiness problem was later alleviated by incorporating lighter and softer materials to be combined with leather, again adding to the comfort of the shoe. In the early 1900's, when electricity allowed for leather laminated devices to be cut down much faster, these arch supports became much easier to make and therefore more affordable for the general population.

The next significant improvement in orthotics came with the introduction of a new generation of thermoplastics introduced in the 1960's. When heat is applied to thermoplastics they become pliable and can be molded to forms such as the foot. Once cooled they hold their shape to create an exact replica. Thermoplastic materials such as polypropylene provide for a strong, durable, thin and extremely lightweight orthotic that can support the body and the contours of the foot while fitting inside all different types and styles of shoes.

The introduction of thermoplastic materials allowed for new theories to be developed on the making of arch supports using casts (or molds) of a patient's foot instead of using the shoe last. Foot supports are developed from the application of scientific principles applied to the foot's structure and to foot orthotics. These supports can be used to control the function of the feet, legs, hips, back and even neck. Today, orthotic design has improved to the point that orthotics can be manufactured to help correct foot deformities and altered gaits that typically may cause pain from the feet all the way up to the neck.

Different people use orthotics for many reasons. Custom prescription foot orthoses are commonly accepted as an effective means of treating and preventing many lower extremity or back injuries and ailments. Often times, orthotics are used as a precautionary measure to help prevent injuries in athletes. For example, injuries are prevalent within the running community. It is very common for runners to sustain injuries from the repetitive high impact movements exerted on the body and in particular the lower extremity joints as well as several kinematic and kinetic factors during the activity. Although the types and general causes of these injuries are relatively well known, there are still many unanswered questions related to the best way to deal with the prevention and recovery aspect of repeated impact injuries.

2.3. Custom Foot Orthotics

There has been very little conclusive research done pertaining to the biomechanical influence of custom foot orthotic intervention. The majority of the clinical studies performed to date have resulted in conflicting results due to significant limitations that will be addressed in the subsequent research.

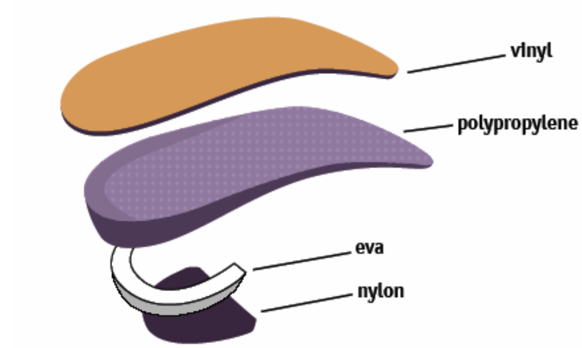


Figure 2 Four layers of a semi-rigid style custom foot orthotic
(Drawing credit to: Kintec FootLabs)

As mentioned above, Custom Foot Orthoses (CFOs) are often used as an acceptable method of managing injuries, and while they usually produce encouraging outcomes, it still remains unclear how the dynamics of the lower extremity are influenced by the device (MacLean et al., 2006). Many previous clinical studies have been performed on the effects of CFO intervention, and many have focused specifically on the effects during running. These studies have focused on rear foot and tibial kinematics, and both lower extremity kinematics and kinetics (MacLean et al., 2006). Variability has been seen in study results due to two main reasons: 1) the types of subjects used; and 2) the design of the experiment (MacLean et al., 2006).

Many research investigations have distributed identical designs to each subject in order to limit the confusing effects of the orthotic (Mundermann et al., 2003). It has been argued that using the same orthotic design for all subjects could be just as or more of a puzzling factor, given that the resulting device may not be comfortable or suitable for each subject's needs (MacLean et al., 2006). CFOs are usually prescribed by podiatrists, physical therapists and sports medicine physicians (Root, 1994) and then manufactured from a volumetric impression of the foot by a certified laboratory to address the specific needs of the patient (MacLean et al., 2006). CFO research has not always included subjects who would normally be candidates for the intervention; many studies have utilized healthy or injury-free subjects (MacLean et al., 2006).

The main findings from CFO clinical studies have been: significant decrease in maximum rearfoot eversion angle (Bates et al., 1979; Smith et al., 1986; MacLean et al., 2006), decrease in maximum rearfoot eversion velocity (Smith et al., 1986; MacLean et al., 2006), decrease in maximum internal ankle inversion moment (Mundermann et al., 2003; Williams et al., 2003; MacLean et al., 2006), decrease in impact peak and maximum vertical loading rate (Mundermann et al., 2003), and decrease in maximum tibial internal rotation angle (Nawoczenski et al., 1995).

Although these studies have shown results, these results have been considered somewhat ambiguous due to the questionable experiement designs mentioned above.

Therefore, the exact effects orthotics have on the kinematics of human locomotion remain unclear due to the fact that it is either extremely difficult, very time consuming or not possible to design a study that will incorporate the appropriate subjects and investigations. This has limited researcher's ability to draw strong conclusions regarding the design and effect of CFOs. This is where modeling and analysis, in particular FEA, can assist in the progression and facilitation of this research. Modeling allows for the ability to control subject variability, thereby minimizing some of the uncertainty found in the current literature.

2.3.1. Engineering Modeling and Analysis

Today's competitive environment has placed a great importance on the ability to reduce the time, effort, prototypes, physical tests, repetitions and expense pertaining to the iterative process used in design of products. Finite Element Analysis, or FEA, has proven to be an excellent tool for analyzing and testing products in a computational environment in order to shorten the time to market, lower development costs and improve product quality. FEA is an engineering analysis method used to determine, among other things, the stresses, strains, structural integrity and fatigue life of many different types of components, structures and machinery. FEA allows for designs to be developed with a high degree of insight and for the ability to perform significant amounts of virtual testing before committing to a particular design for prototyping. FEA is also often used to optimize mechanical designs, increase safety limits, reduce weight, control vibrations and extend life.

Engineering modeling an analysis allows for multiple advancements in the field of biomechanics that could not otherwise be achieved with strictly *in vivo* experimental work. Therefore, this research considers engineering modeling and analysis modeling using FEA as appropriate complementary tools to the current research available to date on custom orthoses through clinical studies in kinesiology. An FEA model can enhance our understanding of orthotics at a micro level, before considering the implications of how they interact with the human body. In practice, clinicians believe that certain design modifications to an orthotic alter the behavior of the orthotic under certain loads. For example, it is know that the stiffness of the support under the medial longitudinal arch can change; however, to what extent is still unknown. Such questions could be answered while avoiding physical tests and experiments using various designs with a valid FE computer model.

Finite element models can be used to enhance the design process of manufacturing orthotics. Presently, clinicians calculate the stiffness of the orthotic based on their experience using the patient's characteristics (i.e. arch height, foot and body mechanics, weight and activity level) and the orthotic design (selected material: orthotic shell, top cover, posting material, alterations to the cast, and additions to the orthotic). In the future, it seems possible to produce interactive software wherein the clinician can complement traditional clinical methods with FE models so as to produce orthotics that are optimal for each particular client.

2.4. Finite Element Related Work on Orthotics

It has been publicized by many researchers that biomechanical factors play a crucial role in the study of orthotics (MacLean et al., 2006). Little biomechanical data is available in the literature to assist in understanding how such factors can effectively be applied to the development of orthoses. It is possible to simulate foot motions, change in material properties, different loading conditions, and different orthotic conditions using accurate FE models. These models can be altered relatively easily, making it possible to further our understanding of the influence that the device has on biomechanical factors.

Currently, the majority of FEAs on orthotics have focused on two types of orthotic inserts: Ankle-Foot Orthotics, or AFOs, and accommodative orthotics. AFO research has focused on analyzing stress points found in the device when in use. This research has allowed for optimal designs leading to the reduction of orthotic fracture and increase in patient compliance. Research on accommodative orthotics has primarily focused on the reduction of peak plantar pressure in the hopes of preventing foot ulcerations. Both will be addressed further in the following sections.

2.4.1. Ankle Foot Orthotic Finite Element Analysis Research

AFOs are designed to help control the motion of the ankle while offering support to the foot. They are often used to treat conditions such as drop-foot, posterior tibial tendon dysfunction, severe flatfoot, arthritis of the ankle and/or foot, ankle sprains, lateral ankle instability and tendonitis.

There are three major objectives for the design of an AFO. The first is to control motion, correct deformity, and compensate for weakness, thereby restoring normal function and ability. The second objective is to make the orthotic as comfortable to wear as possible in order to increase patient compliance. The third objective is to minimize the abnormal appearance of the orthotic. Most advanced AFOs have been unable to improve on all three objectives. The goal of most early research was to either reduce the weight or bulkiness of the orthotic to increase patient compliance or strengthen the weak spots that tend to fail due to high stresses applied by the foot.

Early studies using FEA on AFOs investigated the response of the ankle with and without orthotics (P.C. Lam et al., 1986) by analyzing peak stresses and deformation patterns. Later studies used FEA to predict loads at which AFOs become unstable and analyze the stress distributions (D. Leone et al., 1991; T-M Chu et al., 1991). Due to lack of computational abilities, both of these studies were only able to model the orthotics using linear material properties. More sophisticated models would have been necessary to report accurate numbers such as nonlinear material behavior and accurate geometry. The most recent work on AFOs includes a study using FEA to suggest improvements on lowering the weight and improving the comfort of an orthotic by evaluating real time pressure between the subject and the orthotic during routine actions (walking, chair rise, stair climb, pivoting) via a resistive pad. From the collected data, an accurate model of the orthotic was created and the stress caused by the above activities was evaluated, leading to modification suggestions to reduce orthotic weight

(Khamis S. Abu-Hasaballah et al, 1997). There has not been any FEA research done on ankle-foot orthotics since then.

2.4.2. Accommodative Orthotic Finite Element Analysis Research

Accommodative orthotics are primarily used for the prevention of foot ulcers through the reduction of plantar pressure levels by redistributing the stresses between the foot and orthotic. Foot ulcers are a serious problem for people suffering from diabetes as it can lead to foot amputation and ultimately death. Neuropathy and vascular disease are complications associated with diabetes, and although both may be present, the pathology results in either sensory deficit (neuropathy) or vascular impairment (vascular disease). Skin ulcerations are a result of chronic sensory neuropathy. A protective threshold is when a person possesses adequate sensation to determine when his or her body is at risk of harm from an outside source. At any point below this threshold, there is inadequate sensation to signal the brain to potential harm. When the protective threshold is lost, this allows repetitive, painless trauma to occur to soft tissues and skeletal structures which may further increase the sensory deficit.

Friction, pressure and shearing are the three causes of stress of great concern for diabetics. Friction is considered the surface resistance of one body sliding over another. Blisters are caused by fast and constant friction; the opposite causes calluses. The vertical ground reaction forces applied to the foot is referred to as pressure. Ischemia can be caused by constant pressure and can result in necrosis (tissue death). Shearing is a combination of friction and pressure and can occur when two surfaces slide over each other, with pressure being applied perpendicular to the direction of movement. This force is often produced during normal gait. These forces can cause potential injury to the bones and soft tissues (joint subluxation and skin ulceration).

Orthotic therapy is intended to decrease Ground Reaction Forces (GRF) applied to the foot. An exact mold of the foot is extracted and if localized areas of pressure occur, the GRF's can be reduced by elevating adjacent areas, such as with metatarsal pads. Distributing GRF's over a greater time period will decrease shearing. For example, soft materials will slow the foot by increasing the vertical distance the foot travels before coming to rest. If the orthotic materials are rigid the poor shock absorption and non-accommodative properties will not be helpful for these patients. Corrective components of orthoses aim at decreasing unnecessary pressure on the foot by limiting excess motion and maintaining an unstable foot in proper alignment.

Reduced plantar pressure levels to prevent foot ulcers can be achieved with in-shoe orthoses. They reduce the pressure at bony prominences, especially under the metatarsal heads. Although this method is readily used, very little actual quantitative information is available regarding the effect of thickness and influence of soft tissue characteristics on the cushioning effect of these orthoses. FEA has been used to analyze accommodative orthotics mostly in the late 1990's and most recently in 2003. Nonlinear material properties are difficult to model and only with recent computing advancements has this become more common. The first study used FEA to compare insoles of varying thicknesses by calculating peak plantar pressures and validating these models and values through clinical measurements (Lemmon et al., 1995). Two years later this same group investigated alterations in pressure under the second metatarsal head as a function of insole thickness and foot tissue thickness. The group found that orthoses reduced plantar pressure and offered techniques which allowed for a better approach to understanding plantar cushioning as well as the principals involved in the design of therapeutic footwear(Lemmon et al., 1997). Most recently in Chen et al.'s 2003 research, FEA was used to study the effects of total contact insoles on plantar stress redistribution by analyzing different stress reduction and redistribution. This research allowed for recommendations to be made on the effectiveness of accommodative orthotics. There have been a couple of recent studies on the foot insole interaction, but these studies have primarily been emphasizing the finite element model of the foot as opposed to the orthotic.

Currently nothing comparable exists in the literature regarding the application of FEA to the understanding and design of CFOs and, more specifically, semi-rigid style custom foot orthotics. Traditionally, custom desiging orthotics has been a process primarily using empirical methods. Very little actual quantitative information is available regarding the effectiveness of custom orthoses and little scientific evidence is available to provide guidelines for persons who prescribe insoles. The challenge involved in the modeling and analysis of CFOs lie in the complicated geometry and muti-layering of nonlinear materials.

CHAPTER 3

METHODS

3.1. Research Plan

A four step procedure was proposed to carry out the investigation of the engineering study of CFO. Figure 3 shows a schematic representation of the various stages involved. They include: 1) the estimation of the nonlinear material properties; 2) the generation of complex geometry in CFOs through laser scan and its subsequent conversion to a solid model representation format for further analysis; 3) a thorough FEA analysis using appropriate forces and boundary conditions and, 4) a mechanism for model validation through experimental studies, which will be done at a later time. The following sections detail these steps.

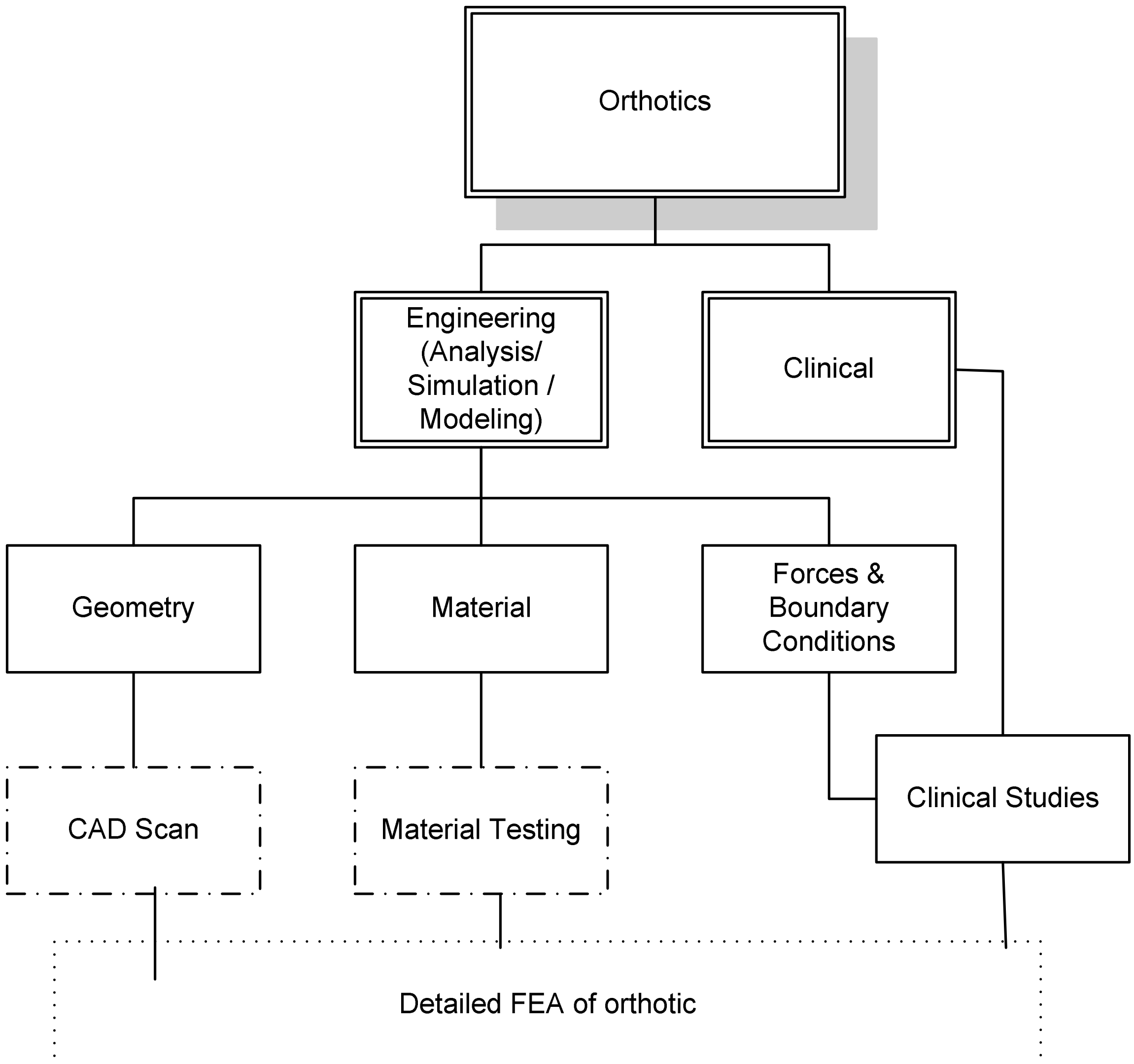


Figure 3 Detailed schematic representation of proposed research tasks

3.1.1. Nonlinear Material Property Estimation

A challenge to executing an engineering analysis of CFOs is the lack of availability of material properties, which are, to begin with, highly complex and nonlinear. To this end, in this research an experimental set-up and an advanced estimation technique were used to assess and calibrate the necessary engineering material properties for the most commonly prescribed orthotic, the semi-rigid style orthotic. The most popular materials used to make up a typical semi-rigid style orthotic are polypropylene, EVA foam, Spenco™, Topy, and McPuff. During normal orthotic design, different thicknesses of polypropylene (support material) are used depending on the person's weight, arch height and the amount of stiffness or flexibility required for support for that patient. The polypropylene is heated up to its melting point in order to be formed to the foot mold. Different material thicknesses are used within a range dictated not by scientific calculations but rather by the prescriber's experience. Currently, there is not enough information in the literature on the material properties that make up these CFOs. As a result, the first step required to create this FE model is to obtain material properties by running uniaxial tensile tests on samples of the materials to obtain their stress-strain behaviors. Using the steps laid out in the "Development of material constants for nonlinear finite element analysis" by Robert H. Finney et al. (1987) and ASTM standard D575-91A, uniaxial tensile tests in the University of Massachusetts Amherst Materials Lab were performed. The resulting stress strain values can be converted into Mooney-Rivlin material constants for material model definitions in ANSYS. Figure 4 shows sample speciments used in uniaxial testing.

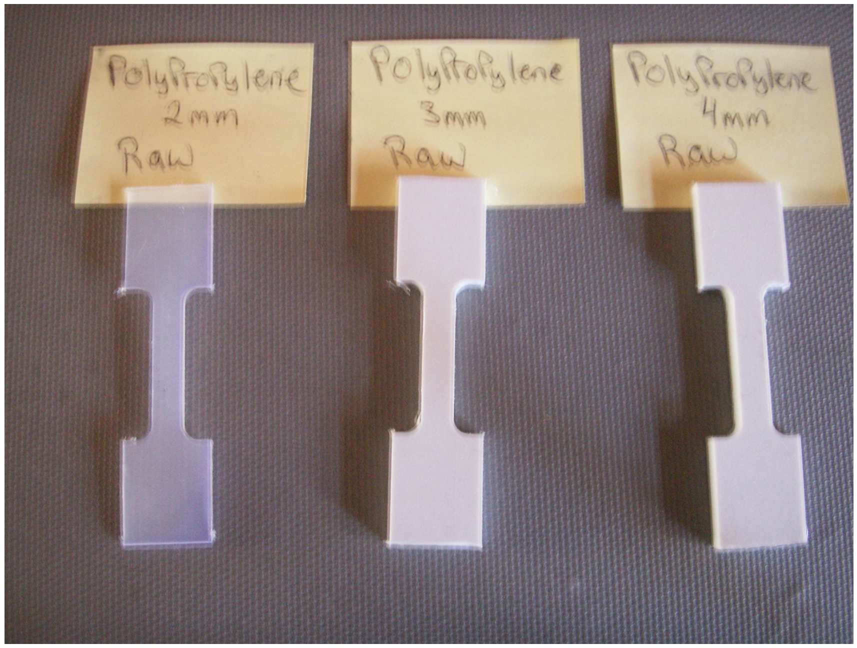


Figure 4 Some samples used in uniaxial tensile testing

One of the challenges encountered in the creation of an FEA model of a semi-rigid style custom foot orthotic (CFO) is estimating the complicated nonlinear material properties. The estimation of material properties can be accomplished by performing uniaxial tensile tests on raw materials and converting the data using a Mooney-Rivlin strain energy function to input those values into ANSYS. The more accurate a finite element model the more useful it will be. Nonlinear materials require a little more detailed knowledge of material properties and characteristics. Custom foot semi rigid style orthotics are primarily comprised of layers of hard and soft nonlinear materials. The material primarily used for the support and therefore investigated in this project is polypropylene. In future work the investigation of the other materials such as Spenco, EVA, and Topy, will be analyzed in their interaction with polypropylene.

3.1.1.1. Material Property Testing of Orthotic Material

The sheets of raw material of polypropylene, 2mm (heated and unheated), 3mm (heated and unheated), and 4mm were supplied by Kintec footlabs, and the uniaxial tensile tests were performed in the University of Massachusetts Amherst Materials Laboratory. The test specimens were cut to standard ASTM D 412 dogbone shape 3" long by .75" wide. Five of each specimen were cut (total of 40) and the stress strain data were recorded from the Instron uniaxial tesile tester in the Materials Lab. Before each sample was run, the width and thickness of each specimen was measured to the nearest 0.025mm (0.001") and loaded into the Instron. The data was exported into labview directly from the Instron machine. The Stress strain data is presented below in figures 5a – 5h.

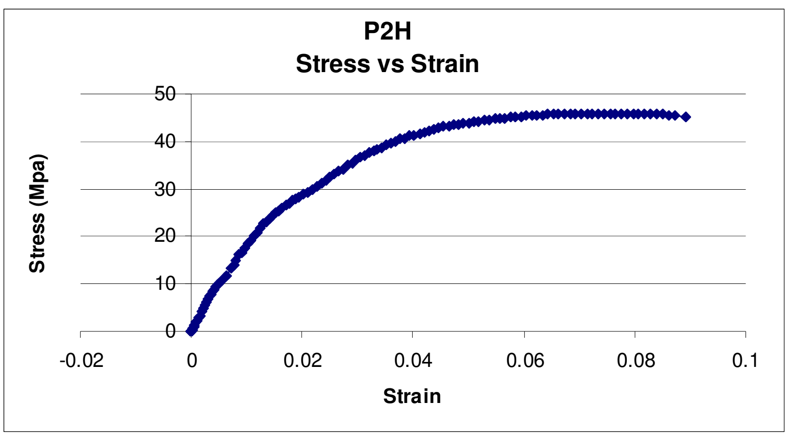


Figure 5a Stress-strain curve for the averaged data set for the 2mm heated polypropylene material

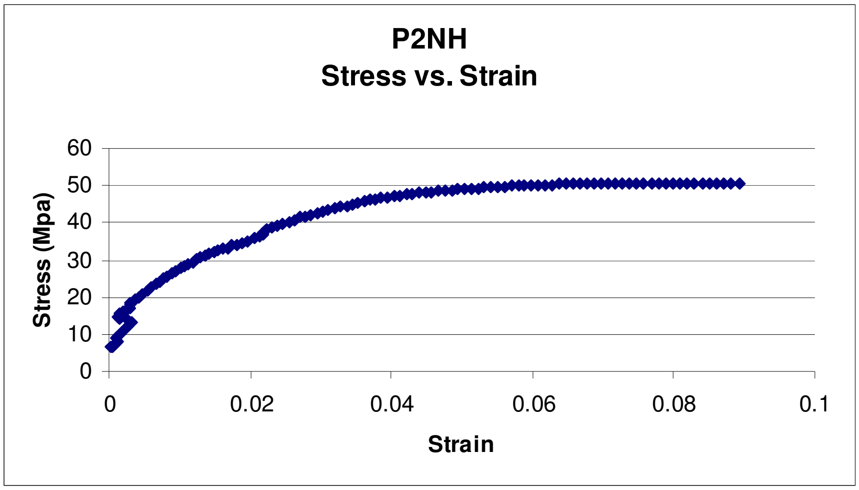


Figure 5b Stress strain curve for the averaged data set for the 2mm raw polypropylene material

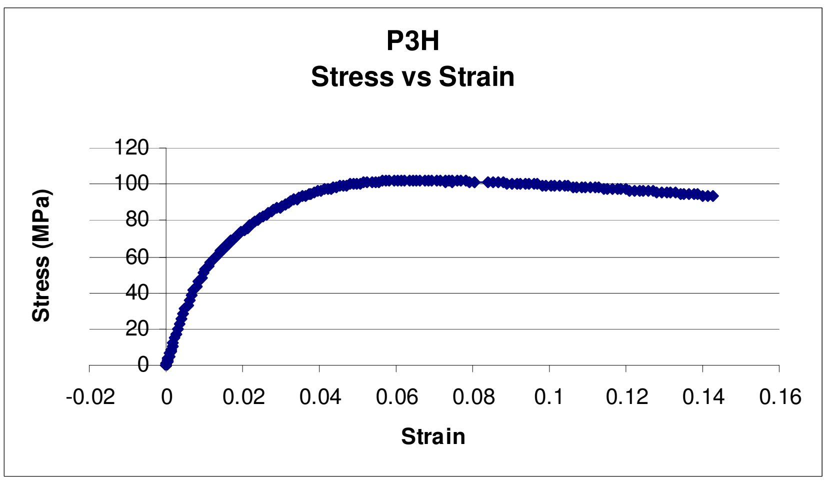


Figure 5c Stress strain curve for the averaged data set for the 3mm heated polypropylene material

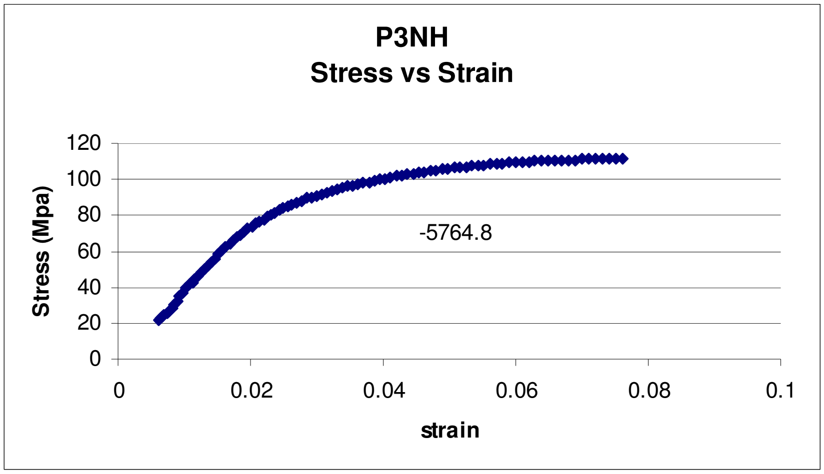


Figure 5d Stress strain curve for the averaged data set for the 3mm raw polypropylene material

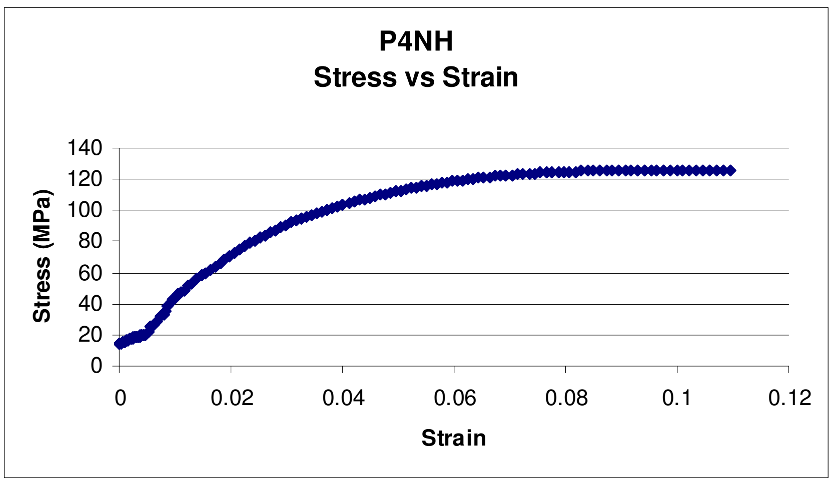


Figure 5e Stress strain curve for the averaged data set for the 4mm raw polypropylene material

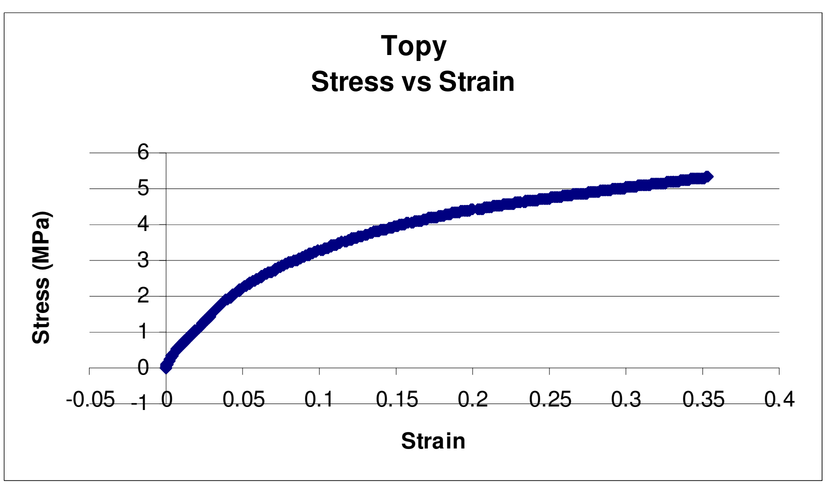


Figure 5f Stress strain curve for the averaged data set for the Topy material

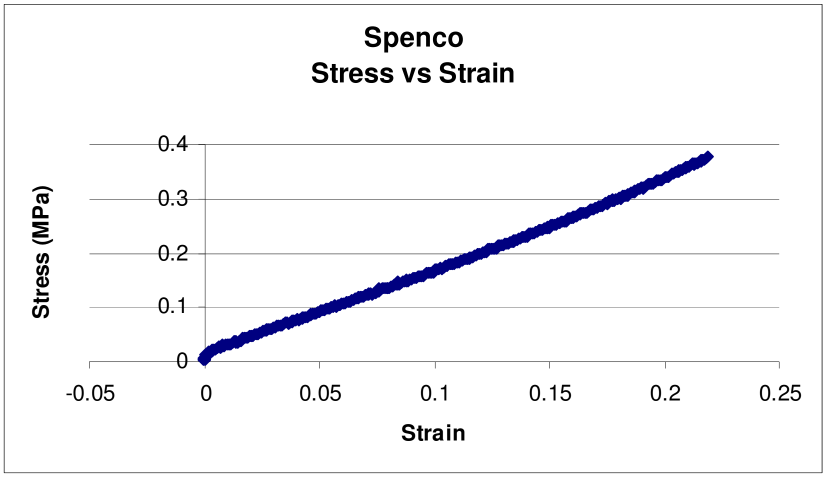


Figure 5g Stress strain curve for the averaged data set for the Spenco material

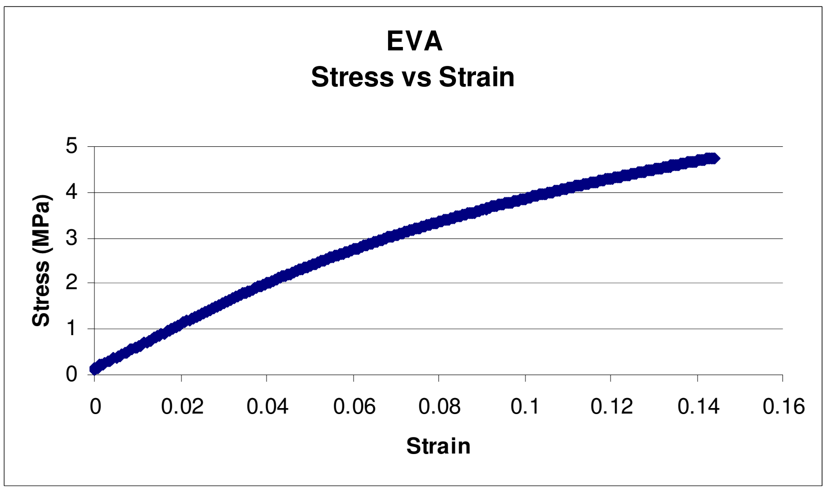


Figure 5h Stress strain curve for the averaged data set for the EVA foam material

Findings for Mooney-Rivlin material constants according to Finney et al., 1987:

According to Finney et al., the most commonly used formula in the analysis of elastomers is the Mooney-Rivlin strain energy function. For uniaxial tension of an incompressible Mooney-Rivlin material, the stress strain equation is expressed by:

$$S = 2*(a - a^{-2})*(C_1 + C_2*a^{-1})$$

Where: S = Cauchy stress (ratio of force to the original area)
a = Principal stretch ratio (1+dL/L)

Using this equation when $(S/(2*(a - a^{-2})))$ is plotted against a^{-1}, the resulting plot will be a straight line with C2 as the slope and $(C_1 + C_2)$ as the intercept at $a^{-1} = 1$. The tension force is converted to tensile stress $S = T/A_{original}$ and the deformation (dL) to principal stretch ratio $a = 1+dL/L_{original.}$ The equation $(S/(2*(a - a^{-2})))$ is plotted against the inverse of the principal stretch ratio (a^{-1}) and as is shown below in figures 6a – 6e the materials analyzed follow the Mooney-Rivlin model since it is in a straight line.

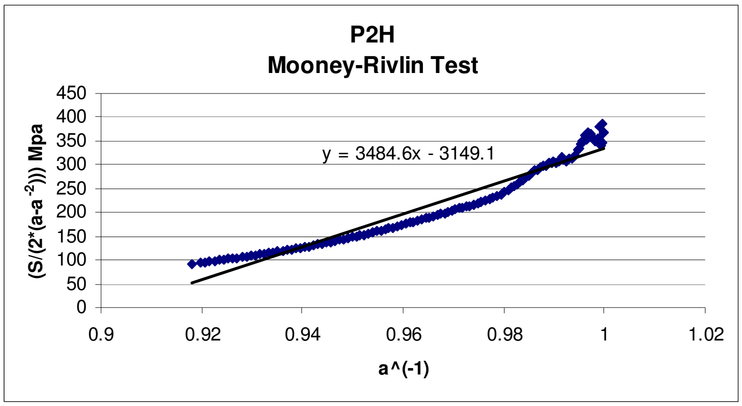


Figure 6a 2mm thick heated polypropylene Mooney-Rivlin test (S/(2*(a – a-2))) is plotted against a-1 validating the ability to use the Mooney-Rivlin strain energy function for the data

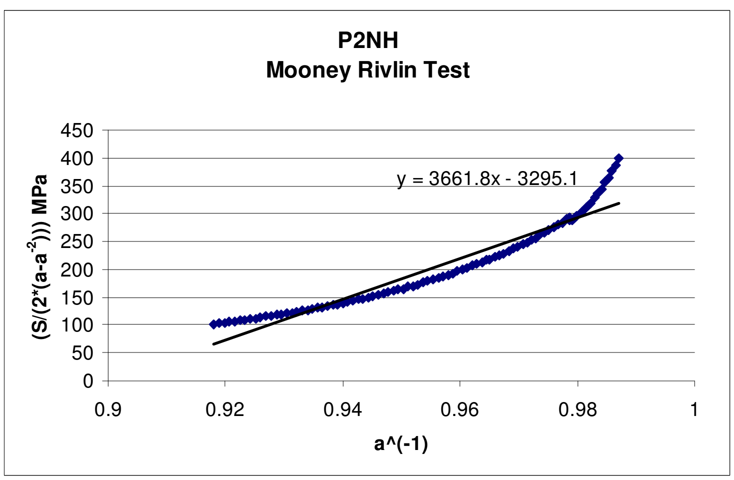


Figure 6b 2mm thick raw polypropylene Mooney-Rivlin test $(S/(2*(a - a^{-2})))$ is plotted against a^{-1} validating the ability to use the Mooney-Rivlin strain energy function for the data

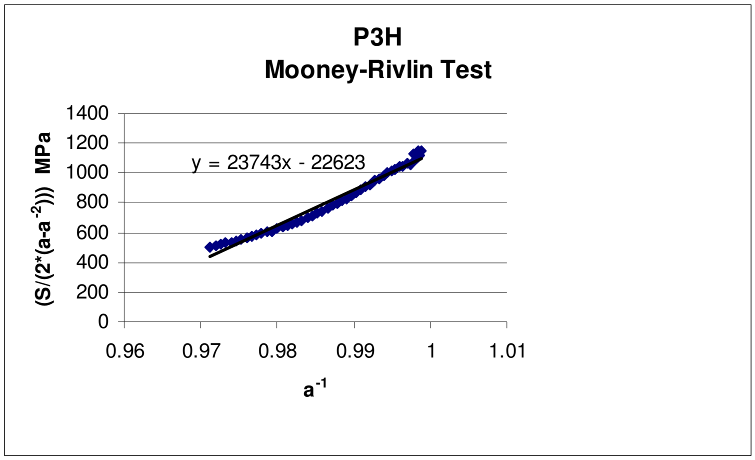


Figure 6c 3mm thick heated polypropylene Mooney-Rivlin test ($S/(2*(a - a^{-2}))$) is plotted against a^{-1} validating the ability to use the Mooney-Rivlin strain energy function for the data

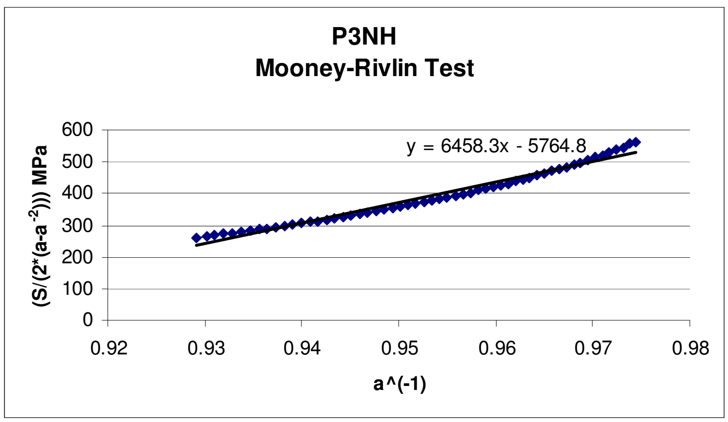


Figure 6d 3mm thick raw polypropylene Mooney-Rivlin test ($S/(2*(a - a^{-2}))$) is plotted against a^{-1} validating the ability to use the Mooney-Rivlin strain energy function for the data

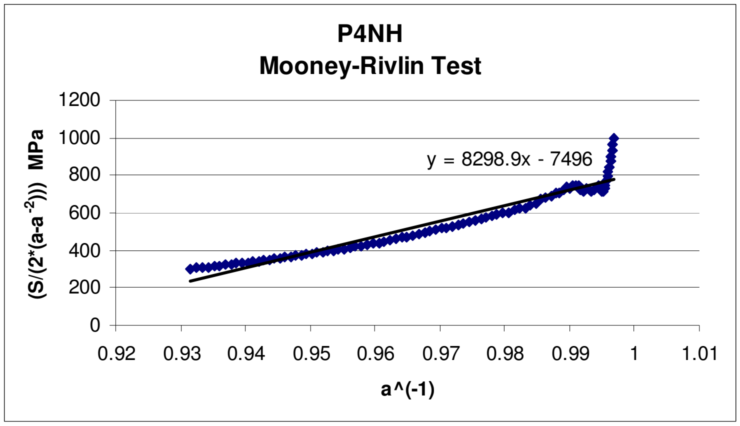


Figure 6e 4mm thick raw polypropylene Mooney-Rivlin test ($S/(2*(a - a^{-2}))$) is plotted against a^{-1} validating the ability to use the Mooney-Rivlin strain energy function for the data

Table 1 Mooney-Rivlin material constants estimated from function and stress strain data. Poisson's ratio of the materials are input through the d value by: d = (1-2*υ)/(C1 + C2) and the material constants are related to the initial shear modulus G by: G

	2mm Heated Polypropylene	2mm Polypropylene	3mm Heated Polypropylene	3mm Polypropylene	4mm Polypropylene
	P2H	P2NH	P3H	P3NH	P4NH
C1 (MPa)	-3149.1	-3295.1	-22623	-5764.8	-7496
C2 (MPa)	3484.6	3661.8	23743	6458.3	8298.9
d (Mpa)	0.000596	0.000545	0.000179	0.000288	2.49E-04
G (Mpa)	671	733.4	2240	1387	1605
E (Mpa)	1342	1466.8	4480	2774	3210

These material constants can be used to designate the appropriate material properties to each material in the finite element model created in ANSYS. The material properties, along with the geometry accuracy, are essential for accurate representation and analyses within those representations. Although different thicknesses of polypropylene were tested, only the results from the 3mm heated polypropylene was used to reduce the number of variables to start out with.

3.1.2. Generation of Complex Geometry

This research recognizes that the replication using CAD tools can be difficult to almost impossible as well as extremely time consuming because of the complex geometry of a custom foot orthotic. Alternatively, accurate geometry can be achieved by laser scanning the device. Laser scanning is a process by which a surface is scanned or sampled by taking multiple scans from all angles in order to capture the object's detail and geometry. This results in a separate point cloud surface image. Although this simplifies the process and allows for the most accurate geometry possible, the process of conversion to a usable format can be challenging. For this research, a semi rigid style orthotic was scanned by a RealScan 3D model 200 laser digitizer (3D Digital Corp, Sandy Hook, CT) by Dr. Saunders Whittlesey. This point cloud data was converted into a solid image and meshed using 3-Matic, a program by Materialise, Inc that allows one to reconstruct and manipulate scanned data directly and exported into ANSYS using the steps outlined in the appendix. These conversion of files were made possible by the very helpful software engineers at Materialise, Inc. The program 3-Matic meshed the model by default with surface element type Shell93. These elements were then converted to 3D 20 Noded tetrahedral elements SOLID186. These elements were used because SOLID186 is a newer version of SOLID95 which means that it tolerates irregular shapes without very much loss of accuracy and it is well suited to model curved boundaries. These elements can also be tetrahedral and can automatically transition between hexahedral and tetrahedral using pyramids. It has capabilities for simulating deformations of nearly incompressible elastoplastic materials and fully incompressible hyperelastic materials which matches well with simulating polypropylene and many of the other materials that will be modeled in the future.

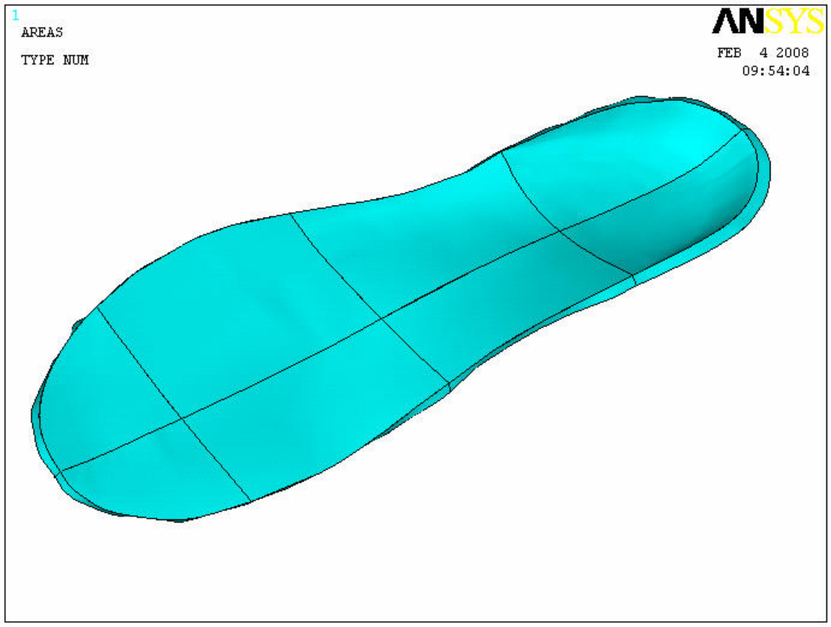


Figure 7 A representative solid model of the scanned orthotic with the areas from ANSYS

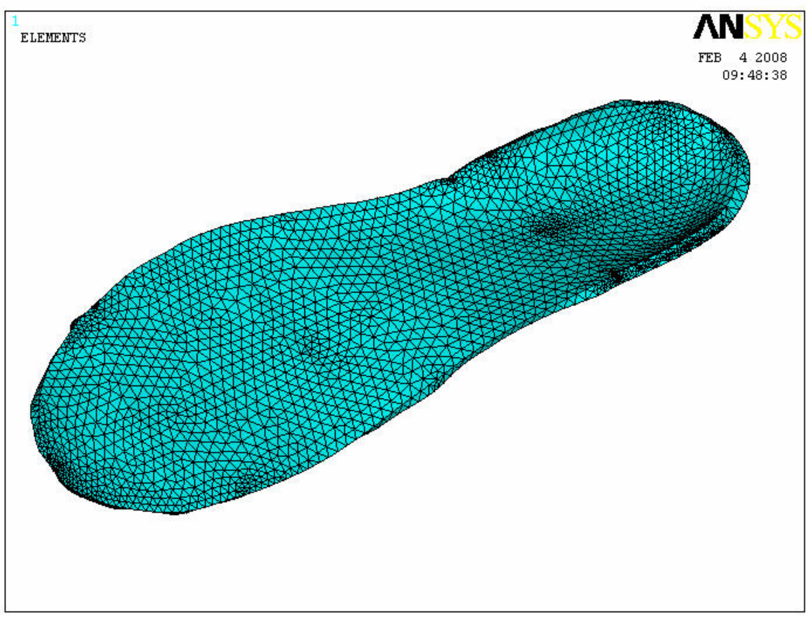


Figure 8 Meshed orthotic representative plot from ANSYS

3.2. Model Development

Utilizing the nonlinear material properties obtained from the material testing described above and the geometry from the laser scanned and converted model, an accurate FEA ANSYS model for the study of force induced deflection in CFOs was developed. Mathematical extrapolation techniques were used to convert the nonlinear material properties from the lab experiments for the inputs into describing the material properties. Once the orthotic was scanned, the file was converted to *.stl format and imported into 3-Matic. 3-Matic then outputs an ANSYS input file of the solid model. The next step after this was to run some in lab clinical trials using the biomechanics lab at the University of Massachusetts Amherst mimicing the loading and constraint conditions used in the FE Model to validate the simulation results.

3.2.1. Preliminary Work: Pilot Study

Custom foot orthoses are vital to the prevention and treatment of many lower extremity ailments but are only effective if they are designed accurately. The effectiveness of the orthotic design can be enhanced with the assistance of the results acquired by finite element models. Two studies were performed with two different simplified objectives by creating and analyzing two sets of models. The first set was created with simplified geometry and the second with simplified material properties.

3.2.1.1.Pilot Study: Study 1

In a first study, an FEA model of a CFO with a simplified geometry was constructed with 4 layers of nonlinear materials. A "heel strike" was simulated on the back outer edge of the model by applying a point load; the deflection and von Mises stresses were analyzed. The purpose of the first study was to analyze the layering of nonlinear material properties modeled in ANSYS.

A very simple geometry representing the area of an orthotic from the heel to the arch (midline) was drawn in ANSYS as an "Area". The surface of the area was then extruded down into a volume, one layer at a time. This volume can be seen in Figure 7 below. Each layer was designated as a nonlinear different material corresponding to the four nonlinear materials in a CFO (Polypropylene, EVA foam, Nylon, and vinyl). Heel strike was simulated by applying a point load to the back of the orthotic representing 1.5* the weight of a 200lb person. A point load was used to simplify the results even though a point load does not accurately replicate heel strike; it was used to simplify the analysis.

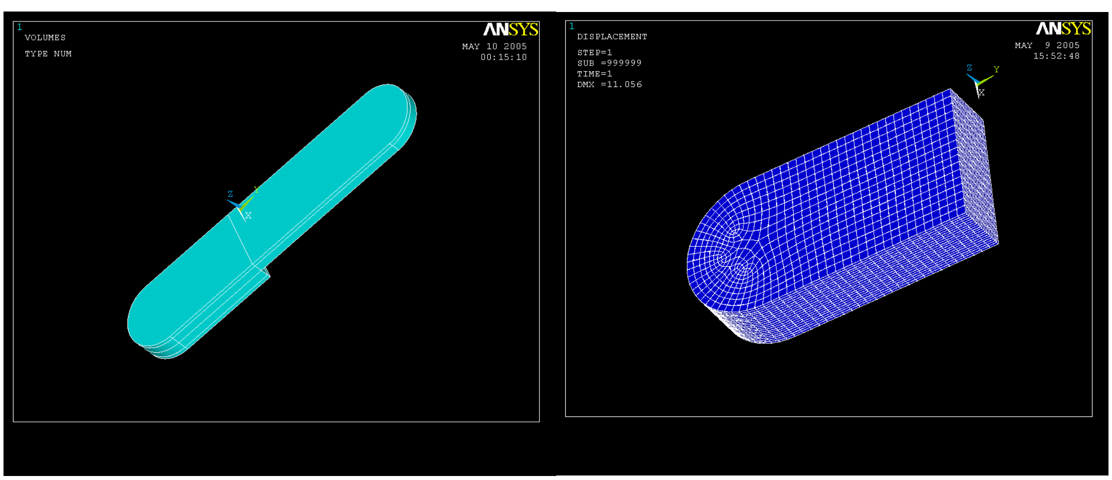


Figure 9 FEA model of simplified geometry orthotic: Full model, and meshed back half of model

Due to the inaccuracies of the nonlinear material property estimation, the strain energy function selection and the application of the point load convergence could not be reached in this model. Layering of nonlinear properties is extremely complicated and as much accuracy as possible is necessary in order for the analyses to converge properly. These results lead to the investigation of accurate material properties and function to use to model these materials properly. The necessity of performing experimental work in order to acquire these numbers became apparent. The research task will be to look in the literature for methods for extracting material properties from experimental studies. This work is not trivial and studies have shown that the use of mathematical extrapolation techniques appear to yield sufficiently accurate results.

3.2.1.2. Pilot Study: Study 2

In the second study the materials were completely simplified by creating one layer of the orthotic (the polypropylene, support layer), modeled as a material with linear properties. Five models with two variables were created. These analyses were accomplished by analyzing the load applied to the outer side of the rear orthotic simulating the heel strike part of the gait cycle using FEA. The 5 models were compared and analyzed using variance of the Ground Reaction Forces (GRFs) to represent stability of the orthotic. The purpose of the second study was to understand the effects of the geometry on the custom foot orthotic and how it affects stability.

Five models were created with two variables. Variable 1 was the wall height and variable 2 was the arch height. The models ranged from "high wall with an arch" to "no wall with no arch". As seen in the two figures below, Variable 1 is the wall height and Variable 2 is the Arch height.

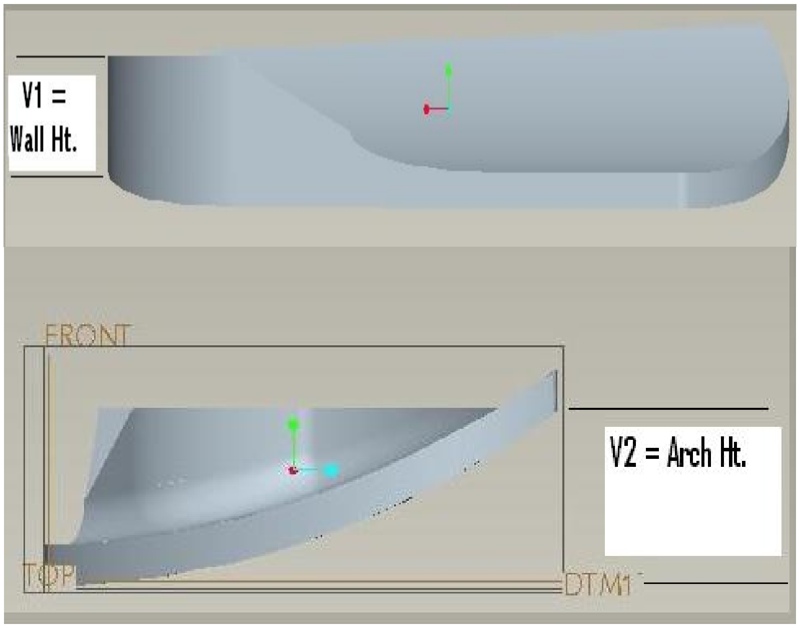


Figure 10 Model of simplified material property orthotic (a) variable 1, wall height, (b) variable 2, arch height

A surface pressure load was applied to the back outer edge of the model to simulate heel strike of the gait cycle. When analyzing the reaction forces, the lower variances would signify more stability.

From the analysis results of this study, it is clear that geometry is a necessary detail and does significantly affect analysis results. The model with "no arch with no wall" had very different results from the model with "an arch and a high wall". The material reaction to the load applied was also not accurately modeled due to the simplification by classifying the materials as linear. The models with "no wall" saw much more deformity than the models with the "medium wall" and "high wall". The arch also affected the stability when the force was applied to the back of the model by showing a much larger variance than the models with "no arch".

Drawing completely accurate geometry is extremely difficult and time consuming to get all of the intricacies of an orthotic design. This will be alleviated by acquiring the exact geometry through other technologically advanced means such as laser scanning. This allows for all the details of the complicated three dimensional geometrical surface to be captured for accurate modeling.

Accurate geometry coupled with an accurate nonlinear material properties and material model function will allow for not only convergence of the model, but accurate results as well.

3.3. Plan of Work

The intent of this research is to build on the two studies discussed above, focusing on two main areas of improvement in order to proceed with an accurate FE model, nonlinear material property estimation and complex geometry replication. These models can replace the use of empirical tables for designing custom foot orthotics and enable the optimal design thicknesses based on end-users' weight and activities. Similarly, they will facilitate the minimization of effort in the simulation of various orthotic and loading conditions, changes in material properties, and foot deformities by simply altering model specifications. Finally, these models and the corresponding results can also form the basis for subsequent design of a new generation of custom foot orthotics.

The first area of improvement is to design an experimental procedure for the estimation of accurate and specific nonlinear material properties from the materials which make up the CFO. Performing uniaxial tensile tests on each material yielded stress strain behavior needed to input precise values resulting in accurate nonlinear model of a custom foot orthotic. The second area is to generate accurate and specific geometry of a CFO through laser scaning and conversion techniques. Specifically, an exact geometry of a CFO will be generated by using a laser scanning device to capture the exact surface geometry and then using the 3-Matics program to convert the image to a solid in a format consistent with CAD representations. The details of this plan of work are shown in Figure 6 below.

3.3.1. Timeline Of Work

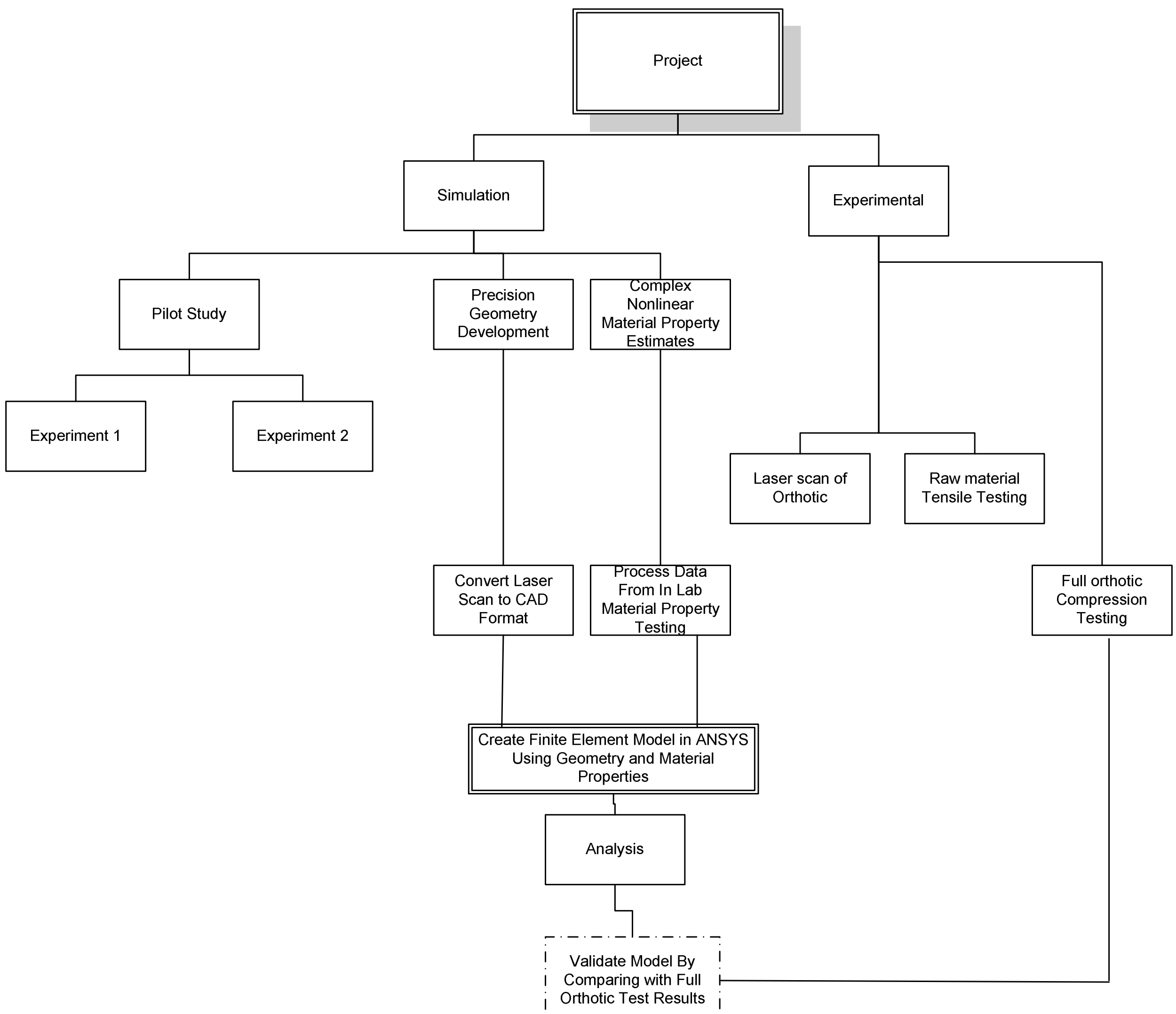


Figure 11 Schematic of detailed plan of work

CHAPTER 4

RESULTS & DISCUSSION

4.1. Results

The objective of this work is to create different models of differing thickness and apply various load conditions to analyze and compare to current thickness to weight classification guidelines. These models will minimize the effort in analyzing various orthotic and loading conditions, changes in material properties, and foot deformities allowing for quick insight into the effect of change to an orthotic without having to build physical models first. This will also allow us to validate and draw more precise guiding principle to prescribers prescription guidelines in reference to the weight to thickness ratio as well as give an alternative reference which can replace the old reference tables. Such as the weight window for each polypropylene thickness and specific material selection. The intent is to have these models and the corresponding results form the basis for subsequent design of a new generation of custom foot orthotics.

4.1.1. Prescribers' Guidelines:Weight-to-Thickness Ratio

The choice of material in the construction of the orthotic is dictated by the amount of control necessary for each patient. The more rigid the material, the more control will be had by the joints of the foot. The choice is decided by the thickness, then density of the material to be used for the shell and post, as well as the weight and activity level of the patient. The thickness of the material will influence its function, and therefore a compromise is drawn between optimum flexibility for weight and maximum control/support. There are different thickness polypropylene materials used depending on the size of the patient and the activity level of that patient. The different thicknesses explored were 2mm, 3mm, 4mm and 5mm. If pure thickness is used to determine the shell to be prescribed by the patients weight; the 2mm thick polypropylene is normally not prescribed on its own, it is normally used for reinforcement of other materials; the 3mm thickness is generally used for a medium sized person in the weight range of less than 45 kg (~100 lbs) but more than 25 kg (~ 55 lbs) and the 4mm thickness for a slightly larger person in the weight range of more than 45 kg, but les than 75 kg (~165lbs). There is also a 5mm thick shell used for people who weigh more than 75 kg.

Currently, pedorthists use a table of guidelines such as the one below in Figure 12 to direct them as to the level of support to choose for each client based on the client's weight and activity level. These guidelines are made based on past experiences, as opposed to scientific methods, and this engineering analysis will give us some insight into whether these guidelines are appropriate or not. These models will also give the prescriber a better tool to follow. The middle column (Category 2) of Figure 12 below lays out the guidelines for the thickness that should be matched with the person's weight for semi-flexible material.

Patient's weight (1 kg = 1000 g = 2.205 lb; 14 lb = 1 stone = 6.35 kg)	Category 1 Rigid material	Category 2 Semi-flexible material	Category 3 Accommodative material
	Europlex TL-2100	NE Polyethylene Polypropylene Aquaplast Acrylonitrile butadiene styrene	Ethylene vinyl acetate (measured in density of foam)
Less than 25 kg	2 mm 1.75mm TL-2100	2 mm of Aquaplast for max. flexibility	220 kg/m^3 Lower densities of polyethylene foam can be used if greater softness is required
More than 25 kg but Less than 45 kg	3 mm 1.75mm TL-2100	3 mm	240 kg/m^3
More than 45 kg but Less than 75 kg	4 mm 2.25mm TL-2100	4 mm	260 kg/m^3
More than 75 kg	5 mm 2.75mm TL-2100	5 mm	300–360 kg/m^3

Figure 12 Sample weight to thickness ratio guideline for orthotics prescribers

4.1.2. Problem Statement

The foot can be subjected to various loading conditions, but in this study we will only consider normal walking in the mid-stance phase as in W.P. Chen et al. (2003) and normal running mid-stance phase. The mid-stance phase of gait was analyzed for this project to remain consistent with the literature as well as for the simple fact that it seemed like the logical place to start the analyses. Other phases of gait such as heel strike and possible toe off will be analyzed in future research. The kinematic constraints for the CFOs under a mid-stance loading scenario were applied as follows: 1) no movement in the back bottom heel area; 2) no horizontal movement on the back and lateral edges, leaving the arch and toe free; and 3) no movement in the vertical direction on the bottom leaving the arch free; and 4) the arch area is free to move in any direction. Finally, a uniform surface pressure load is applied using a static large deformation to the entire top surface.

These constraints were chosen to simulate an orthotic in a shoe during either walking or running at the mid stance phase. So the orthotic will have a support under the bottom side of the orthotic, but since the arch is off of the "support" there is room for that orthotic to flatten out as weight is applied to it, which is the reason for leaving the arch area open and unconstrained to deform in any direction. Then since the orthotic is in a shoe, the back side of the orthotic cannot move in any direction, but as the orthotic flattens out the front part of the orthotic may move forward or the arch may deflect slightly in the lateral direction in addition to the downward direction. That is why there is no constraints on the sides of the arch either. Finally the back outer rim and front medial rim is constrained in the lateral direction so that no sliding/twisting can occur, just as would be in a shoe. These constrains are displayed pictorially in Figures 13(a) – 13(d) below.

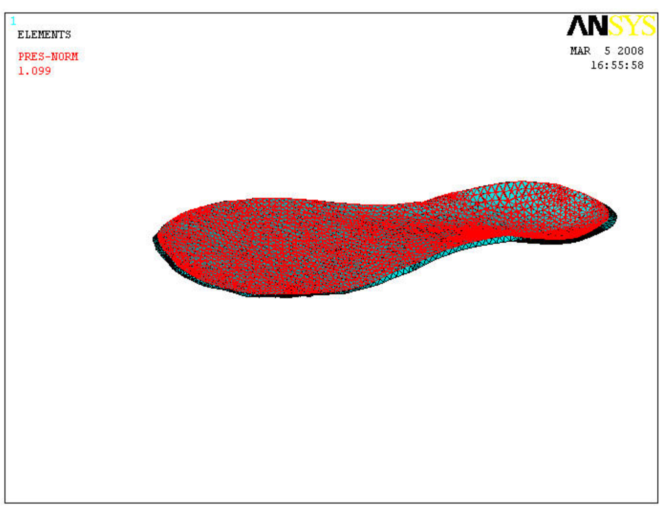


Figure 13a Graphic of applied surface pressure on FE meshed model

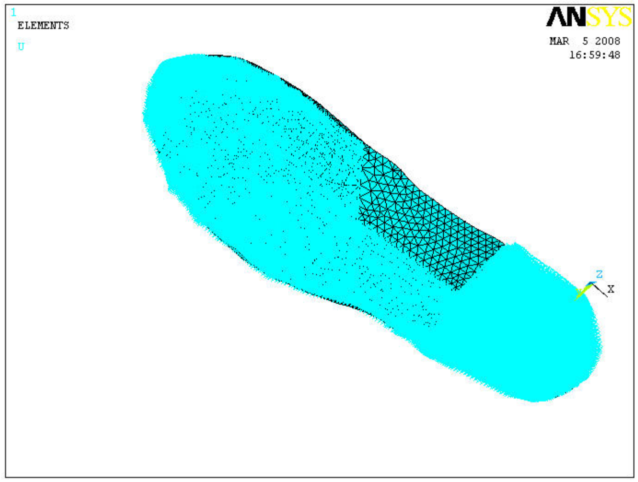


Figure 13b Graphic of constraint on bottom area; zero movement in z-direction (vertical direction)

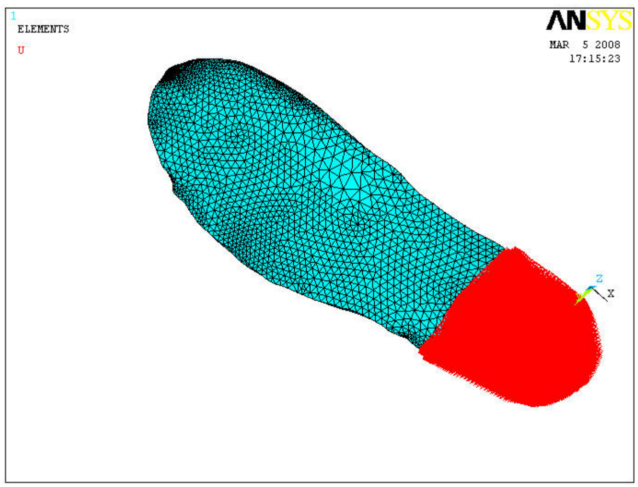


Figure 13c Graphic of applied constraint in heel area; zero movement in all DOF

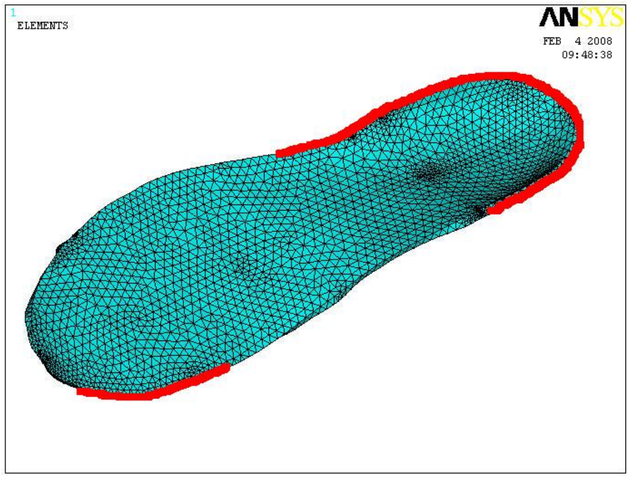


Figure 13d Graphic of applied constraint on lateral and medial edges; zero movement in y-direction (horizontal direction)

The force applied was a static nonlinear step load simulating various weight classifications for walking and running. The results of the different force applied and the resulting maximum Von Mises stress and deflection are listed in Table 2 below.

Table 2 Results of orthotic thickness vs maximum Von Mises stress and maximum deflection. These results correspond to the FEA model results (each thickness is a new model and each weight is a new condition)

Orthotic Thickness (mm)	Weight (Kg (lbs))	Equivalent Pressure (Mpa)	Max stress (Mpa)	Deflection (mm)
2	45 (100)	0.18	38.78	1.31
2	102 (225)	0.4	101.04	3.62
2	140 (310*)	1.09872		
3	45 (100)	0.18	21.57	0.6
3	102 (225)	0.4	45.87	1.4
3	140 (310*)	1.09872	141.62	4.59
4	45 (100)	0.18	13.96	0.35
4	102 (225)	0.4	30.07	0.78
4	140 (310*)	1.09872	73.4	2.27
5	45 (100)	0.18	10.12	0.22
5	102 (225)	0.4	21.87	0.49
5	140 (310*)	1.09872	53.96	1.38

An example of how the weight classifications translate into force applications to the orthotic is:140 kg (~300lbs) is 1373.4 Newtons, which is applied as a 0.55 MPa pressure force for walking and 1.09872 MPa for running.

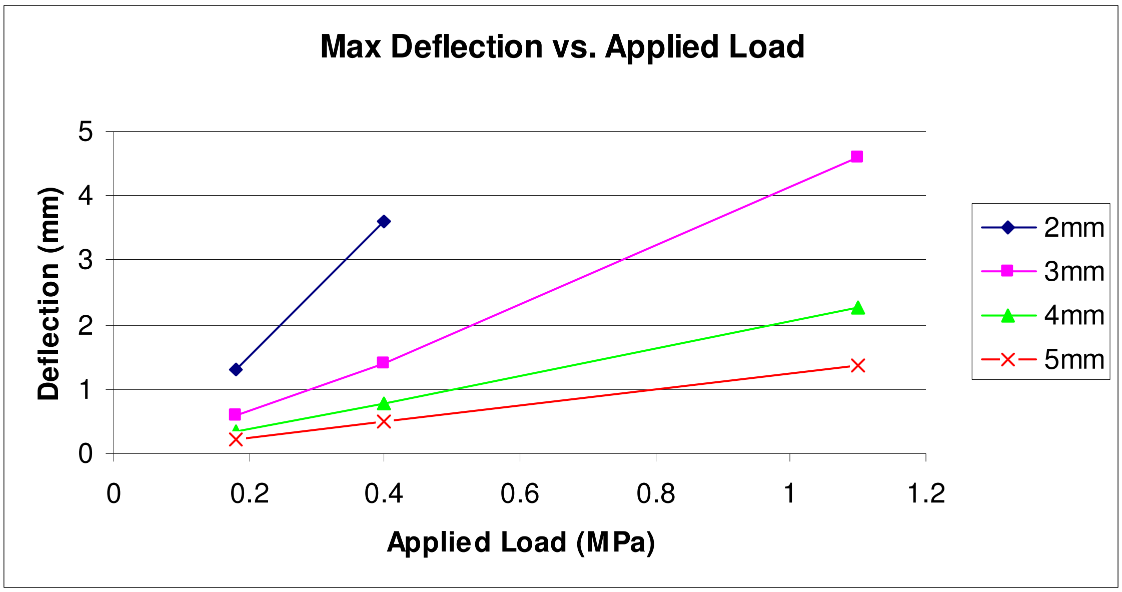


Figure 14 Results of the maximum deflection versus the applied load by orthotic thickness

Figure 14 above show the results of the maximum deflection versus the applied load by orthotic thickness. In this figure the three weight classes used were 45kg, 100kg and 140kg, which translated to an applied pressure load of 0.18, 0.40, and 110 MPa respectively. In the figure it is evident that the lager the applied load, the more deflection is seen in the arch as would be expected. The 2mm deflection line is representative of approximately 10% deflection from original arch height. Each line represents a different model which only varies by thickness. The orange line on the bottom represents the 5mm thick model, the green line right above that represents the

4mm thick model, the pink line above that represents the 3mm model and the top blue line represents the 2mm thick model. Also as expected the thicker the orthotic the less deflection is experienced in the arch. The applied load can also be separated into three distinct regions when from 0 to 0.25 MPa (the far left of the graph) is the load range corresponding to a person standing, the middle area from about 0.25 – 0.5 MPa is representative of a person walking and finally the far right form about 0.5 MPa and up is representative of a more rigourous activity such as running or carrying a heavy weight.

As stated earlier, typically a 2mm thick orthotic is never prescribed on its own, but typically as a reinforcement to another material and the model represents this by showing that this model could not withstand any loads higher than about 0.5 MPa. Then there is some load by thickness interaction seen in particular between the 3mm and the 4mm thick model. There is a small difference between the two down in the standing load range, but as we get in to the running load range (heavier applied loads), the difference between the two is very different (10% deflection versus 20% deflection). These differences are important to know in the design phase. This is explained in a more technical sense by the beam theory described below. Finally at the bottom of the graph there is not much difference between the 4mm and 5mm thick models even out in the running phase.

Preliminary clinical trial were run to gage whether the model results were within an appropriate range. These preliminary trials were run in the Biomechanics Lab of the Kinesiology Department at the University of Massachusetts Amherst. Eight high resolution Qualisys motion capture cameras were set up around a force platform in the center of the room. The Orthotic was placed in the center of the cameras on the force platform and one reflective marker was placed on the highest point on the arch. Two separate trials were run, one non-weight bearing before each weight bearing trial to get a baseline measurement of the arch height. Each weight bearing trial was run with a 70kg person standing on the orthotic. These trials were run to examine the change in arch height when body weight is applied to the orthotic. The FE model deflection ranges fall in line with the numbers from the preliminary clinical results. The model and clinical preliminary trials differ slightly by about 7% – 10% due to two main factors. First, there is some error in the clinical trails and further trials will allow for more accurate numbers. Secondly, the actual FE model is slightly different than the actual geometry of the physical orthotic. In the model the polypropylene layer extends all the way out to the toe region making the material appear slightly stiffer; whereas the actual physical model support phase (polypropylene layer) only extends to just past the arch as seen in Figure 15 below.

(a)

(b)

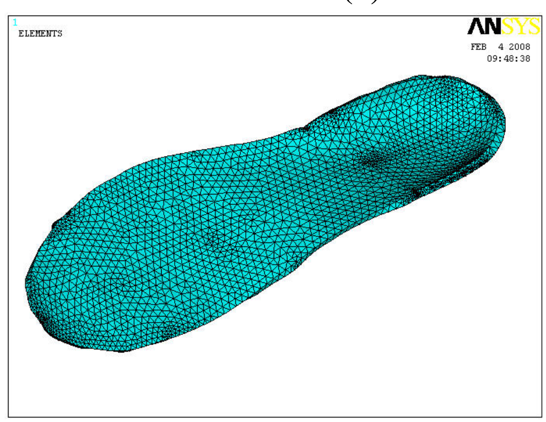


Figure 15 (a) Picture of physical orthotic used for clinical preliminary clinical trials, (b) FEA model used for analysis

Further biomechanics lab clinical trials will be run to validate the model. Figure 16 below shows the results of the deflection versus the applied load including the two data points from the preliminary clinical trials.

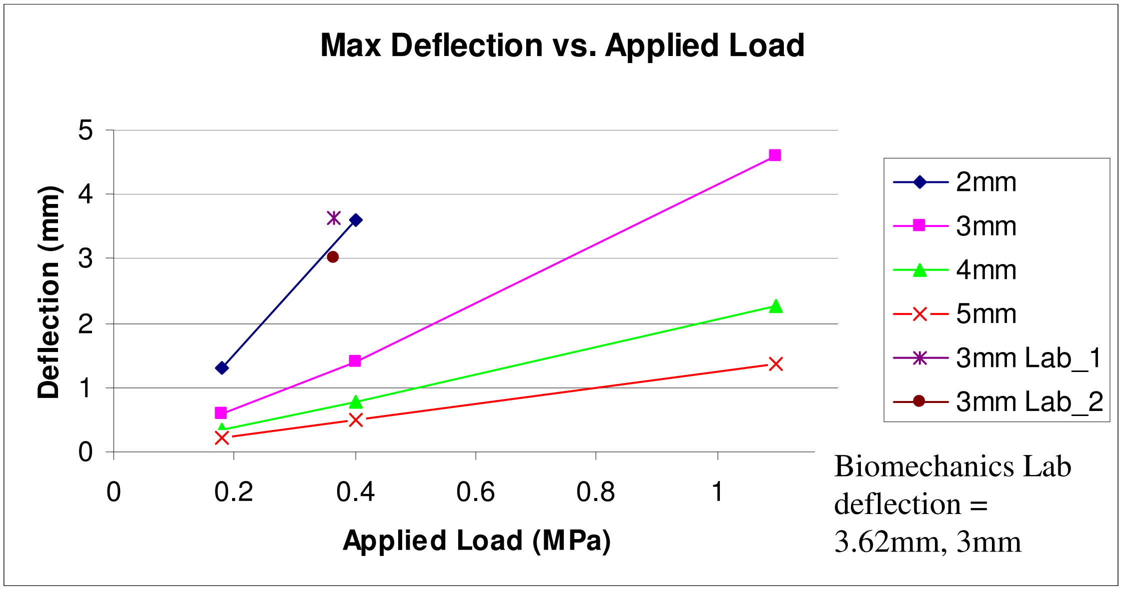


Figure 16 Results of the maximum deflection versus the applied load by orthotic thickness including preliminary clinical trial results

Figure 17 below shows the maximum stress versus the applied load by orthotic thickness and Figure 18 shows the stress distribution and deflection distributions of the finite element model results. As one can see in the figures above, as the applied force increases, the resulting stresses also increase. For each model the maximum stress increases as the applied load increases. This is in line with the expected results because the thinner orthotic will have more deflection and will act as more of a shock absorber (absorbing more of the stress), while the thicker orthotic will be stiffer and more supportive, but will deflect less and therefore absorb less of the shock.

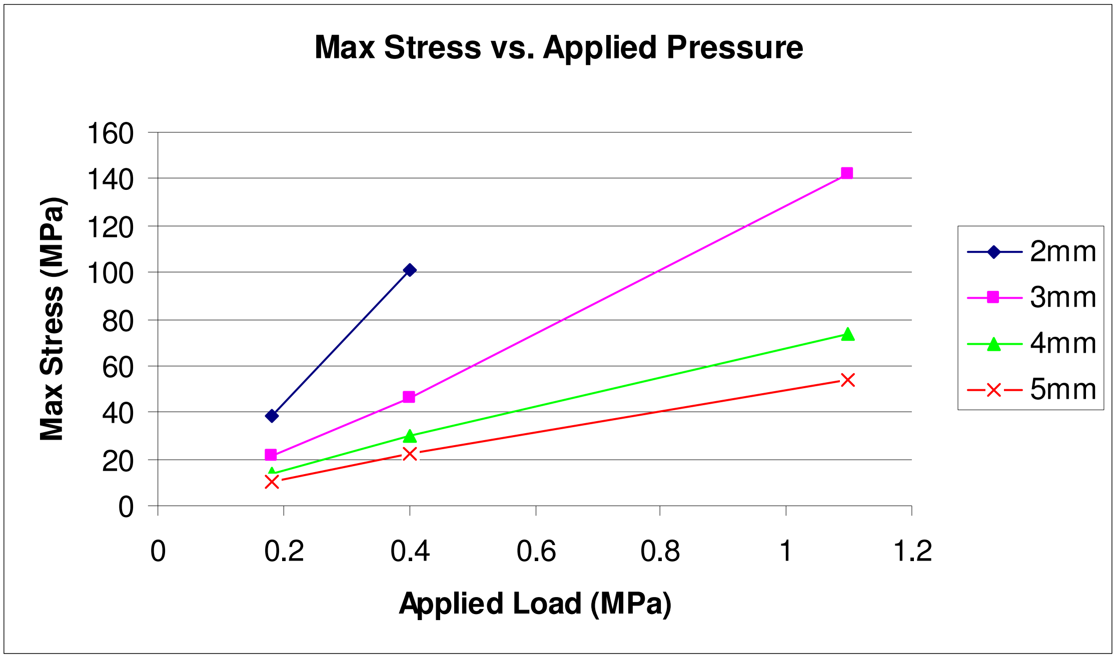
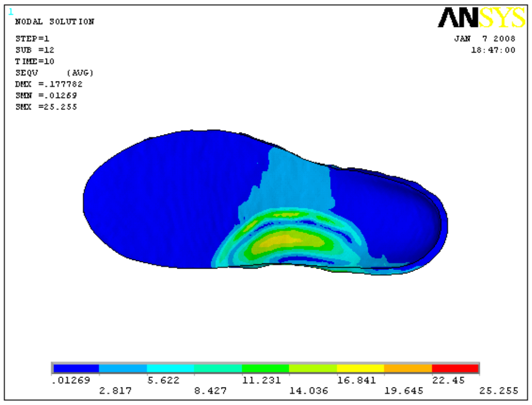


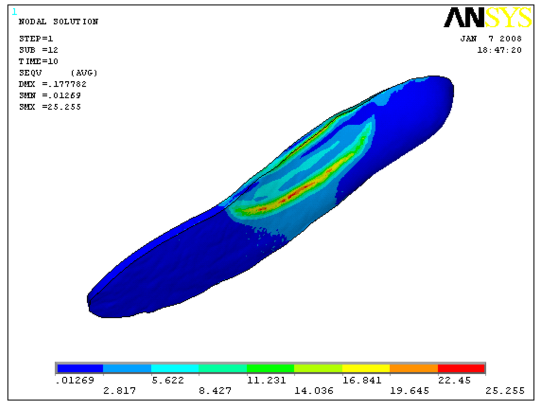
Figure 17 The maximum stress versus the applied load by orthotic thickness

In the stress and deflection distribution result plots below show that the maximum stress areas are right around the edge of the arch and the inside rim of the arch as would be expected. This is important for prescribers to know, since the arch is the area expected to perform. This can also be important when taking into account the long term effects that the high stress areas will cause on the longevity of the orthotic.

(a)



(b)



(c)

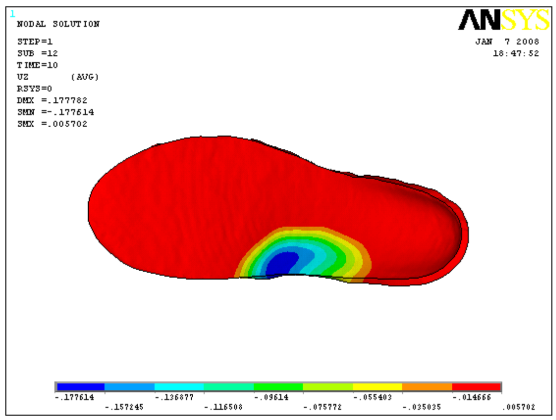


Figure 18 (a) Stress distribution top view, (b) stress distribution bottom view, and (c) deflection distribution top view

4.1.3. Beam Theory

Beam theory is studied to verify the numerical analysis results. The cross section of the beam is rectangular with the width of the beam, b (the dimension parallel to the bending axis), and the height of the beam, h (the dimension perpendicular to the bending axis). A beam’s moment of Inertia (I) is a measure of its stiffness with respect to its cross section and its ability to resist bending. Moment of Inertia (of axis perpendicular to load) is a significant variable in the determination of beam deflection: as I increases, bending decreases where $I = bh^3 / 12$. The deflection (vertical displacement as a result of an applied load) of the simply supported beam is defined by $\Delta = PL^3/48EI$ and the maximum stress $\sigma_{max} = Mc/I$ ($= bh^2/6$).

The arch of an orthotic can be looked at as a simply supported beam and modeled as a structure that carries a load between its two supported ends. Here, the thickness of the orthotic will correspond to the height of the beam's cross section, h. Therefore, as the thickness increases, the moment of inertia will increase, and the corresponding bending deflection will decrease. Similarly, the maximum stress will decrease as the thickness increases. This is evident in the results of the model: as the orthotic thickness increases, the deflection and stress decrease at a proportional rate. This can be seen in Figures 19(a) and 19(b) below.

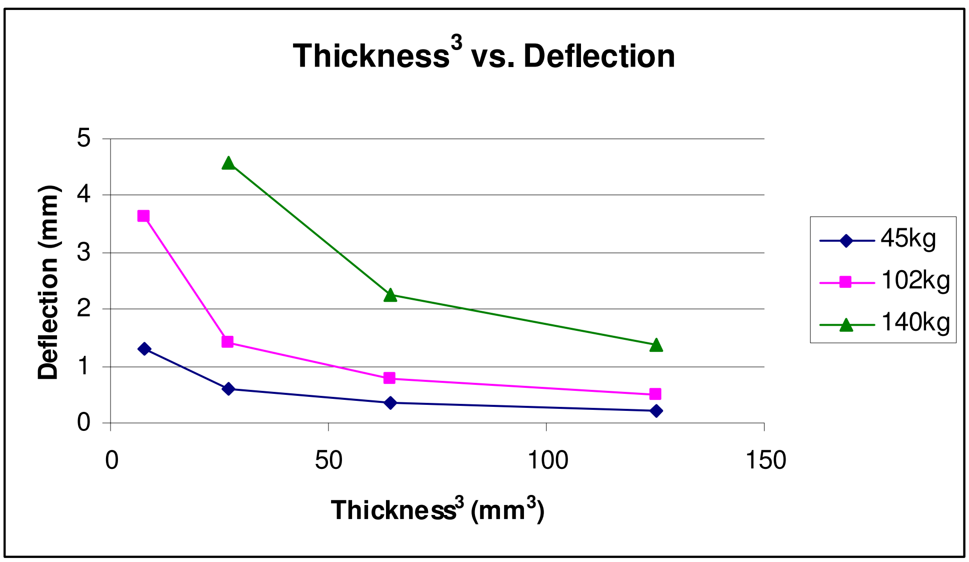


Figure 19a Maximum deflection by orthotic thickness cubed as a function of applied load

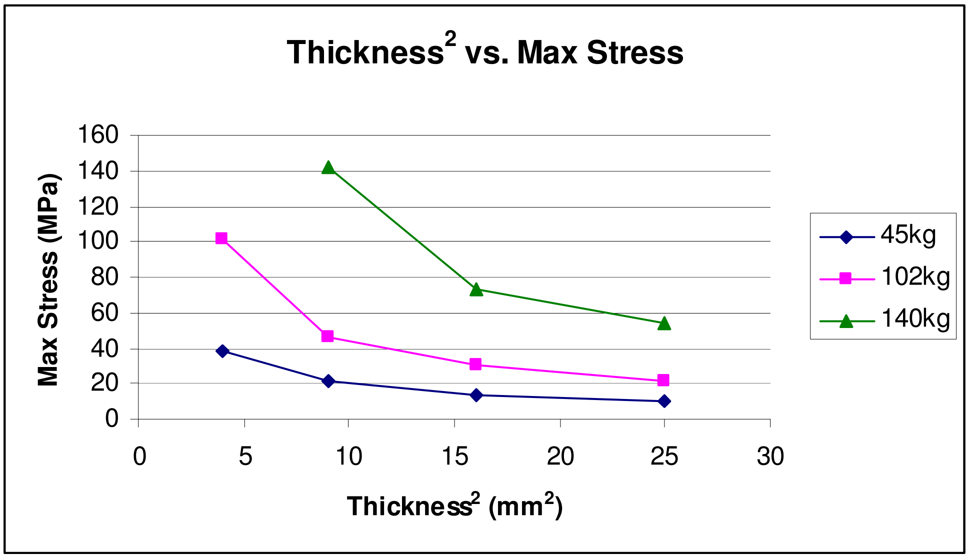


Figure 19b Maximum Stress by orthotic thickness squared as a function of applied load

In Figures 19a-19b, the trends are consistent with the beam theory as stated above. There is a slight deviation from the straight line at the large load/small thickness area because it seems the material deforms beyond its elastic zone and both the deflection and stress differ slightly from the trend.

4.1.4. Early Complications

There were four early complications in this research: conversion of the scanned data; altering of the geometry; material property estimation; and constraints. The first issue was converting the scan data to a usable format. Although scanning the image greatly simplifies the ability to have an accurate geometry, the process of converting the scan data to a usable form by ANSYS can be challenging without the proper tools. Many processes were attempted before finding the 3-Matic program to convert the data. A custom Matlab program was written to create triangular facets, and then brought into ProE to make a solid model and then export as a *.iges file for import into ANSYS.

Unfortunately the IGES file exported from ProE was not readable by ANSYS. Next an attempt was made to import it into and then out of ANSYS Workbench. Workbench was unable to mesh the file in order to complete the conversion to ANSYS Classic. Finally, the 3-Matic program by Materialise, Inc was discovered and the file was able to be converted using the process laid out in the appendix. Once the file was brought into ANSYS, however, the geometry was not able to be altered since the file was meshed in 3-Matic using the data points leaving no areas or volumes, only nodes and elements. Finally, the model was brought back into 3-Matic and a file was created with defined areas to use in ANSYS.

The second main issue had to do with the material property estimation. After getting the stress strain data from the materials testing, the conversion formula was published incorrectly in the Finney et al., paper being used. The paper stated that if $(S/2)*(a-a^{-2})$ is plotted against a^{-1}, the resulting plot will be a straight line with c_2 as the slope and (c_1+c_2) as the intercept at $a^{-1} = 1$. Whereas the actual equation being plotted against a^{-1} is $(S/(2*(a-a^{-2}))$. This resulted in the values being off and the material property definitions being innacuarate, causing the model to behave erroneously. At this early stage, our research did not find any other publications which had used Mooney-Rivlin to define their nonlinear material properties so a comparison in numbers to some standard values was not possible.

Finally, the constraints that were initially defined were constraining the outer rim of the orthotic in lateral direction including the arch area. Since that constraint did not allow the model to deflect in the lateral direction, the polypropylene deflection results suggested that the polypropylene material was stiffer than it actually was. After conferring with experts in the field and removing the lateral constraints on the arch section, the model responded very well. The results were much more accurate and compared favorably to the preliminary in lab trials that were run in the Biomechanics Lab.

In the future another alteration would include cutting the top (toes) portion of the polypropylene layer down in the way that the real CFO's are made, so as to cut down on the inaccurate stiffness further.

CHAPTER 5

SUMMATION & FUTURE WORK

5.1. Summary

In summary, the aim of this research was to make a significant contribution in the understanding and development of CFOs by creating an accurate finite element model of a CFO using FEA. Complicated nonlinear material property estimations were attained through experimental material testing and data conversion. Complex geometry was attained through the use of laser scanning techniques and this coupled with the nonlinear material property estimations, yielded an accurate FEA model of a CFO. The current model is performing as expected showing that as the load increases, the more the arch deflects, as well as that the thicker the orthotic is, the less the arch deflects. The maximum stress areas are seen in the inner and outer edge of the arch region, and both the deflection and stress distributions are as expected.

Creating this model allows for the ability to increase the knowledge and understanding of the effects CFOs have on human movement and performance. The creation of an accurate FEA model will eventually allow for the simulation of various orthotic and loading conditions, changes in material properties, and foot deformities by simply altering model specifications. These FEA CFO models can form a basis for subsequent design of a new generation of CFOs which may lead us to believe that patient compliance to certain designs and desired biomechanical outcome would improve. These models may also be of interest to prescribers of orthotics in the treatment of foot, leg, joint, hip and back problems to complement current clinical practices.

5.2. Future work

Suggested future work topics can include: 1) Alteration of the geometry by further cutting the polypropylene layer down to the more accurate length; 2) a full clinical trial run to validate the model; 3) other polypropylene material properties compared by applying the 2mm and 4mm Mooney-Rivlin material model to the appropriate thicknesses; 4) other material layers added to the models, in particular the soft Spenco material added to the top of the polypropylene and the heel post materials (EVA and Topy rubber); 5) a comparison between the heated and raw materials analyzed; and, finally, 6) different phases of gait will be analyzed such as heel strike and possibly toe off. As can be seen, there is much work to be done in the field and this model lays the foundation for those studies to be undertaken. In the future it seems possible to produce interactive programs where a clinician can complement traditional methods with finite element models so as to produce orthotics that are optimal for each particular client.

APPENDIX

STEPS FOR CONVERTING THE SCAN DATA TO A SOLID MODEL

1. Import the scan data
2. Isolate the points of interest (Point Cloud tab > Use the different tools and views to select and delete garbage points)
3. Once you have the points you need, you can mesh them (Point Cloud tab >"Mesh Range Data")
4. this works best when all points should be directly connected to their adjacent points;
5. a hole filling strength of 5 was used where more is more agressive filling
6. You will now have an STL mesh of your geometry,
7. It may be rough, so it will need to be smoothed
8. Smooth the surface (Mesh tab > Smooth)
9. you can play with the smoothing parameter,
10. A strength of 0.7 to 0.9 was used
11. If some excess fringe data has been captured (i.e. the STL mesh extends slightly beyond where you intend it to), go to Marking tab > Mark Edge and use this to highlight all the triangles on the fringe.
12. If necessary you can use the Expand Marked button to select another "ring" of triangles if your fringe is that large.
13. Then just hit "delete" on your keyboard.
14. Now we need to smooth the contour that remains. Curve tab > Smooth curve. Select the bad contour that defines our surface.
15. Only 3 iterations were done, more than 4 or 5 is not recommended.
16. You may need to use this tool more than once to get a decent smoothing, though over-using it eventually starts to degrade the curve.
17. Look at the edges of the surface now; we want it to be clean and smooth so that the offset function doesn't create artifacts.
18. If you need to, you can mark and delete triangles, and also create triangles (Mesh tab > Create Triangle then click on 3 points)

19. Once the surface is ready to be offset, go to CAD tab > Offset surface.

20. Select your distance and direction

21. remember internal equals in the direction of the red faces

22. You may need to clean up the result of the offset as we did in step 8.

23. To fill holes use CAD tab > Fill Hole Normal

24. Once the geometry is clean, we can remesh. Remeshing tab > Create Inspection scene.

25. On the inspection scene, choose what shape measurement you want

26. The H/B normalized shape was used in this project

27. Make sure the histogram is showing the shape measurement.

28. Now use the Automatic Remesh function;

29. It is suggest to raise the quality threshold by 0.1 each time

30. it is not necessary to go higher than 0.3 unless you have a very specific reason.

31. Once the remeshing is done

32. Remeshing of specific difficult areas was done by marking the area and then doing a local remesh of "Marked Only"

33. "Quality Preserving Triangle Reduction" can be done so that you don't have too many surface elements.

34. Last step was to export to ANSYS

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