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Compliant pediatric prosthetic knee

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Compliant Pediatric Prosthetic Knee

by

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ABSTRACT

We have designed and examined a compliant knee mechanism that may offer solutions to problems that exist for infants and toddlers who are just learning to walk. Pediatric prosthetic knees on the market today are not well designed for infants and toddlers for various reasons. Children at this age need a prosthetic that is light in weight, durable, and stable during stance. Of the eleven knees on the market for children, all but three are polycentric or four-bar knees, meaning they have multiple points of movement. Polycentric knees are popular designs because they offer the added benefit of stable stance control and increased toe clearance, unfortunately this type of knee is often too heavy for young children to wear comfortably and is not well suited for harsh environments such as sand or water, common places children like to play. The remaining three knees do not offer a stance control feature and are equally vulnerable to harsh environments due to ball bearing hinges. Compliant mechanisms offer several design advantages that may make them suitable in pediatric prosthetic knees – light weight, less susceptible to harsh environments, polycentric capable, low part count, etc. Unfortunately, they present new challenges that must be dealt with individually. For example compliant mechanisms are typically not well suited in

applications that need adjustability. This problem was solved by mixing compliant mechanism design with traditional mechanism design methods. This paper presents a preliminary design concept for a compliant pediatric prosthetic knee. The carbon fiber composite spring steel design was first built and then evaluated using Finite Element Analysis. The prototype's instant center was plotted using the graphical method. From our analysis position, force and stress information was gathered for a deflection up to 120º. The instant centers that were plotted indicate that the knee has good potential in offering adequate stability during stance.

Chapter One

Introduction

The objective of my research is to design an adjustable compliant pediatric knee. This knee will better suit children by lowering the over-all weight of the knee while keeping the benefits of heavier rigid link knees. An adjustable knee design is described and its position, force, and stress properties are analyzed.

Figure 1.1. Adjustable compliant pediatric knee prototype

Motivation

Our goal is to improve current pediatric knee technology by designing a knee having lower weight, reduced cost, and increased functionality. By using a *compliant-mechanism-based design*, we are capable of potentially lowering the weight of the knee substantially, and in doing so we will allow younger children to more comfortably wear their prosthetic. Some pediatric knees on the market today offer great stability as well as some of the benefits found in adult prosthetic knees but they are typically too heavy for young children to wear for long periods of time. The benefit of having a child wear their prosthetic longer is that it allows them more time to adapt to wearing their device. By giving a child time to adapt to wearing a prosthetic prior to walking you promote the child to walk at a younger age. Standing and walking up-right promotes the child's ability to move about with their hands free to grasp objects and interact with new surroundings that were otherwise unattainable by crawling. This interaction with their new found world stimulates positive mental and physical growth. [1]

To lower the weight of the overall prosthetic limb, a peg leg is typically prescribed to give the child a way of walking that is easy to use but this oftentimes results bad habits acquired from its use which are difficult to rid. The peg type leg requires that it be adjusted to a length shorter than the sound limb to make it useful. This misalignment forces the child to walk as if one foot was constantly in a hole. This creates problems for the child's gait, forcing the hips to excessively deviate from normal gait to compensate for the difference in height between the two limbs. This constant misalignment is not only bad for the body, it also produces habits which make adapting to better technology later in life become more difficult. By instilling good habits from an articulated knee, like proper hip alignment and good weight transfer, you assist the child now as well as later in life.

A better knee design could potentially offer a reduction in shock that is associated with many prosthetic knees. Because our design is compliant, i.e. allows for elastic deflection, shock felt from ground reaction forces can be greatly reduced. Another benefit our design is the continuous connection that is made with the lower leg and is created by allowing torque to be transmitted through the knee. Prosthetic knees are typically unable to transmit torque through their ball bearing hinges. This feeling through the knee allows the patient to have an increased confidence as to where their lower leg is located during gait, reducing the fear and potential of falling.

Children are rough on their bodies and especially rough on prosthetics. Children are notorious for playing in places that most prosthetics would not be able to endure. For this reason we intend to design a knee that would be particularly suited to children. The compliant knee would be made impervious to sand, water, and other corrosive agents. By eliminating ball bearings or other sensitive equipment, the knee can be submerged in liquid or exposed to sand without causing serious damage to its function. Our design could potentially out last other designs on the market simply because it does not have limitations as to where it can be used. Because children have a mind of their own it is tough to keep them away from potential hazards to their prosthetic.

Given the potential issues that children present to prosthetic devices, small alterations to the design could potentially create a product that would out perform current technology. The compliant knee offers more function and versatility than the competition by offering a lighter weight design with additional features only found in heavier adult prosthetics. Our design offers shock absorption as well as a continuous connection to the prosthetic's lower limb. The knee would be impervious to environmental hazards making it a better choice for small children. For these reasons we feel compliant knees could potentially take over the market for small children.

Contributions

- Design and examined an adjustable pediatric compliant knee prototype
- Created a prototype weighing less than five ounces.
- Formulated a method for analyzing the rotation and translational motion of the compliant knee using nonlinear FEA accomplished through calculation and plotting of the instant centers of the knee's rotation
- Developed a method for simultaneously calculating the external reaction forces and internal stresses of the knee for the anticipated region of motion for a deflection of θ for $0^\circ < \theta < 120^\circ$.

Chapter Two

Background

Phases of Gait

Wheeless' Textbook of Orthopedics (2000) [2] describes the phases of gait in two different sections, the stance phase and the swing phase. In stance phase, the foot is purposefully in contact with the ground, while in swing phase, the foot is suspended in air. The stance phase comprises the larger segment of the entire gait cycle, approximately 60 percent of the cycle, and can be broken down into five separate parts [2]. When an individual demonstrates normal gait, there is a level of symmetry to the process and there is a consistency to the balance between the 60 percent stance phase and 40 percent swing phase [2]. When variations in the ratio of stance phase to swing phase exists, abnormalities can be noted in the gait.

The stance phase, illustrated in *Figure 2.1*, includes initial contact (made by the heel of the foot following an initial upswing), loading response, in which the foot balance is secured and alignment occurs between the hip, knee and ankle to place weight on the foot, mid-stance (entirety of the body weight is on the foot, knee is extended and the ankle is neutral), terminal stance (the end stage and initial lifting), and toe off, which ends at the toe lift [2].

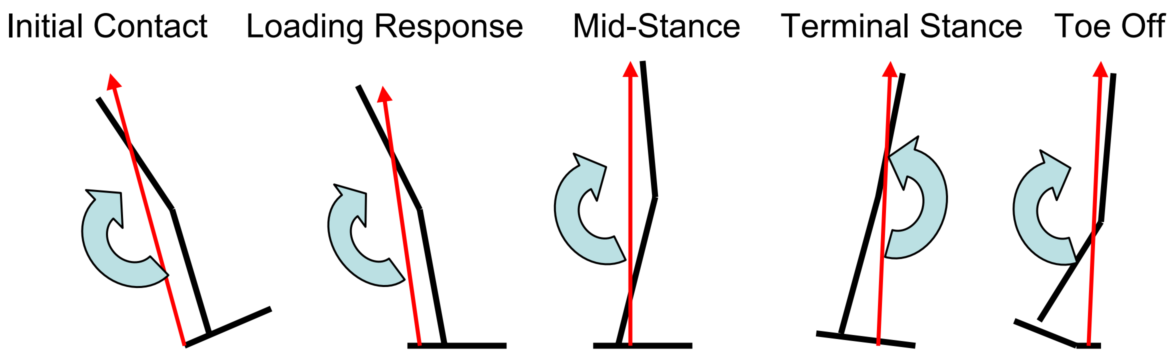


Figure 2.1. The five phases of stance

The key to the stance phase is resistance to falling. Stability and the alignment of the trunk over the base of support which in normal cases, consist of the alignment of hip, knee and ankle over the foot are imperative to the progression from the stance phase to the swing phase. Stability refers to the balancing of the center of mass over the base of support and also resistance to knee buckling, both of which often result in collapse of the amputee. Knee buckling is often caused by the flexion moment, illustrated in *Figure 2.1,* which is present during all phases of stance with the exception of terminal stance. This exception is caused by the ground reaction force falling anterior to the knee joint.

The swing phase is especially important when understanding the dynamics of a prosthetic knee and the functionality. The swing phase comprises 40 percent of the total gait process, and begins with the toe lift-off [2]. After the stance phase, the toe lifts off the ground behind the body. The leg then begins to move forward and the foot extends, in alignment with knee flexion. Subsequently, the forward motion of the body allows for the swing of the leg forward. Key to this, though, is the ability of the toe to clear the ground. In unbalanced prosthetics, it is not uncommon for individuals to purposefully use pelvic rotation to compensate for the inability to move the foot or clear the toe, and so prosthetics that are improperly fitted can result in major gait complications that arise from attempts to adjust for the swing phase [3].

The concept of prosthetic gait synergy relates the functionality of the prosthetic to the capacity to maintain normal gait through motor control. Essentially, the prosthetic has to work in correlation with existing body structures to allow motor control, create repeatable patterns of muscular activity and work in concert with body kinetics [3]. Movements that are synergistic actually require minimal immediate thought processes or neural control and so have been identified as "automatic" in their performance [3]. One of the keys to prosthetic development is the ability to integrate synergistic movements and support independent control of gait in order to create balanced movement.

Prosthetic gait synergy, then, is "the best possible gait pattern of an amputee with a given type of prosthesis/prostheses. Not every deviation from normal gait is considered a component of the prosthetic gait synergy. Only deviations that remain apparent after proper residuum conditioning, socket fit, and prosthesis alignment and adjustment, and after the amputee becomes accustomed to the prosthesis" [3].

Dundass, Zao and Mechefske (2003) studied the use of a hydraulic knee controller in a transfemoral amputee subject, and identified prosthetic deterioration as a significant issue for transfemoral amputees. This study is especially significant for addressing the issue of pediatric transfemoral amputees because it is likely that deterioration will occur because of the longevity of prosthetic use and the need to identify the impacts of growth from the transfemoral amputee.

Dundass, Zao and Mechefske (2003) identify kinetic gait analysis as one of the significant measures of the internal and external factors that influence motion. These researchers maintain that there are some factors that influence the outcomes of assessment kinetic gait process, the least of which is that stride differentials and the fact that many amputees do not put their full body weight on their prosthetic limb during the stance phase, suggests that these are variables that should be considered when creating prosthetic devices [4].

What is evident is that the major purpose of the prosthetic devices, including prosthetic knee components, is the effort to recreate human knee and leg function and the synergy between physiological components and the prosthetic utilized. In doing this, it is valuable to consider the kinematics of the knee and the influence of the knee structure on the development of prosthetic prototypes.

Kinematics of the Anatomical Knee

The structure of the knee is not unlike the structure of the elbow, where three distinct bones come to a central point. The femur, tibia and fibula are static and strong support mechanisms for the musculature and the load forces that result from movement like taking a step are “transmitted over the hyaline joint cartilage, allowing a smooth and easy motion of the joint surface” [5].

“The femorotibial joint surfaces are noticeably incongruent and offer just a two-point contact area. The hip joint, in contrast with its spherical head in its socket, as well as the ankle, have large contact surfaces for load transfer” [5]. This clear differentiation between the structural elements of the lower extremity raises the question of how the effective link between the skeletal components and the musculature is reached and how the formation of the central pivot of the knee is realized [6].

Essential to the formation of the knee itself are two cruciate ligaments and they assist in the creation of a mechanical system of “crossed four-bar linkage” which allows the knee to move in “three rotations (extension-flexion, external rotation-internal rotation, and varus-valgus rotation) and in three translations (anteroposterior, mediolateral, and compression-distraction)" [5]. When prosthetics function in the place of the structure of the knee, it is necessary to develop systems that mimic the rotational component of the knee.

Because of the necessity for flexibility and a range of motion, the ligaments, musculature and support components of the knee work in concert to allow for both rotation and stabilization. The prosthetic must represent the capacity for load transmission from the femur to the tibia (or the structural equivalents in the prosthetic). This must include an identification of methods to absorb or reduce the peak load forces and their impacts on the structures surrounding the joint.

Instantaneous Center

The instantaneous center (IC), in a plane or in a plane figure which has motions both of translation and of rotation in the plane, is the point which for the instant is at rest. (Webster's 1913) A two dimensional instant center is a location within a plane that a body instantaneously rotates around.[7] At this instant all motion within the body is traveling perpendicular to the line that connects the point of interest on the body to the IC. *Figure 2.2* illustrates how the instant center of the rotation of a wheel rotating on flat plane is always located at the point of contact with the surface. Notice how the motion of points *A* and *B* are traveling perpendicular to the line that intersects the instant center. This is the case because at any instant all locations on a rotating body travel perpendicular to the IC.

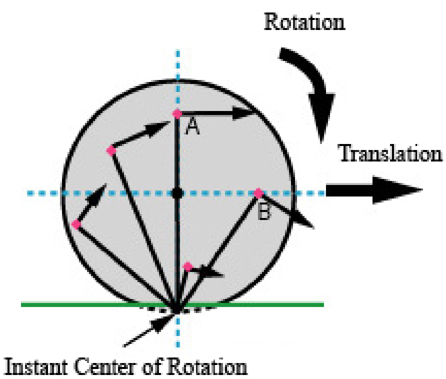


Figure 2.2. Instantaneous center of rotation of a rolling wheel

ICs are calculated in order to fully understand motion characteristics when designing mechanisms. This is especially true for mechanisms designed to be used as prosthetic knees. The IC of a prosthetic knee determines how stable the knee will be during stance phase. The proper IC location is capable of assisting in toe clearance as well as provide vital control feedback, i.e. gait synergy, to patients with short residual limbs. i.e. if the IC is placed near the end of the residuum the amputee is afforded greater control over the prosthesis. For these reasons it is important to understand where the ICs are located during various angles of flexion. The IC for a single-axis knee is located at the center of the joint and doesn't change during rotation. For more complicated, polycentric knees, knees without a fixed center of rotation, this is not the case. During their full range of motion, the IC undergoes constant change.

Polycentric Knees

There are currently more than 100 different types of prosthetic knees on the market today, eleven of which are pediatric knees. Of the many different types available there are two main types of categories that they can fall into, single axis and polycentric. Single-axis knees are designed with one pin joint located near the position of the anatomical knee. Four and six-bar knees fall under another category known as polycentric and have irregular patterns which the ICs follow. Polycentric knees, like the Mighty Mite® illustrated in *Figure 2.4,* have traits that are beneficial for a various reasons. Polycentric knees increase toe clearance during swing phase, they decrease the length of the upper-leg and appearance of abnormality during sitting. Polycentric knees are also particularly stable during stance phase.

Figure 2.3. MightyMite® 4-bar knee by Fillauer

Issues Facing Pediatric Prosthetic Users

Wilk et al. (1999) noted that the common practice with very young children with amputations above the knee is to fit the child with a prosthetic of some kind as soon as possible, as soon as the child is able to pull to a standing position, which generally occurs between the ages of 9 and 16 months. It is common to begin gait training with very young amputees utilizing non-articulating prosthetics, i.e. no knee joint present, because young children can learn to walk without a functional knee easier with the non-articulating prosthetics, and because commercially available pediatric prosthetic devices with articulating knees are not generally made available [8]. The key consideration here is that when the transition to an articulating knee is made, there is often a longer period of adjustment [9]. In addition, researchers have noted that knee prosthetics have relatively low durability and that young children who are so active may require multiple prosthetics that can be cost prohibitive.

Researchers have recognized, though, that whether cost-prohibitive or necessary, an increasing focus on the past processes and the negative implications for pediatric patients should be noted. It is not uncommon, for example, for young children to be fitted with prosthetics without articulating knees and then go many years before a new prosthetic is provided. Children in these situations often incorporate different methods of gait control that are difficult to change after years of use. As a result, the benefits of integrating improved systems can be demonstrated through the use of articulating knee prosthetics from the onset.

The result of the use of non-articulating knee prosthetics for young children is that they often result in gait variations that can be difficult to address when fitting them for different prosthetics later in life. Wilk et al. (1999) maintained though, that children who had prosthetics without articulating knees prior to the use of articulated systems demonstrated immediate adaptations in their gate, and improvements in hip flexion-extension, unbalanced pelvic motion and improvements in balance.

Wilk et al. (1999) argued that children can be fitted prior to 18 months of age with articulated knee prosthetics and that these prosthetics can reduce the chance of problematic gait deviations that cannot be corrected and improve the chances that pediatric patients will demonstrate normal childhood activities. As a result, these authors argue in favor of the development of improved prosthetic devices for young amputees and the discarding of the old practice of the use of non-articulated prosthetics in very young children [8].

Design Criteria

In the paper written by Andrysek "Design Characteristics of Pediatric Prosthetic Knees", functional requirements are examined within the pediatric user community.[9] The main functional requirements are comfort, fatigue, stability, and resistance to falling. Sitting appearance and adequate knee flexion are of lower importance. Stance control and toe clearance were of the highest importance. These criteria were used to rate five knees on each of the five children in the study. From the study it is noted that a single-axis knee is equally suitable in meeting the highly and averagely important functional requirements, meaning it is not as important for a child to have all of the benefits of technology, for proper gait as long as the importance is on function and not on appearance.

The article explains that 8 of 11 commercially available knees are four- or six-bar configurations. They are said to be highly acceptable due to there ability to control stance phase, increased toe clearance, and offer a more natural knee location. The disadvantages are that they are heavier than the single axis counterparts. The added technology hinders rather than helps especially with small children due to the added weight. The results of the study suggest that a single axis knee with particular axis placement and a stance control mechanism satisfy the design parameters similarly to polycentric knee joints. The study was set up to design a more effective, and less complex knee to minimize the size and weight for the pediatric needs. The study concluded that a single axis knee has demonstrated the required function while fulfilling the functional requirements, which could result in a highly functional, less complex, knee joint. [9]

Compliant Mechanisms

It is possible that the benefits of polycentric knees can be achieved without prohibitive weight problems by using a design technology known as “compliant mechanisms.” A mechanism is a device that is intended to transfer motion, force or energy. Mechanisms are typically rigid body and are made up of rigid links connected through movable joints. Mechanisms are common in nearly every part of our lives from a pair of scissors to the steering system on your car. Mechanisms can be used to increase or decreases mechanical advantage depending on the purpose of its use. The scissor jack, shown in *Figure 2.5,* is an example of a mechanism that increases mechanical advantage and allows the user to lift objects much heavier than would be possible without it. Although mechanisms are capable of increasing output velocity or forces with respect to the input velocity and force, they are not capable of increasing the energy output of the system. For example when the output velocity is higher than the input velocity the output force must be lower than the input force and when the output force is higher than the input force the opposite is true. This ability to transfer energy is not entirely efficient though. Energy is wasted by the mechanism due to losses in the system typically caused by friction. These losses can be small or large depending on the type and design of the mechanism.

Figure 2.4. Scissor jack mechanism used to increase mechanical advantage

A compliant mechanism is designed to do the same basic task as a traditional rigid link mechanism, transferring motion, force, or energy, but it does so by allowing deflection of one or more flexible members instead of using movable joints. One example of a compliant mechanism is the plastic hinge found on various plastic bottle caps shown in *Figure 2.6*. The motion of the cap is similar to that of a single axis hinge with the addition of bi-stable positioning. In other words the cap favors two distinct positions along its path of rotation. The advantage of this is that the bi-stability keeps the cap open and out of the way while pouring the contents from the bottle. The ability to fall in either of two separate positions is achieved by the kinematics of the system in combination with the potential energy stored within the flexible segments.

Figure 2.5. Plastic bottle cap with compliant hinge

There are many reasons why compliant mechanisms have advantages over traditional rigid link mechanisms. Compliant mechanisms have the ability to greatly reduce the total number of parts and assembly steps needed to create a mechanism. These compliant mechanisms are oftentimes created from one injection-molded piece. An example of this type of fully compliant mechanism is a plastic injection molded box shown on in *Figure 2.7*. A similar design, shown in *Figure 2.8,* uses traditional moving hinges. Notice how the compliant box uses only one piece of material for the entire mechanism. This single piece mechanism fulfills the same task required from a pensile box but does not require any assembly after the initial molding process. The reduction in required parts and assembly steps for production can greatly reduce the costs and time spent fabricating mechanisms.

Figure 2.6. Fully compliant injection molded plastic box

Figure 2.7. Pencil box with movable metal hinges

Another key feature of compliant mechanisms is the reduction in movable joints. This greatly reduces the amount of internal motion within the mechanism which can potentially reduce the wear, friction, and need to lubricate parts. These attributes can be favorable in situations where the environment is corrosive or harsh. By lowering the amount of wear in a mechanism it is possible to extend the life of the device dramatically. The reduction in total friction losses can be favorable especially in the case of the pediatric knee. The reduction in friction losses can lower the amount of energy expelled by the child during its use which could extend the amount of time a child can comfortably wear the prosthetic. The ability of compliant mechanisms to operate without lubrication can be favorable for extending wear capabilities as well as removing the need to service the device for this cause. In addition to these benefits a reduction in vibration and noise is possible due to the lower friction and absence of additional moving parts.

Compliant mechanisms utilize flexing of segments to generate motion and in this flexing they store potential energy. This potential energy storage can easily be used as a built in feature of the mechanism. By storing energy in the form of strain energy a compliant mechanism can generate force feedback similar to externally mounted springs without the need to add additional parts. A good example of this is a bow and arrow system. The potential energy is initially put into the system by the pulling from the archer's arms. This potential energy within the bow is later released and turned into kinetic energy in the arrow once the bow is fired.

Another feature of compliant mechanisms is the reduction in overall weight of the device. Rigid-body mechanisms require that the links within the system remain rigid and therefore require higher strengths and additional material to build. By allowing the mechanism to flex under pressure, you lower the need for heavier rigid structures. This feature is especially beneficial to the project at hand. By lowering the weight of the pediatric knee you allow younger children the ability to comfortably wear a prosthetic as well as increase the total time one can be worn by any patient.

Compliant mechanisms are favorable in many situations that we may not naturally consider them for. The desire to use rigid link mechanisms is difficult to get around due to the perception that strength and rigidity go hand in hand. This is not always true, especially in the case of compliant mechanisms. Compliant designs can offer many advantages over rigid type mechanisms and for these reasons we feel a compliant knee could easily be an improvement over current technology.

Unfortunately compliant mechanisms are not void of weaknesses. Compliant hinges are not well suited and could potentially be subject to failure under these conditions.

- Excessive flexural deflection
- Cyclic fatigue
- Compressive flexural loading

Specific design consideration must be made to prevent such failure. Limiting of the maximum deflection can be accomplished with proper design. By limiting the flexural deflection, fatigue associated with cyclic loading is greatly reduced. Proper location of the compliant hinges during design is required to prevent compressive failure. [10]

Conclusions

It has been readily recognized that the correct alignment of the prostheses is imperative not only to successful mobility, but also to the longevity of prosthetic use. Longevity and durability are determined by a number of factors, including the underlying causative factor for prosthetic use and the comfort and control that are present as a result of the prosthetic.

Over the course of the last two decades, considerable changes in the development of prosthetics for transfemoral amputees have developed. It has been recognized that the development of modern prostheses reflects the focus on this type of amputation and the need for young children to have access to functional prosthetics from an early age.

Good prosthetic design requires toe clearance, stability, light in weight, and adjustability. To classify knees FEA and calculation of the instant center of rotation can help in determining toe clearance and stability. Use of compliant mechanisms should reduce the weight of the prosthetic and allow young children the benefit of an articulating prosthetic knee.

Evidence suggests that there is a considerable level of improvement that can be made in the existing prostheses for pediatric patients, including improvements that will continue to support durability and longevity as goals when introducing the prosthetics.

Chapter Three

Adjustable Compliant Pediatric Knee

Adjustability is importance in prosthetic knee design as well as prosthetics in general. No two people are the same shape or size, and have varying needs, abilities and disabilities. Children are no different than adults in this regard and require variability in their prosthetics in order to be properly fitted for their needs. From these criteria, we have utilized a design that offers the simplicity of compliance with the versatility of an adjustable return spring. We have created a way of housing a movable spring that is capable of changing the amount of torque required to bend the knee. By moving the position of one end of the adjustable spring along the body of the prosthetic, variable feedback is possible.

From research with Dr. Highsmith and Dr. Maitland, from the Department of Physical Therapy at the University of South Florida, an important concept has made its way into discussion. The ability to adjust the return spring rate of the lower leg in pediatric prosthetic knees would be an improvement over existing technology. Due to the inherently simplistic function of compliant mechanisms a challenge was presented. Dr. Highsmith noted that a well thought out design would offer the ability to change the function and force feedback of the knee from one task to another. [1] Children, being very energetic and active are often times in situations that require variable function of their prosthetic. Just like our own knees, which are capable of adapting to our environment instantaneously, children with prosthetic knees need a way of adapting to their environment. For example, a child would require different types of feedback from their prosthetic depending on whether they were crawling, walking or running. A knee that was suited for crawling would be durable due to constant contact with the ground and flexible with little resistance to bending to allow the child to flex the knee with ease. On the other hand, a good knee for walking would have to be able to lock during the loading and stance phases of gait. As for running, the prosthetic would have to have a slightly higher resistance to bending to allow the lower leg to keep up with the users stride. For this reason we have developed a way of making the knee adjustable while utilizing the simplicity of compliant design.

A static rigid leg is typically prescribed to young children in the early stages of walking to assist them even though its only use is for walking. These knees are not able to flex during crawling or sitting and must be removed many times during the day. These knees are often times not used until the child is ready to walk. It was noted by Dr. Highsmith that the longer the child was able to wear his or her prosthetic prior to walking the quicker the child would be able to stand. This act of standing is critical for a child's mental and physical development. A child's ability to manipulate and interact with his or her surroundings increases dramatically from crawling to standing which promotes mental development. A child's environment becomes more interactive once he or she begins to pull up to walk. For this reason it is crucial that prosthetics be made variable and able to be worn for longer periods of time, allowing the child time to adapt prior walking. [1] The physical benefits of early use of prosthetics are achieved through loading of the bones in stance phase. The simple act of standing is a requirement in growing strong bones and joints. The stimulus generated from standing promotes growth within the muscles, bones and the joints. Without these stimuli our legs would atrophy and deteriorate over time.

As a solution to the problems associated with rigid links, an adjustable compliant four-bar knee was investigated and improved upon. The basic design utilized was initially conceived by Guérinot et al. (2004) [11]. This design was simple and effective but lacked the ability to vary the spring return of the lower limb. For this reason we have developed a new design that allows the user or guardian of the user to change the settings of the knee and alter the response of the output. To do this we took the initial design and made modifications to the body of the knee to allow for a movable segment within the compliant spring system. This movable spring is what allows the knee to adjust. This is achieved by sliding one end of the spring along the body of the knee to various positions. These various positions allow the spring to be forced into different modes of bending. These different modes of bending have different stiffness associated with them. By increasing the mode of bending, the force feedback is increased by a linear factor allowing the knee to be adjusted with one simple motion. This added feature creates ability for the knee to lose flexibility when needed and regain it when needed depending on the task at hand.

Chapter Four

Position/Displacement Analysis

This chapter begins by explaining the significance of the anatomic knee and prosthetic knee function during human gait. The chapter later talks about how we gathered position data and finishes up with how and why we calculate instant centers.

The knee is an important component in walking as well as stance control. The anatomic and polycentric prosthetic knee rotates and translates during swing phase to increase toe clearance and stability during stance. Polycentric knees offer more toe clearance per angle of flexion than single axis prosthetics. The prosthetic knee's shank rotates and translates to an angle of 60º with respect to the socket during the end of the initial swing phase and beginning of mid-swing phase to achieve the required amount of toe clearance to prevent stubbing the toe. [12] In order to determine the rotational characteristics of the compliant knee, the amount of rotation and translation the particular knee designs under go, the shank off each knee was forced to rotate relative to the socket by an amount of, Θ, which ranged between $0° < \Theta < 120°$, in increments of 6°. *Figure*

4.1 illustrates the direction of the motion that takes place during the forced rotation of the compliant knee. The rotation of the shank connection is accompanied by a particular amount of translation, in other words the IC is not fixed but is in constant motion during the flexion and extension of the prosthetic knee. In *Figure 4.1* points *A* and *B* indicate the location of the path taken by points on the upper and lower section of the semi-circle that makes up the lower half of the compliant knee that attaches to the shank. The IC's location is determined by these paths.

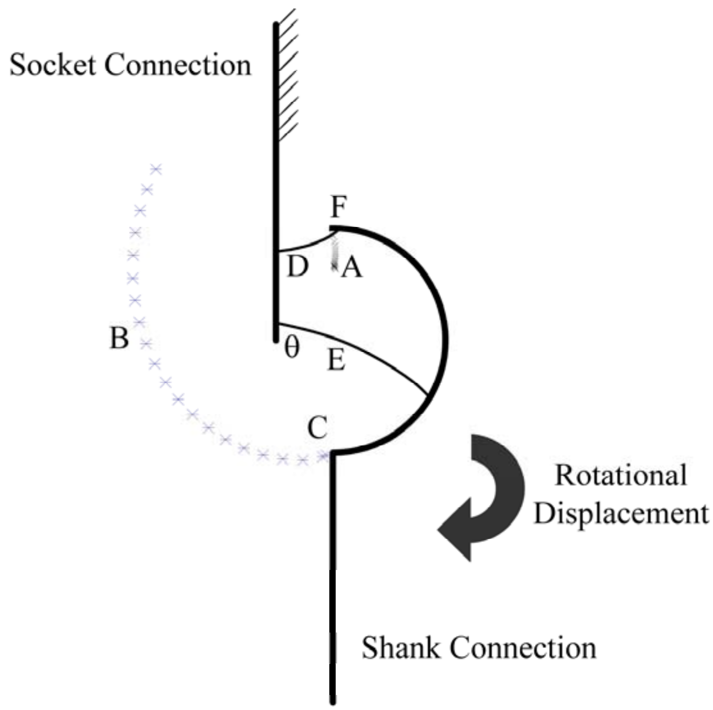


Figure 4.1. Rotational plots using finite element analysis, showing the motion of points F and C when the shank is forced to rotate relative to the socket having a bracket angle θ for θ = 64º

In order to calculate the location of the instant center of the compliant knee, a graphical solution method was utilized. This method is generally used on rigid-body mechanisms because it is relatively simple to graphically track the motion of the links. The links are constrained to specific paths and it is generally easy to solve for their motion. By knowing the direction of motion of any two points on a moving object it is possible to find the instant center or rotation of the object. For compliant mechanisms, knowing where the mechanism will tend to travel at any instant is not easily predicted. Compliant mechanisms direction of motion is controlled by multiple factors. For single degree of freedom rigid link mechanisms there are only two possible solutions for any position attained and can be calculated using vector math. Compliant mechanisms on the other hand are not as easy to solve. Unlike rigid link mechanisms which have physical constraints binding their motion, compliant mechanisms deform to cope with external forces. Generally a pseudo-rigid-body model is utilized to gather information from a compliant mechanism but with the use of finite element analysis we were able to gather the kinematics data as well as force analysis simultaneously without the need for a pseudo-rigid-body model. First FEA, using two dimensional beam3 elements, was used to accurately plot the position of the knee by placing a pure rotation, i.e. no forces, only torque, on the lower half of the compliant knee at location *C* in *Figure 4.1*. The rotation was placed on the knee in steps of six degrees, relative to the socket connection, until a full rotation was reached at 120 degrees. With this data we are able to better predict motion characteristics of the knee and offer educated solutions when generating design changes.

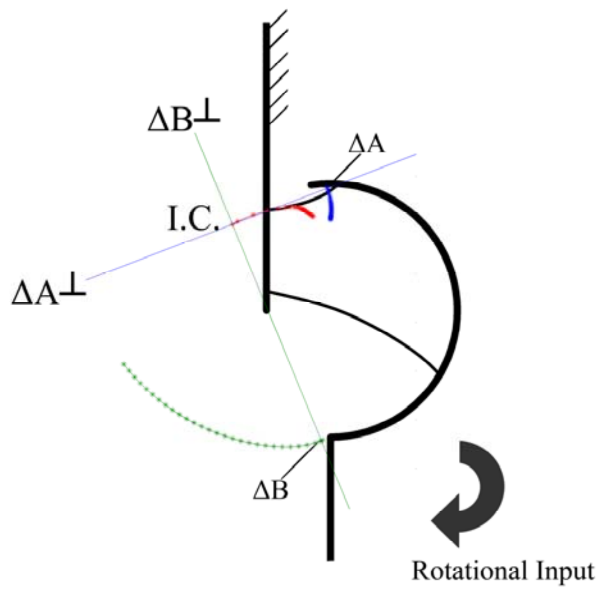


Figure 4.2. Instantaneous center of rotation path of the compliant knee

The data gathered from the rotational displacement was then utilized to find the path of the instant center of the shank relative to the socket, shown in *Figure 4.2*. The perpendicular bisectors *ΔA*⊥ and *ΔB*⊥ of the initial displacement labeled *ΔA* and *ΔB* in *Figure 4.2* cross one another at the instant center. The perpendicular bisectors are displayed to give a visualization of how we obtain the IC as well as to show where the IC path starts and ends. Four-bar knees on the market also utilize this type of analysis to determine the exact location of the IC's.

The data in that case is used to determine what changes are needed in the rigid links to obtain an optimum design. An example of IC analysis done on four bar knees is shown in *Figure 4.3*. This figure shows a lateral view of a typical four bar linkage knee with the instant center plotted at 5-degrees increments of flexion. Points *A, B, C* and *D* represent the axes of rotation of the four-bar knee. The IC is determined by intersecting the line through points *A* and *B* with the line through points *C* and *D.* [13]

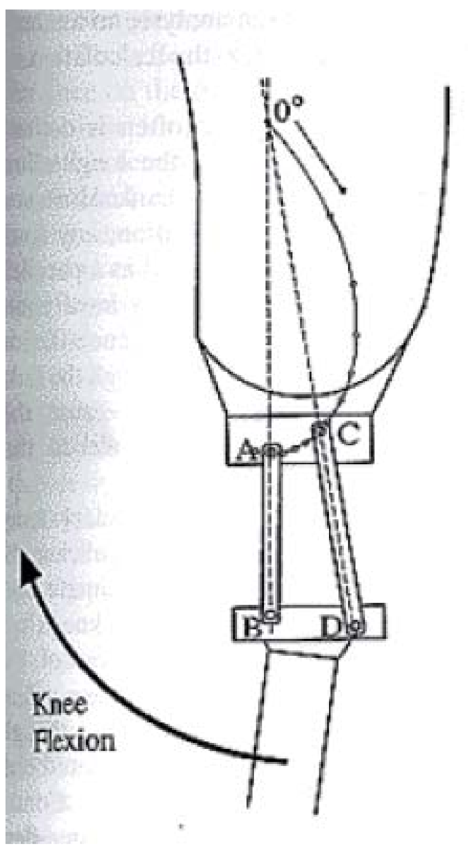


Figure 4.3. Instant center path found in a four-bar knee design [5]

Of the 11 knees currently on the market for children, all but three of the knees are four-bar knees. Four-bar knees provide greater toe clearance than single-axis knees for a given knee-flexion angle and offer stance-phase stability. This added toe clearance allows the user to walk with less concern for stubbing their toes and tripping during the gait cycle. During the gait cycle, as much as 3.2 additional centimeters of toe clearance is achieved over single axis knees in adult testing. [13]

In addition to the added toe clearance achieved, the four-bar knee also offers stance-phase stability. The stability of a four-bar knee during load-bearing is determined by the location of the IC with respect to the ground reaction force vector. Prosthetists are given some control over the degree of stability through prosthetic alignment. For stability during stance, the IC must be posterior to the ground reaction force to maintain the extension moment which keeps the knee in a locked upright position, illustrated in *Figure 1.1* with the exception of the toe off phase. [13] From the position analysis done on the compliant knee we have found that the IC falls well behind what would be the ground reaction force in simulation. This data proves the compliant knee is stable based on the instant center location.

Conclusion

By knowing the IC's path for the full rotation of the compliant knee, it is possible to compare this information with the data found on IC paths of current four-bar knees. If a light weight compliant knee can match or improve on the kinematic behavior of a rigid link knee, it will be a benefit to pediatric amputees.

Future areas for optimization of compliant knees include examining the effect on the instant center caused by varying specific parameters. These parameters include the compliant spring length, bracket angles and compliant spring material.

Simulations could possibly assist in answering these questions. In addition to not knowing how the knee will react to dynamic loads, we feel we need to research where the IC would best be located to offer the most assistance to the user. Although there is significant data available on the proper IC path for adult four-bar knees, less is known about what a child would benefit from. We wish to know if it is possible to relate the motion of the compliant knee back to rigid-body four-bar motion.

Future work also includes creating a simulation of the compliant knee function throughout the gait cycle, in other words there is a particular combination of forces and moments that will be seen by the knee during the amputee's gait. Thus far we have only simulated linear and rotational displacements, not a realistic combination of force and moments.

The first step in obtaining realistic force and moment combinations is to use ground reaction data from a rigid link knee. It is not yet certain that such data would faithfully represent the forces seen by the compliant knee but this has to be the starting point.

Previous research on rigid link knees has indicated that it is desirable for the instant center to be located near a sound joint or the end of the residual limb. We hypothesize that this will still be true and it appears to be the case in the designs researched thus far.

Another research question is whether the same design philosophy as used for adult knees truly applies in the pediatric knee. Certainly the laws of physics apply equally well to adults and children. However, the pediatric knee serves as a training function rather than a re-training function. The child learning to walk does not have prior experience. Thus, the knee may need to be designed to actively discourage poor walking habits by making them either more comfortable or less tiring to use.

It may also be desirable to index the motion of the instant centers in a precise way. One possible approach to this problem would be to determine the rigid four-bar link with instant center motion that most resembles the compliant four-bar motion.

Chapter Five

Force and Stress Analysis

In this chapter, the force and stress analysis of the adjustable compliant knee is discussed. First, investigations into the stiffness of the knee with respect to applied loads are described. Then, the stress is analyzed to validate the knee's ability to function under prescribed loads. Lastly, calculations were made to acquire a safety factor for the compliant knee based on peak forces associated with normal gait.

To better understand how the knee would react to the weight of a child, we studied the force characteristics of the knee throughout a full range of motion for multiple bracket configurations. The adjustable brackets, illustrated in *Figure 5.1,* offer a way to adjust the compliant segments by moving the location of brackets. The bracket angle θ, illustrated in *Figure* 5.1 ranges from $0° < \theta < 140°$. To further analyze the knee's force feedback we took the data from the displacement analysis to give us a baseline starting position for our new displacement paths. This baseline starting position is the path of lowest resistance to rotation for the mechanism. In order to build on the work done in the previous chapter, the data about the path of lowest resistance is used to determine the resistance of the mechanism to motion away from that path. This is done by applying displacements first along the path of least resistance and then in a zigzag pattern about that path as shown in *Figure 5.2.* For each individual displacement along the path made, the knee was forced to move away from the path of least resistance in a way that would cover as much area as possible. To prevent missing data locations, the force magnitude data was overlapped as illustrated in *Figure 5*.2 using a $\log_{10}$ scale. The data from the force plots was later smoothed using interpolation between the adjacent data points, as is shown in *part a)* of *Figures 5.3-5.6*. The length of the vector arrows in *part a)* of each of *Figures 5.3-5.6* indicate the magnitude of the force being applied to the knee at each data location. Because the magnitudes of the forces are very large in the lower left hand side of the plot, the forces at other locations are proportionally too small to see. In *part b),* of *Figures 5.3-5.6,* the arrows are normalized to show the force directions and do not represent the actual magnitude. These steps were then repeated for multiple bracket angle configurations to gather data on the change in force feedback due to the angle deviation in the compliant spring.

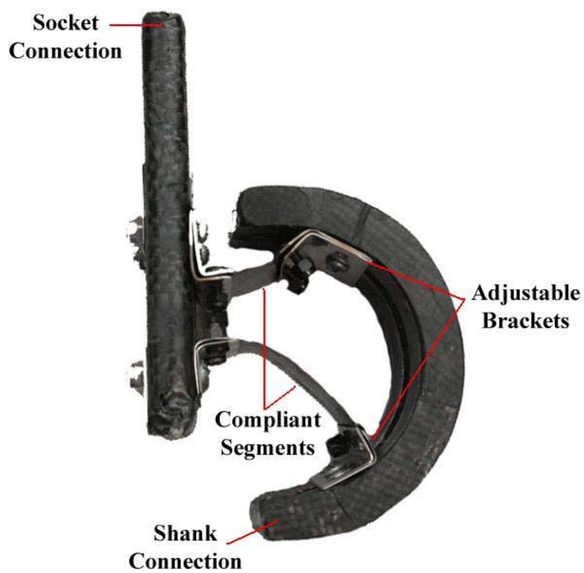


Figure 5.1. Prototype design labeled

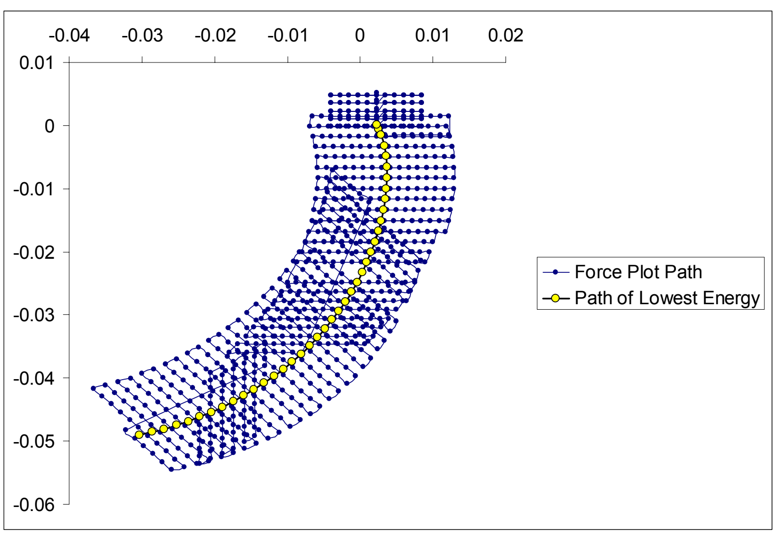


Figure 5.2. Paths used to generate force plots measured in meters

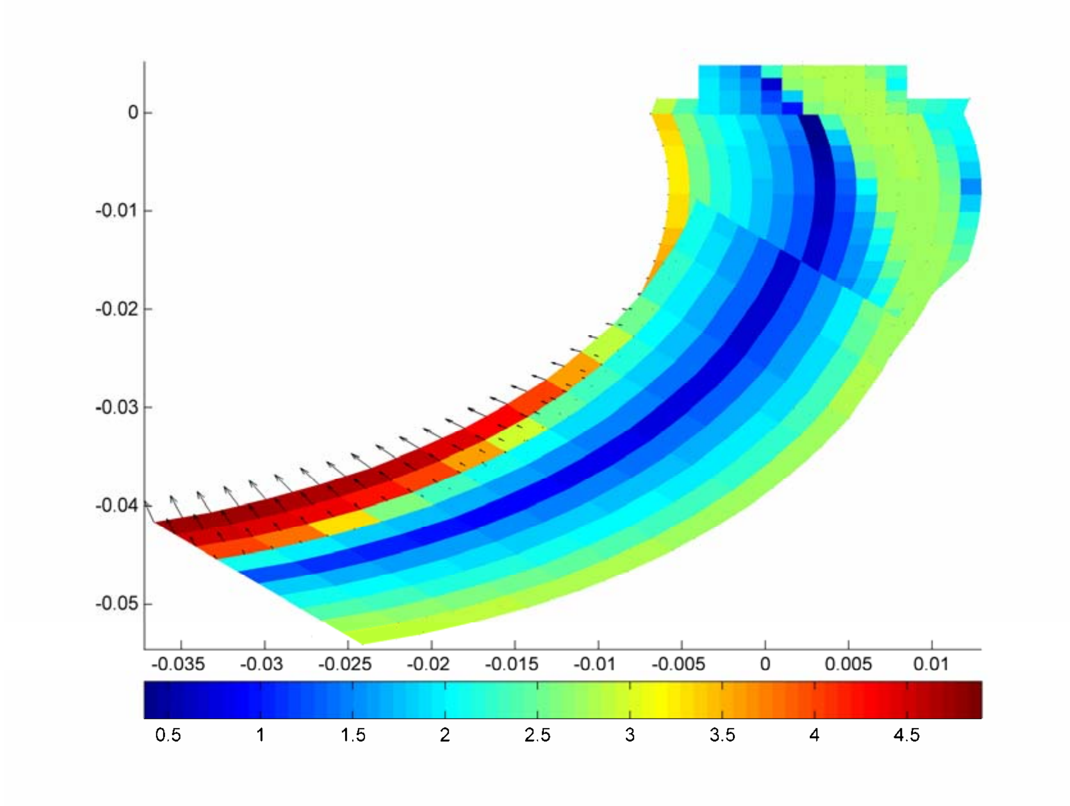
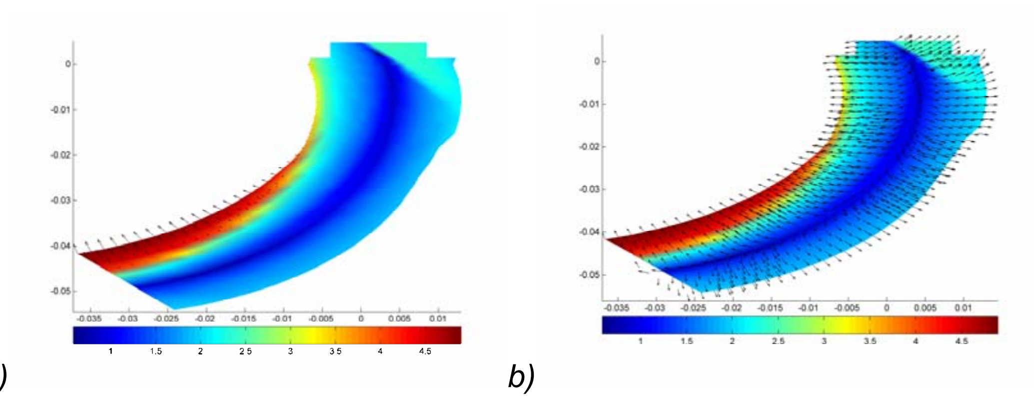


Figure5.3. Force plot data of the compliant knee in kg on a log_{10} scale utilizing overlapping of multiple data sets with a bracket angle of 50 degrees



a) b)

Figure 5.4. Force plot data of the compliant knee in kg on a log_{10} scale with a bracket angle of 80 degrees measured in meters

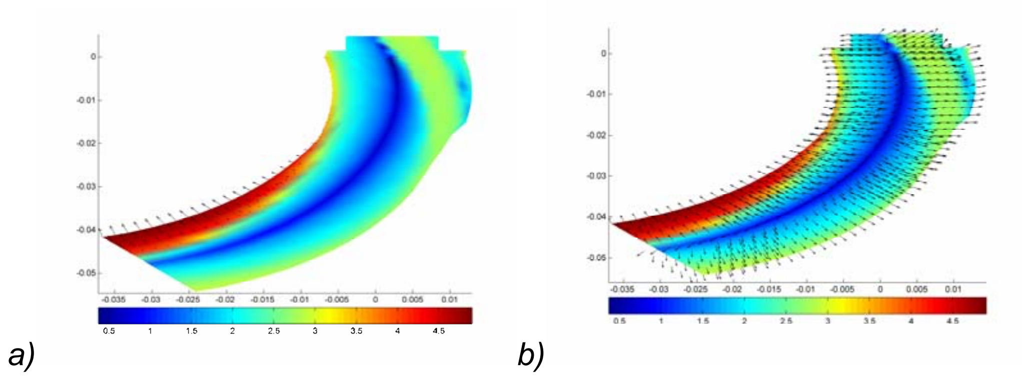


Figure 5.5. Force plot data of the compliant knee in kg on a log_{10} scale with a bracket angle of 70 degrees measured in meters

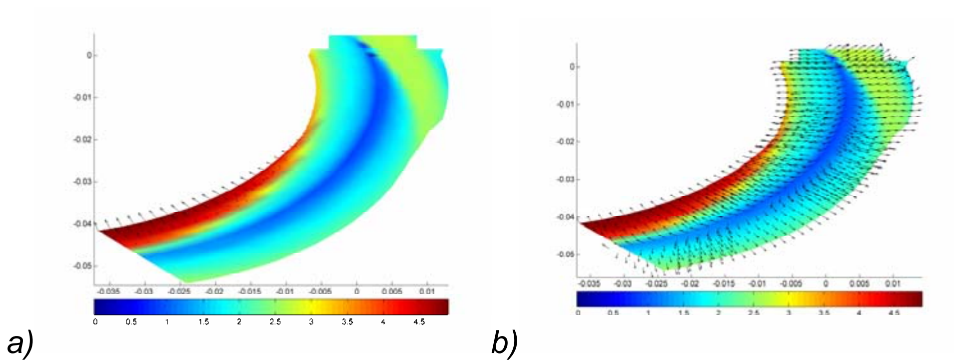


Figure 5.6. Force plot data of the compliant knee in kg on a log_{10} scale with a bracket angle of 60 degrees measured in meters

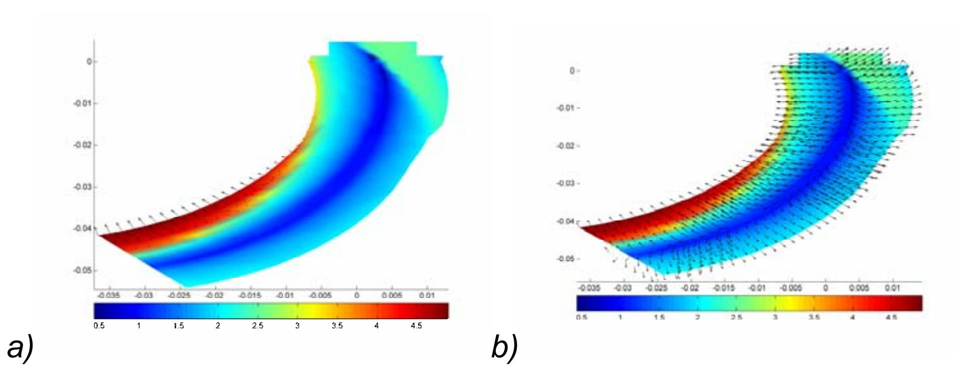


Figure 5.7. Force plot data of the compliant knee in kg on a log_{10} scale with a bracket angle of 50 degrees measured in meters

From the force plots we generate useful information about how the compliant knee will respond to external forces. The plots illustrated in *Figures 5.3-5.6* clearly show the lowest forces fall along the path of least resistance and is illustrated in the figures. The higher forces in *Figures 5.3-5.6* are plotted using red while lower forces appear as blue. From these plots we know the forces increase rapidly away from the path of lowest energy indicating a relatively high resistance to deflection away from the intended path. In other words, the knee requires large forces to displace the shank connection, shown in *Figure 4.1,* from its path of least resistance. This is important because excessive displacements away from the projected path could result in instability and could potentially cause the amputee to fall. From inspection, forces increase more rapidly when the knee is in compression. This is largely due to the tension that results in the shorter compliant segment, labeled *D* in *Figure 4.1.* This result is the case because the shorter compliant segment is mostly in tensile axial happens to be stiffer than the bending mode seen along the path of lowest resistance.

Simultaneous to the FEA force data being calculated, the stress analysis data was also calculated. The plots shown in *Figure 5.8-5.11* illustrate the ultimate stress endured by the compliant knee as it was forced to plot along the path shown in *Figure 5.2.*

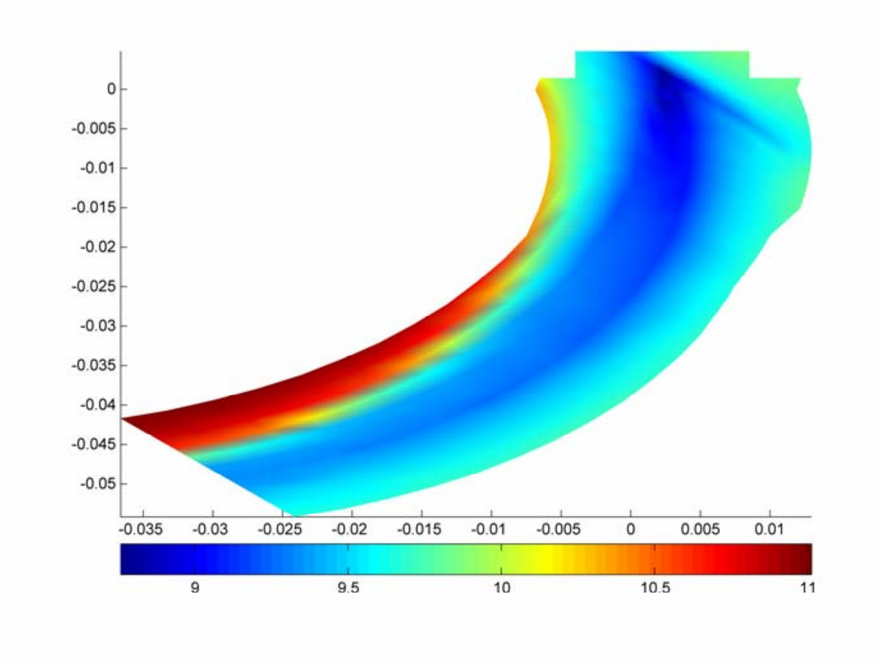


Figure 5.8. Stress plot data of the compliant knee in GPa on a $\log_{10}$ scale with a bracket angle of 80 degrees measured in meters

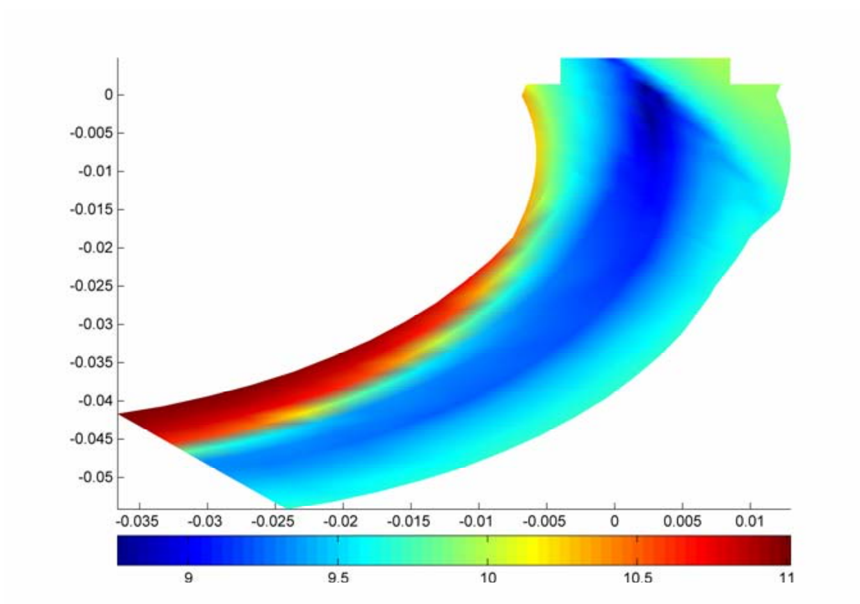


Figure 5.9. Stress plot data of the compliant knee in GPa on a $\log_{10}$ scale with a bracket angle of 70 degrees measured in meters

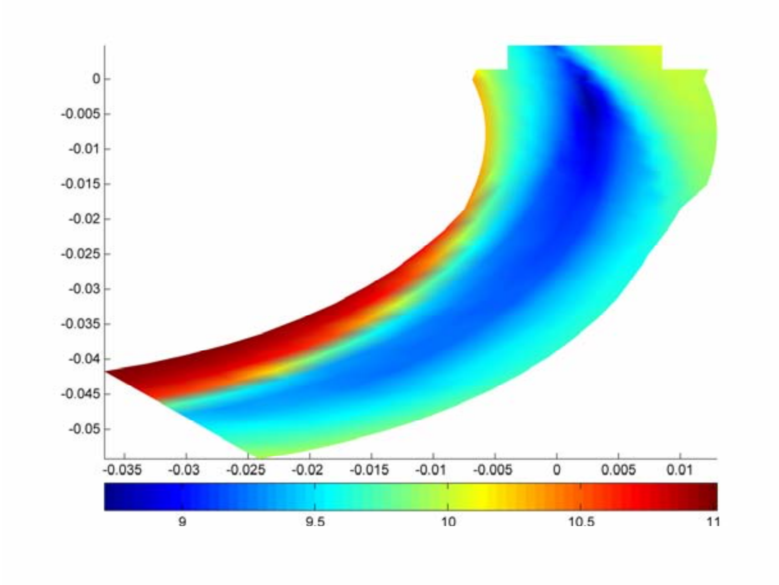


Figure 5.10. Stress plot data of the compliant knee in GPa on a $\log_{10}$ scale with a bracket angle of 60 degrees measured in meters

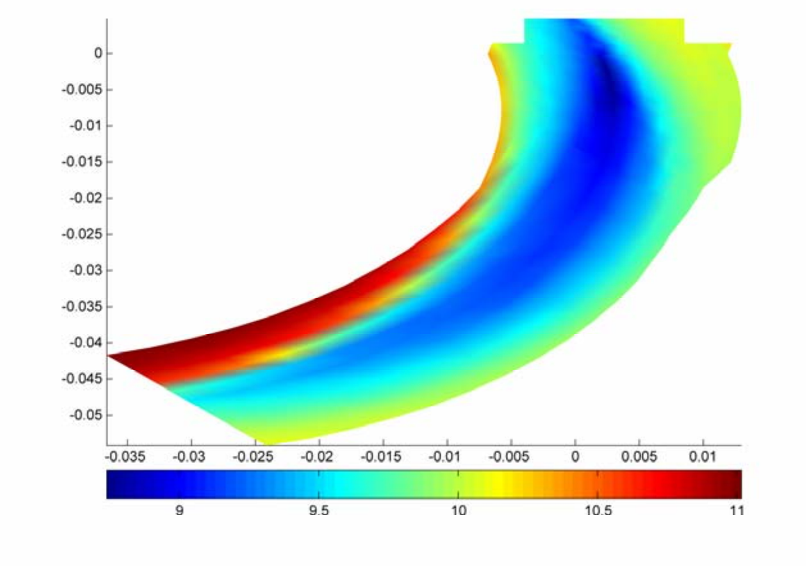


Figure 5.11. Stress plot data of the compliant knee in GPa on a $\log_{10}$ scale with a bracket angle of 50 degrees measured in meters

Conclusion

The stress plots look surprisingly similar to the force plots with the exception of the magnitude of the numbers present in the key. The values of stress are significantly higher than the force.

The force and stress plots indicate that the knee would reach equilibrium at .02 inches if loaded with a point load of 200 lbs in stance. Unfortunately the stress in the compliant members for a 200 pound load reached approximately 3.5 GPa, nearly 2 times the acceptable tensile strength of 1.758 GPa for SAE 1070-1090 high carbon tempered spring steel, indicating a failure would result. [14] The color plot color that corresponds to 1.758 GPa, which translates to 9.25 on the $\log_{10}$ scale, is displayed as light blue shown in *Figure 5.11.* The exceeded stress levels indicate that modification will need to be made to the design in order to overcome the stresses acquired during the test load. For these tests it is assumed that the ends of compliant members have fixed position and angle using beam3 type elements.

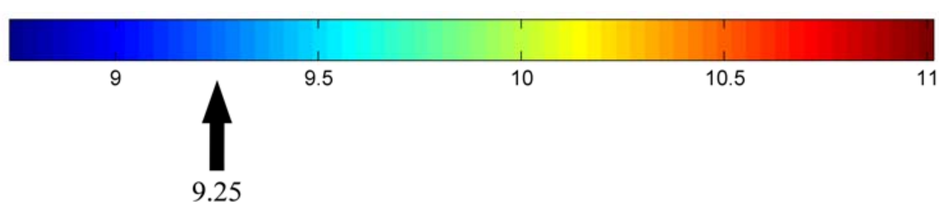


Figure 5.12. Corresponding plot color for 1.758 GPa on a $\log_{10}$ scale

Future work involves plotting precise regions of acceptable stress levels for the knee. This would allow us to better determine how much deflection would be required to reach the stress limitations for various positions and could be used to calculate the exact force required to generate these stress levels.

Future areas for optimization of compliant knees include examining the change in forces and stresses within the knee when changing the following parameters.

- Spring lengths
- Angles of the spring
- Stiffness of the springs
- Contact elements

Chapter Six

Discussion and Conclusions

We have designed and examined a pediatric compliant knee mechanism that may offer solutions to problems that exist for young children who are just learning to walk. One of our achievements was in mechanism design by way of creating a pediatric compliant knee mechanism with the ability to adjust external rotational resistance. In addition to the functional achievements, we have created a prototype weighing less than five ounces. We formulated a method for analyzing the rotation and translational motion using nonlinear FEA of the compliant knee accomplished through calculation and plotting of the instant centers of the lower shank with respect to the socket connection. In addition, we have developed a method for simultaneously calculating the external reaction forces and internal stresses present for displacements made within the anticipated region of motion. From our analysis position, force and stress data was gathered for a deflection of θ for $0° < \theta < 120°$. The instant centers and prototype both indicate that the knee offers adequate stability for stance loading. Unfortunately the force and stress plots indicate that the knee will not support a point load of 200 lbs in. The stress in the compliant members for a 200 pound load reached approximately 3.5 GPa, twice the acceptable tensile strength of 1.758 GPa for SAE 1070-1090 high carbon tempered spring steel. [14] The exceeded stress levels indicate that modification will need to be made to the design in order to overcome the stresses acquired during the test load.

Future work includes:

- Design a gait simulator for FEA
- Research the effect of IC location
- Find the IC location that would benefit children
- index the motion of the instant centers in a precise way
- Find and relate a four-bar knee that most resembles the compliant four-bar motion
- Introduce contact elements that will redirect forces through the rigid frame of the knee and away from the compliant members

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Appendices

Appendix A

Ansys Code Used to Find Force and Stress Data

Modify the value “theta4degrees” to attain the force and stress data

```
! knee.bat

!

! This ansys batch file analyzes compliant four-bar knee designs

!

/COM,

/COM,Preferences for GUI filtering have been set to display:

/COM,  Structural

/COM,  Thermal

!*

!*

!******************BEGIN PREPROCESSOR STEPS ***********

/PREP7


!*******************************************************

!************SET UP MODEL PARAMETERS********************

!*******************************************************

!********** UNITS IN NEWTONS, METERS, ETC.
```

```
!***

!***


!--------------number of divisions-------------------

Segments3 =120

Segments5 = Segments3/3


!-------------------Variables-----------------------


 theta4degrees = 45


!---------------Calculate parameters-----------------

in_m   = .0254

PI     = acos(-1.)

R      = in_m*1.02       ! Knee arc radius ---> 1.02 inches

theta2 = 176.5*PI/180     ! first compliant beam origin on knee arc in radians

L3     = in_m*.5          ! first compliant segment length ---> .5 inch

theta3 = 300*PI/180

b3ang  = 275*PI/180       ! bracket angle for first compliant segment

theta4 = theta4degrees*PI/180  ! second compliant segment origin on knee arc in radians

L5     = in_m*(1.5)        ! second compliant segement length ----> 1.5 inch

theta5 = theta4+165*PI/180

b5ang  = 265*PI/180       ! bracket angle for second compliant segment

Lleg   = 4*in_m          ! guess lower leg mass center at 4 inches from center of knee
```

```
delta3 = .625*in_m      ! assembly distance for point 5 from point 3

x3     = R*cos(theta2)+L3*cos(theta3)

y3     = R*sin(theta2)+L3*sin(theta3)

x5     = R*cos(theta4)+L5*cos(theta5)

y5     = R*sin(theta4)+L5*sin(theta5)

dx5    = x3 +delta3 - x5  ! displacement in x for point 5

dy5    = y3 - y5          ! displacement in y for point 5

t2 = theta2*180/PI

t3 = theta3*180/PI

t4 = theta4*180/PI

t5 = theta5*180/PI

b3 = b3ang*180/PI

b5 = b5ang*180/PI

d3 = delta3/in_m


!---------------file name subscript----------------------------




!!!!!!!!!!!!!!!!!!!!!!!!!!!!!!!!

! define segment properties !!!!

! compliant segments        !!!!

!!!!!!!!!!!!!!!!!!!!!!!!!!!!!!!!
```

```
t1 = 1/200*in_m  ! thickness is 1/200 inch
w1 = 1.5*in_m    ! width is 1.5 inch
A1 = t1*w1          ! Area
I1 = w1*t1*t1*t1/12   ! Moment of Inertia
Esteel = 200E9       ! Young's Modulus in GPa
psteel = 0.28        ! poisson's ratio
!*
! rigid segments
t2 = 0.5*in_m    ! thickness is 1/200 inch
w2 = 1.5*in_m    ! width is 1.5 inch
A2 = t2*w2          ! Area
I2 = w2*t2*t2*t2/12   ! Moment of Inertia
Ealuminum = 70E9      ! Young's Modulus in GPa
paluminum = 0.38     ! poisson's ratio


ET,1,BEAM3
KEYOPT,1,6,1
KEYOPT,1,9,0
KEYOPT,1,10,0
R,1,A1,I1,t1, , , ,
R,2,A2,I2,t2,0,0,0,
MPTEMP,,,,,,,,
MPTEMP,1,0
MPDE,EX,1
```

```
MPDE,PRXY,1

MPDATA,EX,1,,Esteel

MPDATA,PRXY,1,,psteel

MPTEMP,,,,,,,,

MPTEMP,1,0

MPDATA,EX,2,,Ealuminum

MPDATA,PRXY,2,,paluminum


!-----------CREATE KEYPOINTS: K,Point #, X-coord, Y-COORD, Z-Coord

K,1,0,0,0

K,2,R*cos(theta2),R*sin(theta2),0

K,3,R*cos(theta2)+L3*cos(theta3),R*sin(theta2)+L3*sin(theta3),0

K,4,R*cos(theta4),R*sin(theta4),0

K,5,R*cos(theta4)+L5*cos(theta5),R*sin(theta4)+L5*sin(theta5),0

K,6,R,0,0

K,7,-R,0,0

K,8,R+Lleg,0,0


!----------Create knee using arcs and lines--------

! arcs in the knee:  7 to 2,  2 to 4, and 4 to 6

LARC,7,2,1,R

LESIZE,1,,,5

LARC,2,4,1,R

LESIZE,2,,,5
```

```
LARC,4,6,1,R

LESIZE,3,,,5

! lines in model: 2 to 3, 4 to 5, and 6 to 8

L,2,3

LESIZE,4,,,Segments3

L,4,5

LESIZE,5,,,Segments5

L,6,8

LESIZE,6,,,10


!------------------- MESH ------------

type,1

real,1

mat,1

LMESH,4

LMESH,5

real,2

mat,2

LMESH,1

LMESH,2

LMESH,3

LMESH,6


FINISH
```

```
!-----------Get Nodes at Chosen Keypoints-----------

ksel,s,kp,,3

nslk,s

*get,nkp3,node,0,num,max

nsel,all

ksel,all


ksel,s,kp,,5

nslk,s

*get,nkp5,node,0,num,max

nsel,all

ksel,all


ksel,s,kp,,6

nslk,s

*get,nkp6,node,0,num,max

nsel,all

ksel,all


ksel,s,kp,,7

nslk,s

*get,nkp7,node,0,num,max

nsel,all
```

```
ksel,all


!********************************************

!***********SOLUTION STEPS *****************

!********************************************


/SOLU


! Set to Nonlinear Deflection Analysis

NLGEOM,on


CNVTOL,ROT,,0.01,,0

CNVTOL,U,,0.01,,0

CNVTOL,F,,0.001,,0

! Set Analysis Type to Static (0)

ANTYPE, 0


!*****************************************************************

!******************* Set up Boundary Conditions *****************

!*****************************************************************


!---------------------------------------------|

!  Keypoint3  is fixed with a twist on the end  |

!---------------------------------------------|
```

```
DK,3,UX,0,,

DK,3,UY,0,,

DK,3,ROTZ,b3ang-theta3,,


DK,5,UX,dx5,,

DK,5,UY,dy5,,

DK,5,ROTZ,b5ang-theta5,,

steps = 0

steps = steps+1

lswrite,steps

        *enddo

  lssolve,steps

move1=steps


!*************************************************************************

! _________________________ Displacement Input _________________________ *

!*************************************************************************

        *do,nn,1,5,1

DK,6,UX,.005*nn

Dk,6,UY,0

steps=steps+1

lswrite,steps

disx11 = .005*nn
```

```
*enddo
lssolve,move1+1,steps
move2=steps

        *DO,ll,1,10,1
DK,6,UX,disx11
Dk,6,UY,-.005*ll
steps=steps+1
lswrite,steps
dis22y = -.005*ll
        *ENDDO
lssolve,move2+1,steps
move3=steps

        *DO,kk,0,10,2
        *DO,rr,1,10,1
        DK,6,UX,disx11-rr*.005
        Dk,6,UY,dis22y+kk*.005
        steps=steps+1
        lswrite,steps
        disx33 = disx11-rr*.005
        *ENDDO
        lssolve,move3+1,steps
move4 = steps
```

```
*DO,ss,0,9,1

            DK,6,UX,disx33+ss*.005

            Dk,6,UY,dis22y+kk+1*005

            steps=steps+1

            lswrite,steps


            *ENDDO

            lssolve,move4+1,steps

move5 = steps




            steps = steps+1

            lswrite,steps

            *ENDDO

            lssolve,move5+1,steps







FINISH
```

```
!*****************************************************************
!  ++++++++++++++++++++  Results  ++++++++++++++++++++++++    *
!*****************************************************************

/POST1
*DIM,Smax,TABLE,Steps
*DIM,Ydis7,TABLE,Steps
*DIM,Xdis7,TABLE,Steps
*DIM,Ydis6,TABLE,Steps
*DIM,Xdis6,TABLE,Steps
*DIM,Qdis6,TABLE,Steps                  !INPUT ROTATION ON NODE 6
*DIM,Xforce6,TABLE,Steps
*DIM,Yforce6,TABLE,Steps
*DIM,Xforce3,TABLE,Steps
*DIM,Yforce3,TABLE,Steps
*DIM,Xforce5,TABLE,Steps
*DIM,ZMoment5,TABLE,Steps
*DIM,ZMoment3,TABLE,Steps
*DIM,Yforce5,TABLE,Steps

        *DO,mm,1,steps,1
```

```
SET,mm

ETABLE,smxi,NMIS,1

ESORT,ETAB,smxi,0,1

*GET,stress,SORT,0,MAX

*SET,Smax(mm),stress

*GET,disX7,NODE,nkp7,U,X

*SET,Xdis7(mm),disX7

*GET,disY7,NODE,nkp7,U,Y

*SET,Ydis7(mm),disY7

*GET,disY6,NODE,nkp6,U,Y

*SET,Ydis6(mm),disY6

*GET,disX6,NODE,nkp6,U,X

*SET,Xdis6(mm),disX6

*GET,disQ6,NODE,nkp6,ROT,Z

*SET,Qdis6(mm),disQ6

*GET,forceX3,Node,nkp3,RF,FX

*SET,Xforce3(mm),forceX3

*GET,forceY3,Node,nkp3,RF,FY

*SET,Yforce3(mm),forceY3

!*GET,forceX6,Node,nkp6,RF,FX

!*SET,Xforce6(mm),forceX6

!*GET,forceY6,Node,nkp6,RF,FY !not needed, Fxy = 0 when rotating

!*SET,Yforce6(mm),forceY6

*GET,MomentZ3,Node,nkp3,RF,MZ
```

```
*SET,ZMoment3(mm),MomentZ3
*GET,forceX5,Node,nkp5,RF,FX
*SET,Xforce5(mm),forceX5
*GET,forceY5,Node,nkp5,RF,FY
*SET,Yforce5(mm),forceY5
*GET,MomentZ5,Node,nkp5,RF,MZ
*SET,ZMoment5(mm),MomentZ5


    *ENDDO
    /output,knee_rotation_%theta4degrees%deg,txt,,Append
    *MSG,INFO1,'Theta2','Theta3','BrAng3','Theta4','Theta5','BrAng5','Delta3'
     %-9C %-9C %-9C %-9C %-9C %-9C %-9C
    *VWRITE,t4,t2,t3,b3,t5,b5,d3
     %9.2G %9.2G %9.2G %9.2G %9.2G %9.2G %9.2G
    *MSG,INFO2,'Stress_Max','Ydis7','Xdis7','Xdis6','Ydis6','Rot6'
     %-11C %-11C %-11C %-11C %-11C %-11C
    *VWRITE,Smax(1),Ydis7(1),Xdis7(1),Xdis6(1),Ydis6(1),Qdis6(1)
     %11.4G %11.4G %11.4G %11.4G %11.4G %11.4G
    *MSG,INFO2,'Xforce3','Yforce3','Xforce5','Yforce5','ZMoment3','ZMoment5'
     %-11C %-11C %-11C %-11C %-11C %-11C
    *VWRITE,Xforce3(1),Yforce3(1),Xforce5(1),Yforce5(1), ZMoment3(1), ZMoment5(1)
     %11.4G %11.4G %11.4G %11.4G %11.4G %11.4G
    !*MSG,INFO2,'Xforce6','Yforce6'
```

```
        ! %-11C %-11C
        !*VWRITE,Xforce6(1),Yforce6(1)!not needed, fxy = 0 when rotating
        ! %11.4G %11.4G
/output

FINISH

!****************************************************
!  ************  PLOT FINAL POSITION  ************  *
!****************************************************

/POST1
/EFACE,1
SET,LoadSteps
AVPRIN,0,0,
PLNSOL,U,X,1,1
FINISH

/POST1

PLDISP,1
```

Ansys Output File

Theta2	Theta3	BrAng3	Theta4	Theta5	BrAng5	Delta3
64.	1.27E-02	3.00E+02	2.75E+02	2.29E+02	2.65E+02	0.63

Appendix A (Continued)

Stress_Max	Ydis7	Xdis7	Xdis6	Ydis6	Rot6
4.5919E+08	1.0575E-03	2.5879E-03	2.5641E-03	-5.1223E-04	-3.0298E-02
4.5919E+08	1.0575E-03	2.5879E-03	2.5641E-03	-5.1223E-04	-3.0298E-02
5.6029E+08	9.6236E-04	2.3375E-03	2.3297E-03	5.8006E-05	-1.7453E-02
4.9449E+08	1.1950E-03	2.9858E-03	2.9147E-03	-1.5168E-03	-5.2360E-02
6.6028E+08	1.3574E-03	3.5262E-03	3.3290E-03	-3.1585E-03	-8.7266E-02
7.7049E+08	1.4718E-03	3.9684E-03	3.5821E-03	-4.8428E-03	-0.1222
8.3983E+08	1.5559E-03	4.3377E-03	3.6998E-03	-6.5497E-03	-0.1571
8.8151E+08	1.6208E-03	4.6579E-03	3.7059E-03	-8.2659E-03	-0.1920
9.0469E+08	1.6726E-03	4.9454E-03	3.6173E-03	-9.9832E-03	-0.2269
9.1519E+08	1.7147E-03	5.2106E-03	3.4450E-03	-1.1696E-02	-0.2618
9.2888E+08	1.7490E-03	5.4600E-03	3.1959E-03	-1.3400E-02	-0.2967
9.9700E+08	1.7768E-03	5.6977E-03	2.8747E-03	-1.5093E-02	-0.3316
1.0607E+09	1.7988E-03	5.9263E-03	2.4847E-03	-1.6770E-02	-0.3665
1.1203E+09	1.8156E-03	6.1473E-03	2.0282E-03	-1.8430E-02	-0.4014
1.1761E+09	1.8277E-03	6.3620E-03	1.5073E-03	-2.0070E-02	-0.4363

1.2281E+09	1.8354E-03	6.5708E-03	9.2328E-04	-2.1688E-02	-0.4712
1.2767E+09	1.8390E-03	6.7743E-03	2.7763E-04	-2.3282E-02	-0.5061
1.3218E+09	1.8388E-03	6.9725E-03	-4.2843E-04	-2.4848E-02	-0.5411
1.3653E+09	1.8351E-03	7.1656E-03	-1.1938E-03	-2.6386E-02	-0.5760
1.4076E+09	1.8281E-03	7.3535E-03	-2.0172E-03	-2.7892E-02	-0.6109
1.4470E+09	1.8183E-03	7.5362E-03	-2.8976E-03	-2.9365E-02	-0.6458
1.4835E+09	1.8059E-03	7.7135E-03	-3.8338E-03	-3.0803E-02	-0.6807
1.5172E+09	1.7911E-03	7.8852E-03	-4.8247E-03	-3.2203E-02	-0.7156
1.5533E+09	1.7744E-03	8.0512E-03	-5.8689E-03	-3.3564E-02	-0.7505
1.5868E+09	1.7560E-03	8.2112E-03	-6.9652E-03	-3.4883E-02	-0.7854
1.6176E+09	1.7363E-03	8.3651E-03	-8.1123E-03	-3.6159E-02	-0.8203
1.6504E+09	1.7156E-03	8.5128E-03	-9.3087E-03	-3.7390E-02	-0.8552
1.6830E+09	1.6943E-03	8.6540E-03	-1.0553E-02	-3.8574E-02	-0.8901
1.7133E+09	1.6726E-03	8.7886E-03	-1.1844E-02	-3.9709E-02	-0.9250
1.7494E+09	1.6509E-03	8.9164E-03	-1.3179E-02	-4.0794E-02	-0.9599
1.7832E+09	1.6295E-03	9.0375E-03	-1.4557E-02	-4.1827E-02	-0.9948
1.8233E+09	1.6087E-03	9.1516E-03	-1.5977E-02	-4.2806E-02	-1.030
1.8630E+09	1.5889E-03	9.2589E-03	-1.7436E-02	-4.3730E-02	-1.065
1.9125E+09	1.5701E-03	9.3592E-03	-1.8933E-02	-4.4598E-02	-1.100
1.9657E+09	1.5528E-03	9.4526E-03	-2.0465E-02	-4.5408E-02	-1.134
2.0258E+09	1.5370E-03	9.5391E-03	-2.2030E-02	-4.6159E-02	-1.169
2.1000E+09	1.5231E-03	9.6190E-03	-2.3627E-02	-4.6851E-02	-1.204
2.1854E+09	1.5111E-03	9.6924E-03	-2.5254E-02	-4.7481E-02	-1.239
2.2845E+09	1.5012E-03	9.7594E-03	-2.6907E-02	-4.8050E-02	-1.274

2.3990E+09	1.4935E-03	9.8204E-03	-2.8584E-02	-4.8556E-02	-1.309
2.5349E+09	1.4880E-03	9.8756E-03	-3.0284E-02	-4.8999E-02	-1.344
2.6904E+09	1.4848E-03	9.9254E-03	-3.2003E-02	-4.9379E-02	-1.379
2.8607E+09	1.4839E-03	9.9703E-03	-3.3739E-02	-4.9694E-02	-1.414
3.0589E+09	1.4854E-03	1.0011E-02	-3.5490E-02	-4.9944E-02	-1.449
3.2690E+09	1.4891E-03	1.0047E-02	-3.7252E-02	-5.0129E-02	-1.484
3.5053E+09	1.4952E-03	1.0081E-02	-3.9023E-02	-5.0249E-02	-1.518
3.7633E+09	1.5034E-03	1.0111E-02	-4.0799E-02	-5.0304E-02	-1.553
4.0401E+09	1.5140E-03	1.0140E-02	-4.2579E-02	-5.0293E-02	-1.588
4.3384E+09	1.5267E-03	1.0168E-02	-4.4359E-02	-5.0218E-02	-1.623
4.6608E+09	1.5417E-03	1.0195E-02	-4.6136E-02	-5.0076E-02	-1.658
5.0109E+09	1.5589E-03	1.0223E-02	-4.7906E-02	-4.9870E-02	-1.693
5.3925E+09	1.5785E-03	1.0253E-02	-4.9667E-02	-4.9599E-02	-1.728
5.8105E+09	1.6005E-03	1.0285E-02	-5.1415E-02	-4.9263E-02	-1.763
6.2704E+09	1.6250E-03	1.0322E-02	-5.3148E-02	-4.8862E-02	-1.798
6.7788E+09	1.6521E-03	1.0364E-02	-5.4860E-02	-4.8397E-02	-1.833
7.3437E+09	1.6819E-03	1.0413E-02	-5.6549E-02	-4.7869E-02	-1.868
7.9742E+09	1.7143E-03	1.0470E-02	-5.8212E-02	-4.7278E-02	-1.902
8.6810E+09	1.7491E-03	1.0537E-02	-5.9843E-02	-4.6624E-02	-1.937
9.4764E+09	1.7859E-03	1.0616E-02	-6.1441E-02	-4.5910E-02	-1.972
1.0374E+10	1.8242E-03	1.0708E-02	-6.3000E-02	-4.5136E-02	-2.007
1.1390E+10	1.8631E-03	1.0814E-02	-6.4518E-02	-4.4305E-02	-2.042
Xforce3	Yforce3	Xforce5	Yforce5	ZMoment3	Zmome

0.9787	-0.3347	-0.9787	0.3347	-4.7121E-02	4.1809E-02
0.9787	-0.3347	-0.9787	0.3347	-4.7121E-02	4.1809E-02
1.972	1.705	-1.972	-1.705	-5.7641E-02	4.4153E-02
-0.7198	-3.427	0.7198	3.427	-3.0569E-02	3.6918E-02
-3.211	-7.200	3.210	7.198	-8.5401E-03	2.7942E-02
-5.325	-9.748	5.325	9.747	8.8893E-03	1.8915E-02
-7.074	-11.38	7.074	11.38	2.2877E-02	1.0837E-02
-8.533	-12.39	8.533	12.39	3.4502E-02	4.0528E-03
-9.774	-12.98	9.774	12.98	4.4524E-02	-1.4188E-03
-10.85	-13.27	10.85	13.27	5.3428E-02	-5.6685E-03
-11.80	-13.35	11.80	13.35	6.1516E-02	-8.8052E-03
-12.64	-13.28	12.64	13.28	6.8982E-02	-1.0928E-02
-13.40	-13.08	13.40	13.08	7.5948E-02	-1.2118E-02
-14.07	-12.79	14.07	12.79	8.2492E-02	-1.2443E-02
-14.67	-12.42	14.67	12.42	8.8667E-02	-1.1953E-02
-15.20	-12.00	15.20	12.00	9.4505E-02	-1.0690E-02
-15.67	-11.53	15.67	11.53	0.1000	-8.6849E-03
-16.07	-11.02	16.07	11.02	0.1052	-5.9644E-03
-16.40	-10.49	16.40	10.49	0.1102	-2.5482E-03
-16.67	-9.941	16.67	9.941	0.1148	1.5478E-03
-16.88	-9.386	16.88	9.386	0.1191	6.3108E-03
-17.02	-8.831	17.02	8.832	0.1231	1.1731E-02
-17.10	-8.286	17.10	8.287	0.1268	1.7800E-02
-17.12	-7.760	17.12	7.760	0.1302	2.4512E-02

-17.07	-7.260	17.07	7.260	0.1333	3.1864E-02
-16.97	-6.796	16.97	6.796	0.1360	3.9852E-02
-16.81	-6.377	16.81	6.377	0.1384	4.8475E-02
-16.59	-6.012	16.59	6.012	0.1405	5.7734E-02
-16.32	-5.710	16.32	5.711	0.1422	6.7632E-02
-16.00	-5.483	16.01	5.484	0.1436	7.8173E-02
-15.64	-5.340	15.64	5.340	0.1447	8.9366E-02
-15.24	-5.292	15.25	5.292	0.1455	0.1012
-14.81	-5.350	14.81	5.351	0.1459	0.1137
-14.35	-5.527	14.35	5.528	0.1461	0.1270
-13.87	-5.836	13.88	5.837	0.1459	0.1409
-13.39	-6.291	13.39	6.292	0.1455	0.1556
-12.89	-6.907	12.89	6.908	0.1448	0.1710
-12.41	-7.700	12.41	7.702	0.1439	0.1873
-11.95	-8.691	11.95	8.692	0.1429	0.2044
-11.52	-9.900	11.52	9.901	0.1417	0.2225
-11.14	-11.35	11.14	11.35	0.1404	0.2415
-10.82	-13.07	10.82	13.07	0.1390	0.2616
-10.59	-15.10	10.59	15.10	0.1377	0.2829
-10.46	-17.47	10.46	17.47	0.1365	0.3055
-10.47	-20.22	10.47	20.22	0.1354	0.3295
-10.64	-23.42	10.64	23.42	0.1347	0.3550
-11.01	-27.13	11.01	27.13	0.1343	0.3823
-11.63	-31.43	11.63	31.43	0.1344	0.4116

-13.84	-42.23	13.84	42.23	0.1367	0.4772
-15.58	-49.01	15.58	49.01	0.1393	0.5143
-17.90	-56.97	17.90	56.97	0.1431	0.5547
-20.91	-66.36	20.91	66.36	0.1485	0.5991
-24.83	-77.51	24.83	77.51	0.1556	0.6481
-29.88	-90.85	29.88	90.85	0.1651	0.7026
-36.40	-107.0	36.40	107.0	0.1772	0.7636
-44.84	-126.6	44.84	126.6	0.1926	0.8325
-55.82	-150.8	55.82	150.8	0.2119	0.9108
-70.23	-181.1	70.23	181.1	0.2361	1.001
-89.34	-219.4	89.33	219.4	0.2662	1.105
-115.0	-268.8	115.0	268.8	0.3036	1.228

Matlab Code for Plotting Force and Stress Plots

Modify the “text” value to change files used to generate the plots

```
%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%
%%%%%%%%%%

% Ansys data analysis file                    %

% For an Ansys batch file                     %

% which produces an output file named knee_output.txt    %

% Version 1: May 18,2007                      %

%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%
%%%%%%%%%%
```

```
clear all
  text = '50deg'
  type = 'interp'
%   type = 'flat'
 filename = ['knee_force_plot_',text,'.txt'];
 string1 = ['C:\DOCUME~1\smahler\MYDOCU~1\lusk_ansys\ansys_output_files\']
fid1 = fopen([string1,filename]);    % opens the file
ABT = fread(fid1);                    % reads the file into variable ABT
fclose(fid1);                         %closes the data file
GBT = native2unicode(ABT)';          %changes data from machine code to text
s_iB = findstr('Rot6', GBT);          % finds end of header
s_if = findstr('Xforce6', GBT);          % finds end of header
s_iB2 = findstr('Yforce6', GBT);          % finds end of header
A=str2num(GBT(s_iB(end)+5:s_if(end)-1))% turns the data into a numerical matrix
A2=str2num(GBT(s_iB2(end)+9:end))% turns the data into a numerical matrix


%figure(1)
                %plot(90-A(1:50,1)*180/pi,A(1:50,2))
    %plot(A(:,1))
    %axis equal
    %title('stress')


figure(1)
clf
```

```
x=A(:,4)'

y=A(:,5)'

% u_sign = sign(A2(:,1)');

% v_sign = sign(A2(:,2)');

% u=log10(abs(A2(:,1)')).*u_sign;

% v=log10(abs(A2(:,2)')).*u_sign;

 u=A2(:,1)';

 v=A2(:,2)';

%C=(A(:,2).^2+A(:,3).^2).^.5;



xp = [];

yp= [];

vp=[];

up=[];

xp1 = [];

yp1= [];

vp1=[];

up1=[];

xp2 = [];

yp2= [];

vp2=[];

up2=[];
```

```
xp3 = [];

yp3= [];

vp3=[];

up3=[];

xp4 = [];

yp4= [];

vp4=[];

up4=[];


%1st run

for i = 1:5,

ls = 22

sv = 2

xp4 = [xp4;[x(ls*(i-1)+5+sv:ls*i+sv-6-1)]]

yp4 = [yp4;[y(ls*(i-1)+5+sv:ls*i+sv-6-1)]]

vp4 = [vp4;[v(ls*(i-1)+5+sv:ls*i+sv-6-1)]]

up4 = [up4;[u(ls*(i-1)+5+sv:ls*i+sv-6-1)]]

end

%2nd run

for i = 1:6,

ls = 32

sv =115

xp = [xp;[x(ls*(i-1)+sv+24:-1:ls*(i-1)+sv+9)];[x(ls*(i-1)+sv+25:ls*(i-
1)+sv+32)],[x(ls*(i)+sv+1:ls*(i)+sv+8)]]
```

```
yp = [yp;[y(ls*(i-1)+sv+24:-1:ls*(i-1)+sv+9)];[y(ls*(i-1)+sv+25:ls*(i-1)+sv+32)],[y(ls*(i)+sv+1:ls*(i)+sv+8)]]

up = [up;[u(ls*(i-1)+sv+24:-1:ls*(i-1)+sv+9)];[u(ls*(i-1)+sv+25:ls*(i-1)+sv+32)],[u(ls*(i)+sv+1:ls*(i)+sv+8)]]

vp = [vp;[v(ls*(i-1)+sv+24:-1:ls*(i-1)+sv+9)];[v(ls*(i-1)+sv+25:ls*(i-1)+sv+32)],[v(ls*(i)+sv+1:ls*(i)+sv+8)]]

end

i=7

xp = [xp;[x(ls*(i-1)+sv+25:ls*(i-1)+sv+32)],[x(ls*(i)+sv+1:ls*(i)+sv+8)]]

yp = [yp;[y(ls*(i-1)+sv+25:ls*(i-1)+sv+32)],[y(ls*(i)+sv+1:ls*(i)+sv+8)]]

up = [up;[u(ls*(i-1)+sv+25:ls*(i-1)+sv+32)],[u(ls*(i)+sv+1:ls*(i)+sv+8)]]

vp = [vp;[v(ls*(i-1)+sv+25:ls*(i-1)+sv+32)],[v(ls*(i)+sv+1:ls*(i)+sv+8)]]

sv=ls*(i-1)+sv+1

%3rd run

for i = 1:5,

ls = 31

%xp1 = [xp1;[x(ls*(i-1)+sv:ls*i+sv-1)]]

%yp1 = [yp1;[y(ls*(i-1)+sv:ls*i+sv-1)]]

%vp1 = [vp1;[v(ls*(i-1)+sv:ls*i+sv-1)]]

%up1 = [up1;[u(ls*(i-1)+sv:ls*i+sv-1)]]

xp1 = [xp1;[x(ls*(i-1)+sv+22:-1:ls*(i-1)+sv+8)];[x(ls*(i-1)+sv+23:ls*(i-1)+sv+30)],[x(ls*(i)+sv+1:ls*(i)+sv+7)]]

yp1 = [yp1;[y(ls*(i-1)+sv+22:-1:ls*(i-1)+sv+8)];[y(ls*(i-1)+sv+23:ls*(i-1)+sv+30)],[y(ls*(i)+sv+1:ls*(i)+sv+7)]]

up1 = [up1;[u(ls*(i-1)+sv+22:-1:ls*(i-1)+sv+8)];[u(ls*(i-1)+sv+23:ls*(i-1)+sv+30)],[u(ls*(i)+sv+1:ls*(i)+sv+7)]]

vp1 = [vp1;[v(ls*(i-1)+sv+22:-1:ls*(i-1)+sv+8)];[v(ls*(i-1)+sv+23:ls*(i-1)+sv+30)],[v(ls*(i)+sv+1:ls*(i)+sv+7)]]
```

```
end

i=7
xp1 = [xp1;[x(ls*(i-1)+sv+22:-1:ls*(i-1)+sv+8)]]
yp1 = [yp1;[y(ls*(i-1)+sv+22:-1:ls*(i-1)+sv+8)]]
up1 = [up1;[u(ls*(i-1)+sv+22:-1:ls*(i-1)+sv+8)]]
vp1 = [vp1;[v(ls*(i-1)+sv+22:-1:ls*(i-1)+sv+8)]]
%4th run
sv=ls*i+sv+6
for i = 1:15,
ls = 22
xp2 = [xp2;[x(ls*(i-1)+sv+10:-1:ls*(i-1)+sv)];[x(ls*(i-1)+sv+11:ls*(i-1)+sv)+21]]
yp2 = [yp2;[y(ls*(i-1)+sv+10:-1:ls*(i-1)+sv)];[y(ls*(i-1)+sv+11:ls*(i-1)+sv)+21]]
vp2 = [vp2;[v(ls*(i-1)+sv+10:-1:ls*(i-1)+sv)];[v(ls*(i-1)+sv+11:ls*(i-1)+sv)+21]]
up2 = [up2;[u(ls*(i-1)+sv+10:-1:ls*(i-1)+sv)];[u(ls*(i-1)+sv+11:ls*(i-1)+sv)+21]]
end

%5th run

% for i = 1:3,
```

```
% ls = 24
% xp3 = [xp3;[x(ls*(i-1)+sv:ls*i+sv-1)]]
% yp3 = [yp3;[y(ls*(i-1)+sv:ls*i+sv-1)]]
% vp3 = [vp3;[v(ls*(i-1)+sv:ls*i+sv-1)]]
% up3 = [up3;[u(ls*(i-1)+sv:ls*i+sv-1)]]
% end
Cp = (up.^2+vp.^2).^.5;
Cp1 = (up1.^2+vp1.^2).^.5;
Cp2 = (up2.^2+vp2.^2).^.5;
Cp3 = (up3.^2+vp3.^2).^.5;
Cp4 = (up4.^2+vp4.^2).^.5;


clf
hold on


 pcolor(xp1,yp1,log10(Cp1))
 shading(type)


 pcolor(xp,yp,log10(Cp))
 shading(type)
```

```
 pcolor(xp4,yp4,log10(Cp4))
shading(type)


%colorbar('horiz')
%[cmin cmax] = caxis;
quiver(x,y,u,v,'k')


% %colorbar('horiz')
% %caxis([cmin cmax]);
%
 pcolor(xp2,yp2,log10(Cp2))
shading(type)
 colorbar('horiz')
 %caxis([cmin cmax]);
%axis equal
axis('tight')
%[c,h]=contour(xp,yp,Cp)
%clabel(c,h)
%for j=1:length(x),
%   text(x(j),y(j),num2str(j))
%end
fname = ['forcemag_',text,'_',type];
print('-dtiff', '-r600', fname)
```