ABSTRACT

A Review of Energy Storing Prosthetic Feet and Computer Aided Structural Optimization of a Below-knee Prosthesis.

by
Poonam Gope Lalwani

Because people with physical disabilities have shown an interest in participation in sports, a new class of prosthetic feet known as "energy storing prosthetic feet" has been developed. These new developments in prosthetic foot design utilize energy storage and return to improve ambulation. This thesis reviews the design, materials, advantages and disadvantages of various energy storing prosthetic feet. Research studies, comparing gait in below-knee amputees using different prosthetic designs, can be applied to the design of prosthetic feet that are lighter, stronger and more reliable. Comparisions among these feet are reviewed in the context of functional capability and patient satisfaction. This study indicates a significant improvement in the amputees overall function with the use of energy storing prosthetic feet compared to the conventional feet.

In this thesis, a model of a below-knee prosthesis is constructed and its response to two different loading conditions studied by finite element stress analysis using the Computer Aided Engineering package of I-DEAS. The main criterion in the design of a prosthesis is a balance between minimizing stress and weight, for a required level of functional capability. The effect of different geometry, material properties and loading conditions on minimizing the weight of the prosthesis and on stress distribution within the prosthesis is determined. An optimal prosthesis with minimum weight is designed for use by geriatric amputees.

A REVIEW OF ENERGY STORING PROSTHETIC FEET AND COMPUTER AIDED STRUCTURAL OPTIMIZATION OF A BELOW-KNEE PROSTHESIS

by
Poonam Gope Lalwani

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This Thesis is dedicated to My Wonderful Family

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TABLE OF CONTENTS

LIST OF TABLES

LIST OF FIGURES

Figure **Page**

Figure	**Page**

CHAPTER 1
INTRODUCTION

1.1 Introduction to Energy Storing Prosthetic Feet

Presently, there is an increased demand among amputees for a prostheses that would allow them to participate in sports. Recent developments in the design of energy-storing feet have enabled lower-extremity amputees to significantly increase and improve their ability to jog and run. New prosthetic materials have resulted in improvements in prosthetic durability, weight and cosmesis.

Energy storing prosthetic feet store energy during stance and release the energy as body weight progresses forward, thus helping to passively propel the limb. Manufacturers claim that these feet increase the duration of daily use of the prostheses and improve overall amputee function and satisfaction. These feet reduce the energy consumption of gait. This is especially important when considering prosthetic prescription for the young athlete and the compromised geriatric amputee.

Requirements for energy-storing prosthetic feet vary because of different physical characteristics and activity objectives of patients. The amputees age, body weight, activity level and the level of amputation are major considerations in the selection of an energy storing prostheses. Proper selection of a prosthetic foot allows an appropriate balance of energy storage and release over a wide range of walking and running speeds. Design, materials, alignment and suspension are all contributing factors in the success of an energy storing prostheses. This thesis will review the design philosophy, materials and applications of five different types of energy-storing prosthetic feet that are now commercially available.

1.2 Comparative Analysis of Energy Storing Feet

At present, there is limited information available that describes how energy storing prosthetic feet perform in achieving optimal and symmetrical gait. Several methods have been used to evaluate and compare the advantages and disadvantages of different energy storing feet. Most of these comparison studies have focused on the extent by which amputees benefit from wearing different prosthetic feet, as far as load-bearing structure, metabolic cost of walking, aesthetics and comfort are concerned. Significant research has been published so far in this field and some of these comparative research studies will be reviewed in this thesis with respect to improvement in the amputees overall function and satisfaction with the prosthesis.

1.3 Purpose of this Thesis

In order to design and develop a model of a lower-limb prosthesis, it is very important to incorporate analytical tools such as mechanical Computer-Aided Engineering into the design process at an early stage. This allows us to assess the loads imposed on the musculo-skeletal system due to the use of a given prosthesis-amputee interface. In this study, an optimal prosthesis with minimum weight will be designed for use by geriatric amputees.

A finite element model of a lower-limb prosthesis and altered variations of the model will be developed using the Computer Aided Engineering package of I-DEAS. The models will consist of shell elements of 0.5 cm thickness. The original finite element model will be used to investigate the stress distributions on the prosthesis for different prosthetic materials. Finite element stress analysis will be performed on the models in order to determine the response of the prosthesis to load during weight-bearing and during heel strike.

The objective of this study is to determine the effects of altering the model on the stress distribution and magnitudes within the model and on the weight of the model. Since the prosthesis with the minimum mass is considered to be the best possible design for geriatic amputees, the structural mass of the models will be determined to select the optimum design. The original design will be modified to decrease weight and improve function. A hollow model of the prosthesis will be constructed and its weight determined. Changes in the model that result in minimizing the weight of the prosthesis will be identified.

CHAPTER 2
ENERGY STORING PROSTHETIC FEET

2.1 Introduction to Energy Storing Prosthetic Feet

Anatomists, biomechanical engineers, and clinicians have studied the foot and ankle complex for centuries. Each discipline has provided its unique insight into the structure and function of this unit (1). The human foot is a very complex structure. The pair contain 52 separate bones, dozens of intrinsic muscles, and scores of extrinsic muscles. The feet are composed of multiple layers of ligaments, fascia, and muscle, and contain numerous interrelated articulations (2).

Hundreds of times a day, with every step we take, we crash down upon our heel with a force that often reaches several times our total weight. The foot sustains these impacts and reduces potential injury to the body by deforming upon striking the ground. If the foot were extremely rigid, the ground reaction force would be of great magnitude and short duration. The foot, however, is restrained by flexible tendons, and its many bones are held together by flexible ligaments.

As the foot deforms, the ligaments and tendons stretch to absorb much of the shock, and the impulse is a more sustained force of smaller amplitude, reducing the potential for injury to the body. During running our muscles provide kinetic and potential energy as we simultaneously speed up and rise, but later in the gait cycle this energy is lost as we slow down. Ligaments and tendons store some of this energy and return it during the cycle, reducing the work our muscles must do (3). Thus, the foot-ankle complex provides the dual functions of support and propulsion by combining the characteristics of flexibility and rigidity as the foot adapts to the gait cycle.

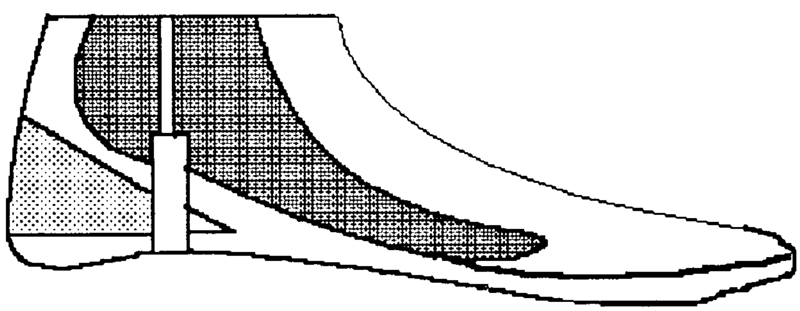


Figure 2.1 The SACH Foot

Although there have been many attempts to imitate the foot-ankle complex structure, very few designs have been widely accepted. All prosthetic feet simulate the general contour of the human counterpart. However, they differ in the internal design characteristics, which enable them to simulate some actions of the human foot.

Several functions are common to all prosthetic feet. They all provide 1) a base of support when the wearer stands or is in the stance phase of gait, 2) shock absorption at heel-strike as the device plantar flexes, and 3) simulation of the push-off phase as the device dorsiflexes. Foot motion occurs passively, in response to the load applied by the wearer (4).

2.1.1 The SACH Foot

The conventional SACH foot (4) has for years been the industry standard. It was developed at the University of California, Berkley in 1950's. The Solid-Ankle Cushion Heel (SACH) shown in Figure 2.1 consists of a central rigid heel (solid ankle), a posterior wedge of resilient material (cushion heel) and a covering of resilient synthetic rubber.

The compressible heel enables the SACH foot to bend only slightly and absorb shocks on uneven surfaces. However, the rigid keel prevents ankle dorsiflexion. The rate of plantarflexion after heel-strike determines the time required for the amputee to place his foot flat on the ground and achieve stability. Plantarflexion is delayed in the SACH foot. As a result, the amputee has to apply substantial load to the prosthesis to achieve the flat-foot phase of stance and thereby gain knee stability (4). The SACH foot is available in a variety of sizes, weights and heel heights. It has no moving parts and requires little maintenance. Because of its simple design and durability, the SACH foot is the most popular prosthetic foot in the U.S.

Traditional prosthetic feet such as the SACH foot were designed only for walking. Most amputees, even young athletic amputees, were not expected to run or participate in sports. Even highly motivated amputees were severely limited in their activities because of a lack of appropriate prosthesis and training (5). Little or no exercise often led to excessive weight gain, hypertension and abnormal glucose tolerance among amputees.

The major problem encountered in sports is running. During running, the downward force during heel-strike exceeds body weight by two to three times. Similarly large differences are seen between normal and amputee gait. The loading of the foot that occurs during running leads to further deterioration of the amputee gait pattern as speed increases. An asymmetrical gait could contribute to the development of degenerative disease. This type of loading may also shorten the life of the prosthesis. Thus any design of prosthetic feet should be able to deal with the loading of the feet that occurs during running. In addition amputees are unable to push off after foot-flat. Therefore, prosthetic feet should be able to simulate the push-off phase of normal running. It is also important for prosthetic feet to be as light as possible, without sacrificing strength.

In the past several years there have been numerous advances in the development of prosthetic feet for the amputee. New materials having better mechanical properties and reduced weight are now available. New designs in prosthetic feet assemblies now offer amputees a much wider choice than was the case a few years ago. Because amputees have increased participation in sports, there is a demand for prosthetic feet that will endure and improve athletic performance. This has resulted in the development of a new class prosthetic feet known as "energy-storing" prosthetic feet.

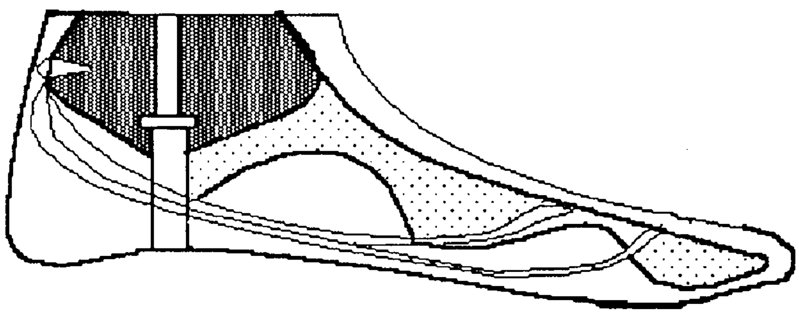


Figure 2.2 The S.A.F.E. Foot

Energy-storing prosthetic feet (ESPF) represent an attempt to approach normal running gait patterns by responding to the downward force during heel-strike. The energy stored during heel contact is later used during push-off to initiate the swing phase of the gait cycle, while increasing forward acceleration of the leg and body. This allows the patient to walk and run smoothly and conserve more energy than when using traditional prosthetic feet (6). ESPF have more dynamic action than the SACH foot, which for years has been the industry standard.

Design, materials, alignment and suspension all contribute to a successful energy-storing prosthesis. The components can be selected to obtain the most suitable balance of energy storage and release over a wide range of walking and running speeds. Energy-storing prosthetic feet vary in performance levels and can vary from providing a little energy storage, to a great deal of energy storage. The amputee's age, body weight ,activity level and the level of amputation are major considerations in the selection of an energy storing prosthetic foot.

2.2 Types of Energy Storing Prosthetic Feet

There are two basic types of energy storing prosthetic feet : 1) models that are bolted to conventional prosthesis - Solid Ankle Flexible Endo-skeletal (S.A.F.E.) Foot, Seattle Foot, Stored Energy (STEN) Foot and Carbon Copy II Foot, and 2) models that incorporate the foot and pylon into a single unit which is attached to the socket e.g. Flex-Foot.

2.2.1 S.A.F.E. Foot

The S.A.F.E. Foot (Figure 2.2) was developed by Campbell and Childs in 1979. It was designed and built to meet five predetermined criteria : a dome

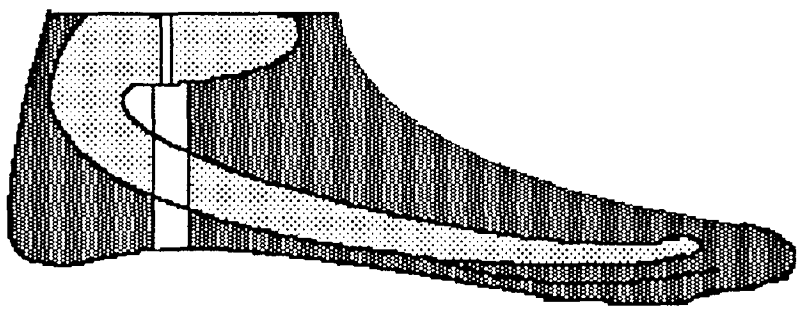


Figure 2.3 The Seattle Foot

shaped arch, a long plantar ligament band, a flexible endoskeleton, a subtalar joint and made entirely out of plastic materials with no mechanical joints (8). All of these criteria were met in the design. It has a keel composed of a rigid polyurethane bolt block (Stationary Ankle) joined to a polyurethane elastomer section (Flexible Endoskeletal) at a 45-degree angle in the sagittal plane to simulate the subtalar joint. The S.A.F.E. foot is bolted to the shin through the bolt block. The S.A.F.E. Unit also has two Dacron(polyester fiber) bands on the plantar surface, which tighten at heel-off to make the foot more rigid during late stance.

The flexible keel dissipates energy as it accommodates to irregular surfaces. The S.A.F.E. Foot is designed to conform well to uneven ground by passively inverting and everting and to remain stiff during push-off. It is relatively heavy and noncosmetic in appearance. The S.A.F.E. Foot is probably unparalleled for use on uneven surfaces, and many amputees report an increase in residual limb comfort because it absorbs much of the shock of everyday walking. It has the advantage of requiring no maintenance and being moisture and grit-resistant.

2.2.2 Seattle Foot

The Seattle Foot (Figure 2.3) was designed by the Prosthetic Research Study and Engineers from Boeing Aircraft in 1981. The design specifications have varied over the years as the concept was refined. Originally the keel was a fiberglass multi-leaf design, somewhat similar to an automobile suspension spring. The fiberglass keel extended into the metatarsal area to form a spring, and as the patient walked or ran, the keel deflected and sprang back, thrusting the patient forward (5). As the patient increased his speed, stiffer portions of the spring came into play. Excessive weight, an unacceptable failure rate, and the labor requirements of the fabrication technique were the drawbacks of this version of the Seattle Foot.

The commercial version of the Seattle foot consists of a Delrin keel, a Kevlar reinforcement toe pad and an exterior polyurethane foam matrix. Delrin is a hard plastic (acetal resin) and Kevlar is a strong, lightweight yellow fiber (aramid fiber). The keel extends from the top of the foot, back around the heel, and into the metatarsal area.

Shaped as a leaf spring, the keel compresses during heel contact (stores energy) and extends during push-off (gradually releases stored energy). Delrin provides the necessary combination of strength, lightness, moldability and dampening. Additional keel combinations are available to correspond to the weight and activity of the user.

The Kevlar pad is beneath the heel in the metatarsal area. It is designed to reinforce the metatarsal area and keep the keel from pushing its way through the foam. The polyurethane foam matrix provides the most life-like external replication of any commercially available foot.

The Seattle Foot is designed to control and store energy that is available at heel strike and foot flat, releasing it during push-off to increase the forward movement of the foot. It is suitable for both walking and running, but is especially suited to running because it provides more spring. It provides damping to accommodate to individual gaits. The Seattle foot is available in two styles, an athletic model derived from human feet, and a noncosmetic model which resembles a SACH foot. The cosmetic model can be difficult to fit into some shoes because of its wide profile.

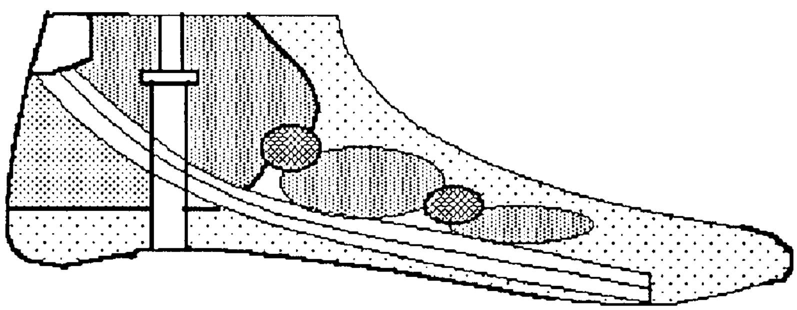


Figure 2.4 The STEN Foot

The Seattle foot weighs just over a pound which is heavier than the Carbon Copy II and Flex-Foot. The return of stored energy exceeds that of the Carbon Copy II and STEN Foot. There is no uniformity in dimension from size to size. The Seattle foot is compatible with most standard prosthetic components and fabrication techniques. It can be tailored to the individual and is available in a wide range of sizes for both men and women. A newer version of Seattle foot is the SLF30 Seattle Light Foot. It weighs one third less than the Seattle foot and is also lighter than most other energy-storing prosthetic feet.

2.2.3 STEN Foot

The STEN Foot (Figure 2.4) consists of : 1) three hardwood blocks that extend to the toe region, 2) two high-density rubber blocks, which act as compressible spacers between the wooden blocks, 3) a layer of high-strength fabric attached to the plantar surface of the wood blocks, 4) a SACH cushion heel wedge, 5) plantar reinforcement fabric bonded to a rubber sole, and 6) an exterior polyurethane foam matrix. The wood/rubber/fabric composite is designed to store energy during the stance phase as the rubber blocks deform. This energy is released when the amputee advances over the foot during push-off.

The STEN Foot accommodates to uneven walking surfaces and smooths stance transition with a mild spring action. It is light and inexpensive compared to the SAFE Foot. The STEN Foot is similar to the SACH Foot in appearance. It is the easiest design to fit in a variety of shoe styles, and comes in the greatest selection of sizes and heel heights.

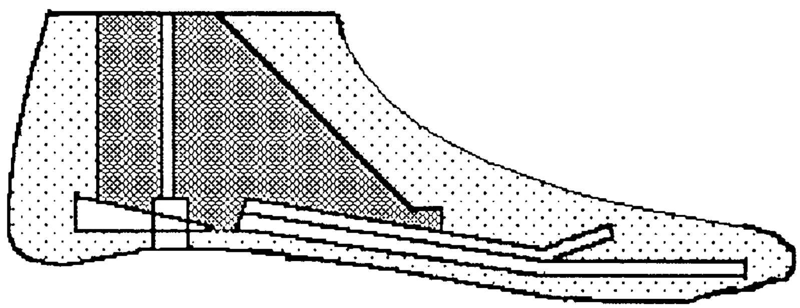


Figure 2.5 The Carbon Copy II Foot

2.2.4 Carbon Copy II Foot

The Carbon Copy II Foot (Figure 2.5) was introduced by Ohio Willow Wood Company in May 1986. The most recent design of Carbon Copy II prosthetic foot is a result of the "better way" design that began at Ohio Willow Wood Company in 1974. It was decided that the new design would be such that it could be used by a wide range of amputees.

New materials, that had better mechanical properties and reduced weight, were beginning to become available. It was therefore decided to use new fiber reinforced composites to assist in initiating heel rise at toe-off and in moving the prosthesis forward. Such composites also allow an amputee's center of gravity to remain at a constant vertical height over a period of time, provided that the material's fatigue life was long. A material that met all the above mentioned criteria was a high performance carbon fiber-epoxy laminate.

The Carbon Copy II Foot consists of a rigid Kevlar bolt, two carbon composite deflection plates, a SACH heel, a Kevlar glide sock and a polyurethane exterior matrix. The keel of the Carbon Copy II Foot consists of block and the deflection plates. This keel is designed to deform from heel-strike to late stance, storing energy like a compressed spring, and then to release this stored energy just before toe-off to initiate swing.

The bolt block of the Carbon Copy II Foot is a special ultralight reinforced Kevlar/nylon design. The failure of the bolt due to fatigue is very common in conventional wooden keels. Due to the mechanical properties of this composite material, failure of the bolt block is considerably reduced. This composite material also has low moisture absorbency. It has high impact resistance and a tendency to dampen vibration associated with impact which permits the keel to act as a shock absorber.

The deflection plates produce a "graded" or two-stage energy storage system. In normal walking, the thin primary deflection plate (which extends to the joints of the toes) provides a gentle energy return. At higher speeds or during more vigorous activities, the auxiliary deflection plate (which curves upward to terminate at the midfoot) provides additional push-off. This provides an amputee with two basic levels of resistance for different walking velocities.

Three durometers of heel cushion are used for simulated plantar flexion. Compression of the flexible methane heel wedge permits a large evenly distributed load base while weight is being transferred to the prosthesis. Kevlar is a material that is extremely resistant to abrasion. A Kevlar glide sock protects the anterior ends of the keel segments to prevent them from piercing the urethane foam body of the foot. The "protection sock" also bonds very well to the flexible methane foam, subsequently acting as a reinforcement fiber for the methane. The outer shell is made up of "life-molds", which is a special microcellular polyurethane elastomer blend and is shaped in molds derived from human feet.

The plantar surface of the Carbon Copy II Foot is broad and flat for maximum mediolateral stability. It does not yield very much to pavement irregularities. The Carbon Copy Foot is the lightest of the "conventional" ESPF feet and is very cosmetic. It is also somewhat more expensive than the SACH and SAFE feet, but comparable in cost to the Seattle foot. The Carbon Copy II Foot is very versatile and is suitable for many levels of amputation.

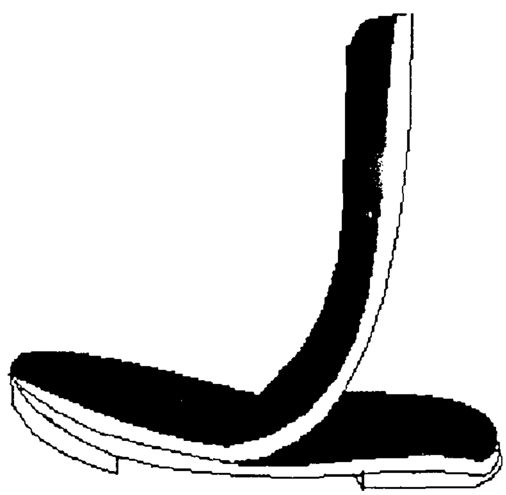


Figure 2.6 The Flex Foot

2.2.5 Flex-Foot

The Flex-Foot (Figure 2.6) is a light-weight low-modulus carbon-graphite composite structure. It has a graphite composite keel that is the core of the foot and shank. Graphite composite was chosen because it can be designed thin enough to flex, and thick enough to maintain structural rigidity.

The Flex-Foot consists of two flat, broad carbon leaves in a soft foam cover. A large main leaf extends from the base of the prosthetic socket (below knee) or knee unit (above knee) through the ankle and into the toe region of the foot. The smaller leaf is attached to the larger leaf at the level of the mid-foot and extends posteriorly, forming a heel component.

The leafspring shape of the Flex-Foot prosthesis stores and releases energy as it flexes. The smaller posterior leaf is designed to attenuate the shock of heel-strike and force the prosthesis forward. The main leaf is designed to dorsiflex under stress during stance phase and to extend forcibly during push-off.

The entire prosthesis bends as downward force is applied. The Flex-Foot utilizes the entire distance distal to the socket for function. Since it stores energy throughout its entire length rather than just within a four inch keel, this results in a very responsive and resilient component.

The Flex-Foot is available in an " original " model in which the carbon leaf is bonded to the prosthetic socket and a " modular " model in which the carbon leaf is bolted to the socket. The weight of the Flex-Foot prosthesis is less than a "conventional" ESPF-equipped prosthesis and the mass (less than 1lb) is concentrated proximally. Actual weight savings of 10-15 % are common, but patients typically perceive that the Flex-Foot weighs "half-as-much".

The Flex-Foot is hand made from a computer generated design specific to each individual patient. Data such as weight, activity level, and residual limb characteristics determine the specific orientation and thickness of reinforcement fibers. Ultra high pressure, high temperature molding insures the greatest possible strength to weight ratio, but requires several weeks for fabrication. Although this is a very costly approach, it does permit fitting the widest range of individuals.

The Flex-Foot offers the highest level of energy storage. This is due to its long spring leaf design and high modulus carbon construction. It is very responsive, provides the smoothest running and walking gait patterns, and allows the highest jumping performance of all the ESPF. It makes walking, up inclines, much easier for patients. It has been reported that for below-knee amputees, walking with the Flex Foot conserves energy at higher walking velocities resulting in lower relative levels of exercise intensity and enhanced gait efficiencies (13). This, in combination with its light weight, makes it a good alternative for many above- and below-knee amputees.

The chief restriction of the Flex-Foot is that it requires a minimum of five inches from the end of the residual limb to the floor, and seven inches or greater is preferred. Thus, the Flex-Foot is not suitable for small children, Symes and similar amputations, and very long below-knee residual limbs. Patients may require a longer time to adjust to the Flex-Foot because it may force them to walk more quickly. Some other disadvantages are high cost and complexity in fabrication and alignment. Successful use of the Flex-Foot requires a high level of commitment on the part of the patient and prosthetist.

CHAPTER 3
COMPARATIVE ANALYSIS
OF ENERGY STORING FEET

3.1 Introduction to Comparative Analysis of ESPF

The desire of amputees to participate in sports, and the high demands of athletics, have resulted in the development of energy-storing prosthetic feet. These prosthetic feet store energy during stance and release energy during push-off as body weight progresses forward, thus helping to thrust the prostheses forward.

New prosthetic materials and designs have broadened the range of ESPF available. As a result, it is more difficult for prosthetists and physicians to choose the best foot for individual amputees (14). In one aspect however the advantage of energy storing prosthetic feet over conventional prosthetic feet is obvious : as a rule, amputees do not run when fitted with conventional prostheses : many of those fitted with ESPF can and do run very well (15). There is a decrease in the effort required for walking due to decreased weight and increased responsiveness of the energy storing prosthetic feet.

Design, selection and alignment of an energy storing prosthetic foot should be directed towards obtaining optimal gait. Although ESPF have achieved widespread clinical acceptance, the effect of these feet on the biomechanics of below-knee amputee gait is poorly understood. It has been established that comparison of sound-limb to prosthetic-limb symmetry is the best method to analyze and evaluate gait (16).

Presently there is limited information available that describes how energy storing feet perform in achieving optimal and symmetrical gait. Also there is no universally accepted definition of an energy storing prosthetic foot.

Besides observations and anecdotes, there is little evidence available that shows energy savings at all. Most of the research done in evaluating ESPF is subjective and the relative merits of different prosthetic feet are unclear. Therefore it is very important to be able to evaluate and rank the various designs quantitatively and scientifically to justify selection of a particular prosthetic foot.

3.2 Methods of Comparative Analysis of ESPF

Quantitative research, comparing gait in below-knee amputees using different energy storing feet, has been performed in the following areas :

1. Biomechanical analysis
2. Energy-Cost measures
3. Subjective preferences

3.2.1 Biomechanical analysis

Biomechanical evaluation consists of dynamic evaluation of the foot through motion analysis and evaluation of the forces created by and acting on the body when wearing a prosthetic foot (17). In the modern biomechanics laboratory, it is possible to measure the kinetic and kinematic gait characteristics of amputees by using motion analysis. This gives us the opportunity to test different prosthetic feet under the conditions in which they will ultimately be used. The walking gait of amputees differs significantly from that of normal individuals. It is very important to identify these differences (or variables) in order to evaluate and compare ESPF. Some of these variables are stride characteristics, kinematic variables and foot-floor contact forces. Stride characteristics include velocity, cadence, step length, gait-cycle duration, single-limb stance, etc.

Several different researchers have measured these variables in an effort to evaluate different prosthetic feet. Torburn *et al.* (18) investigated the gait biomechanics of five below-knee amputees wearing four different energy storing feet and a SACH foot. Only cadence and gait-cycle duration were slightly different for the five feet during free-velocity walking. Cadence was greater (and gait-cycle duration shorter) for the Carbon Copy ll Foot than for the SACH or Flex-foot. Cadence was 102 steps/min for Carbon Copy II versus 98 steps/min for the SACH and Flex-foot. However, this difference is not clinically significant.

The Flex-foot had greater dorsiflexion (Figure 3.1) at the ankle joint during late stance compared to the four other feet. There was no significant difference in torque or motion at the hip or knee joints. Thus no clinical significant changes in gait were detected in this study.

Macfarlane, Nielsen *et al.* (19) compared the gait of seven below-knee amputees using flex-foot versus a conventional foot. Single-limb stance was longer for the flex-foot as compared to the conventional foot (Figure 3.2). Also walking with the flex-foot resulted in a smoother vertical trunk motion. This suggests an increase in the biomechanical efficiency of the flex-foot when compared to the conventional foot.

Menard *et al.* (15) performed gait analysis on eight below-knee amputees and nine control subjects to compare energy-storing capabilities of the Seattle Foot and the Flex-Foot. Larger late vertical forces and smaller later anterior-posterior and medial-lateral forces were observed when the Flex-Foot was used. The researchers believed that this was due to the forces produced by the Flex-Foot very late in stance. These forces are responsible for the liveliness and springiness of the Flex-Foot during sports and athletic activities.

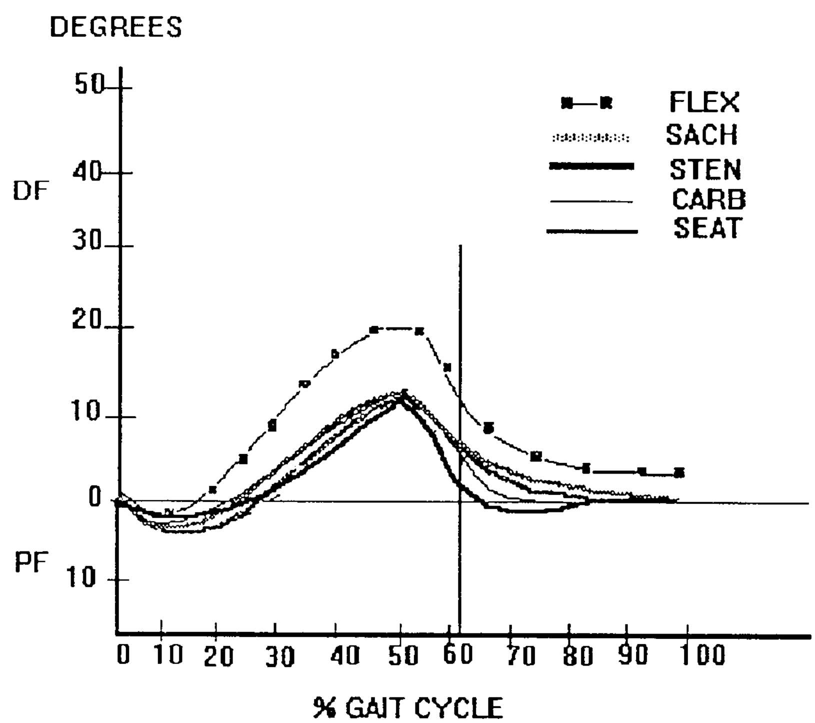


Figure 3.1 Dorsiflexion at the ankle joint during late stance

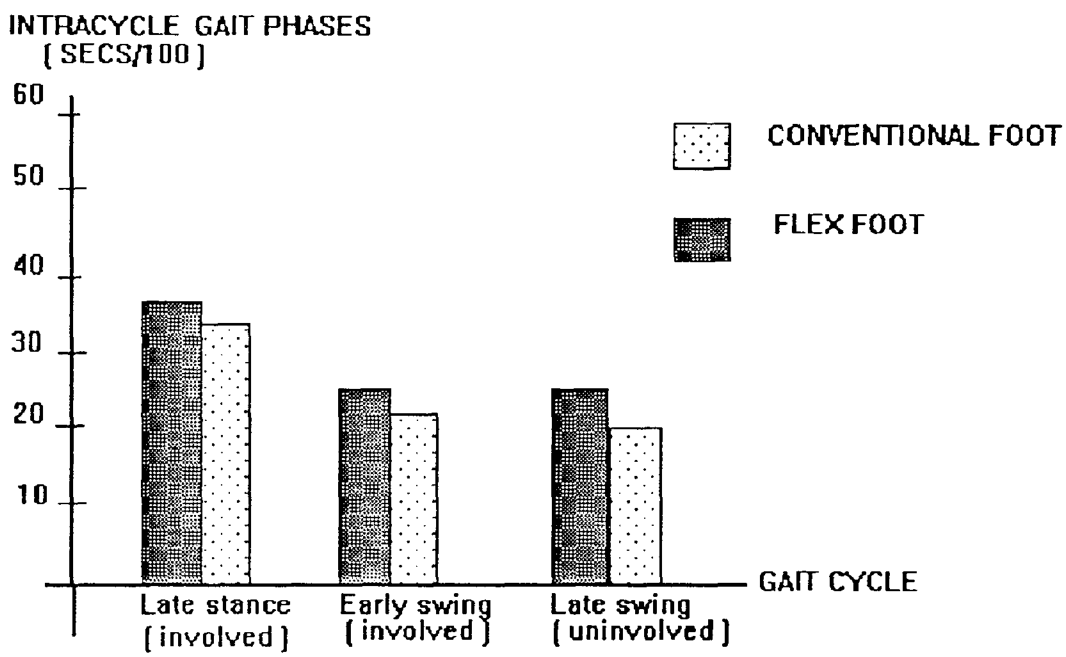


Figure 3.2 Single-Limb stance is longer for the Flex-Foot when compared to the conventional foot

Barr *et al.* (20) compared the gait of an amputee using SACH and Carbon Copy II prosthetic feet. They found no significant differences in stride characteristics, angular kinematics and moments of the SACH and Carbon Copy II prosthetic feet. However, the Carbon Copy II showed slower unloading in late stance and a later peak propulsive force than the SACH foot (Figure 3.3).

Angular acceleration occurring in the prostheses was the primary measurement in a study done by Wirta *et al.* (22). Angular acceleration can be used to study the capacity of prosthetic feet to absorb shocks. Effective shock absorption is necessary to avoid joint diseases in amputees.

This study showed that the shock absorbed by the Seattle foot was the greatest followed by the SAFE foot and then the SACH foot. These objective findings, when related to the subjective ratings, showed that amputees preferred prosthetic feet that developed lesser shock and greater damping immediately following heel-strike.

Barth *et al.* (14) have investigated several variables in an effort to compare energy-storing capabilities of SACH, SAFE, Seattle, Carbon Copy II and Flex foot. The SACH foot exhibited less ankle dorsiflexion in late stance compared to the flex foot (Figure 3.4).

This indicates that the SACH foot provides good late-stance stability which may be appropriate for lower activity level amputees. More ankle dorsiflexion and increased change in dorsiflexion in the Flex foot makes it appropriate for higher activity level amputees including amputees walking on inclines or uneven terrain.

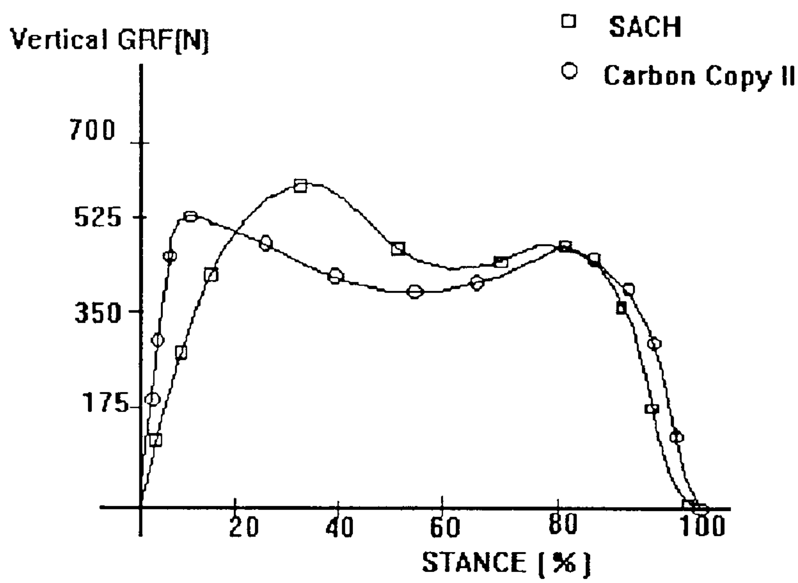


Figure 3.3 Carbon Copy II shows slower unloading in late stance and a later peak propulsive force than the SACH

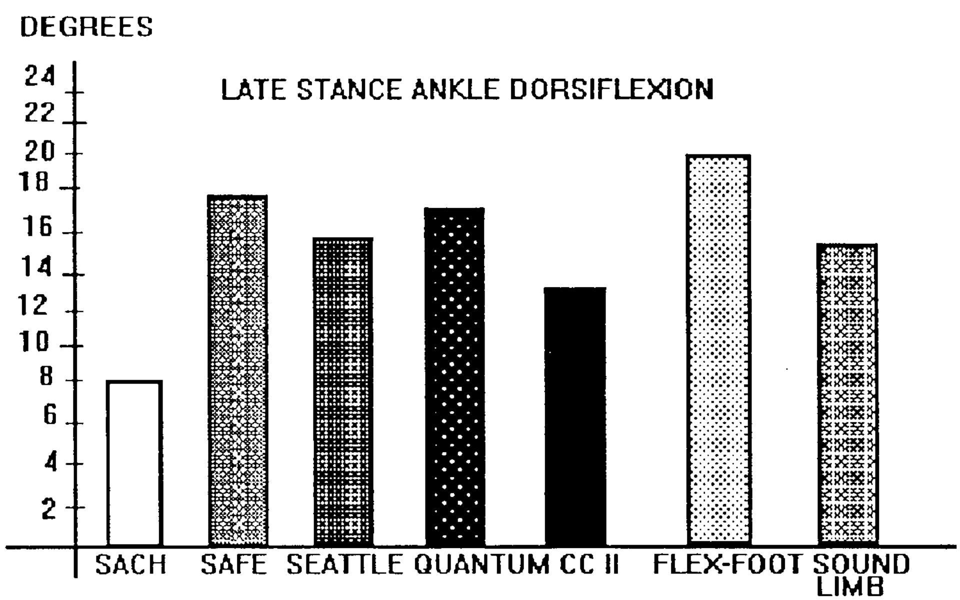


Figure 3.4 The SACH Foot exhibits less ankle dorsiflexion in late stance as compared to the Flex Foot

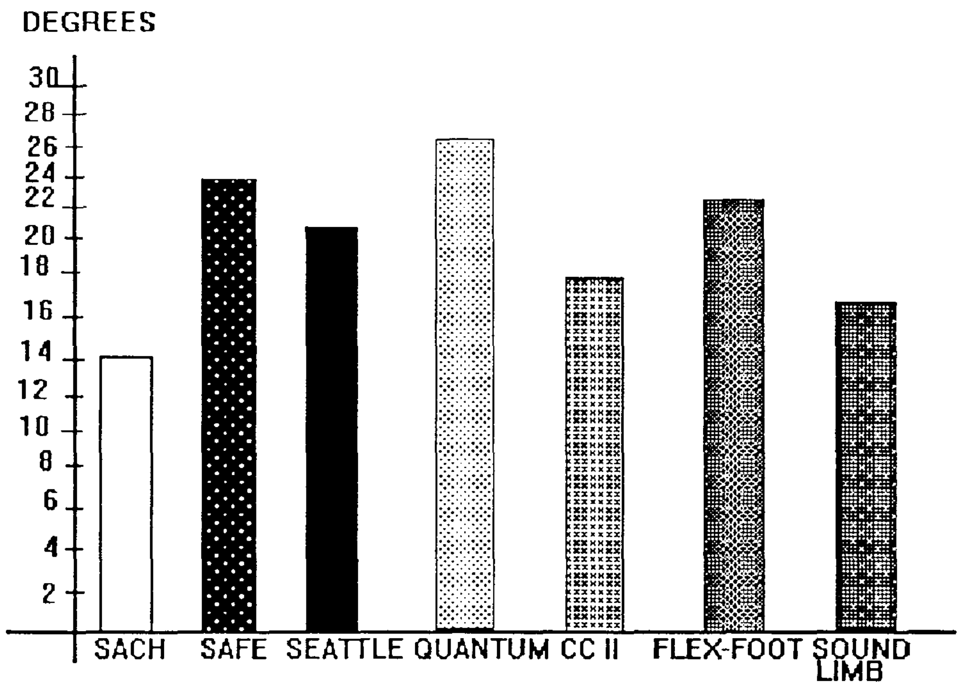


Figure 3.5 The SAFE Foot and the Flex Foot have an increased change in ankle dorsiflexion

The SAFE foot also has an increased change in dorsiflexion which makes it suitable for amputees walking on different surfaces and at different velocities (Figure 3.5). The Seattle foot exhibited the most symmetrical gait of all the prosthetic feet. This makes it a good choice for average activity level amputees. The Carbon Copy ll showed significantly greater sound-limb weight acceptance forces.

Another variable of interest in the study of ESPF is joint muscle power. Czerniecki *et al.* (23) compared the running performance of two non-disabled individuals with a subject wearing below-knee prostheses with Flex-Foot, SACH and Seattle foot assemblies. In normal subjects, early stance phase was characterized by energy absorption due to eccentric contractions of the knee extensors and ankle plantar flexors (24). Energy generation occurs during late stance by power output of the ankle plantar flexors and knee extensors. In the amputee, the greatest energy absorption during early stance and the maximum hip extensor output was observed at the hip.

The Flex foot performance of the amputee most closely approached normal values with regard to energy absorption & power output. The SACH foot produced the lowest power output and the Flex foot the highest. Thus ESPF result in a more normal pattern of power absorption and generation during the stance phase of running.

In a recent study done by Czerniecki *et al.* (25), mechanical power outputs of the lower extremity in 5 normal & 5 below-knee amputee subjects using the SACH, Seattle and Flex foot were studied at a walking speed of 1.5 m/s. While wearing the SACH foot, negligible energy generation occurred at the prosthetic foot during push-off. The ESPF demonstrated increased energy generation during push-off. However,no significant differences were found in the pattern or magnitude of power outputs of ESPF compared to the SACH.

Table 3.1 Work performed and energy stored and returned (in Joules) by the SACH and Carbon Copy II prosthetic feet

Foot	Negative Work (storage)	Positive Work (Return)	Net Work	Energy Returned(%) (Return/Storage)
SACH	-1.1565	0.3417	-0.8148	30
CC II	-2.3348	1.3323	-1.0024	57

Barr *et al*. (20) showed that the Carbon Copy II foot returned energy at 57% efficiency compared with only 30% efficiency for the SACH foot during level, non vigorous gait. The Carbon Copy II performed greater work in both the energy storage (CC II = 2.33J, SACH = 1.16J) and energy return(CC II = 1.33J, SACH = 0.34J) phases of stance (Table 3.1). During late stance, the peak power for the Carbon Copy II foot was 19 W as compared with 5 W for the SACH foot.

3.2.2 Energy-Cost measures

Energy expenditure reduction is of clinical importance in prosthetic selection and training. Some foot manufacturers have suggested that their prosthetic feet reduce the amputees energy cost. This energy cost can be evaluated by comparing different prosthetic feet. Researchers have found that each amputee walks at a self-selected optimal speed. By evaluating the amputee at self-selected speed for each foot, the energy cost for that foot can be determined. Earlier studies have shown that below-knee amputees have a higher energy cost than normal individuals.

Fisher and Gullickson (26) have written a review paper on the studies of the energy cost of ambulation of normal people and amputees. According to an average of the results available from literature, a normal person walks at about 83 m/min, with an energy expenditure of 0.063 Kcal/min/Kg. The average below-knee amputee walks 36% slower expending 2% more Kcal/min. This study indicates that an amputee consumes more energy than a non-amputee at comparable walking velocities.

For optimum gait efficiency, it is imperative that prosthetic feet keep energy expenditure to a minimum. Nielsen *et al*. (13) have carried out a study to investigate the differences in self-selected walking velocity, relative exercise intensity, oxygen consumption and gait efficiency in below-knee amputees while wearing the SACH foot and the Flex foot. Seven below knee amputees were selected and tested at different walking speeds on a treadmill using the Flex foot and the SACH foot on alternate days. Heart rate and oxygen uptake measurements were taken at the end of each testing stage. The heart rate measurements were used to calculate % MHR (relative work load of walking expressed as a % of age-predicted maximum heart rate) which was considered as the measure of relative exercise intensity.

$$\% \text{ MHR} = \frac{\text{Steady State Exercise Heart Rate}}{\text{Age-Predicted Maximum Heart Rate}} * 100$$

Relative exercise intensity was used to evaluate gait performance in amputees wearing the SACH and Flex foot. The reduced % MHR values for ambulation with the Flex foot, which occurred at all walking velocities, indicated decreased levels of stress (Figure 3.6).

Oxygen uptake for walking in below-knee amputees was higher than normal oxygen consumption resulting in increase in energy-costs compared to normal energy costs. Reduced oxygen consumption can be considered as a definition of the energy-conserving component of ESPF and can be measured in order to determine which energy storing foot produces the more optimal gait.

Lower energy cost values were obtained at higher walking velocities for Flex foot walking (17 ml/Kg.min at 3 mph) compared to SACH foot walking (19 ml/Kg.min at 3 mph) (Figure 3.7). Energy cost values at slower walking velocities was similar for both the Flex foot and SACH foot. Self-Selected walking velocities for both the feet were found to be below normal values. However self-selected walking velocity for Flex foot walking was higher compared to SACH foot walking.

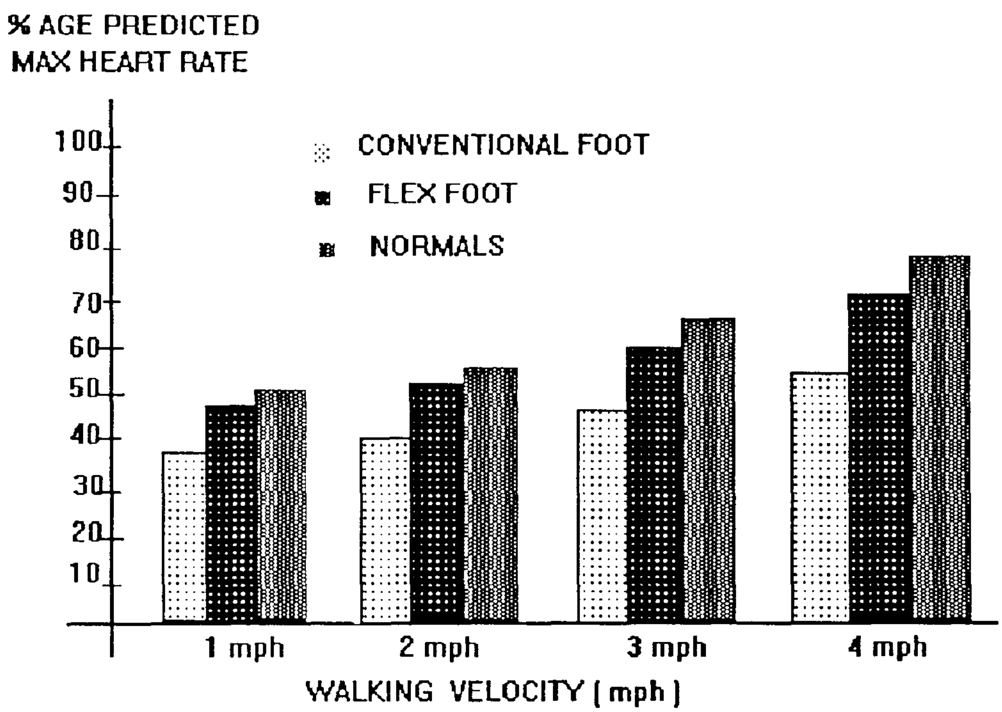


Figure 3.6 Relative exercise intensity at different walking velocities

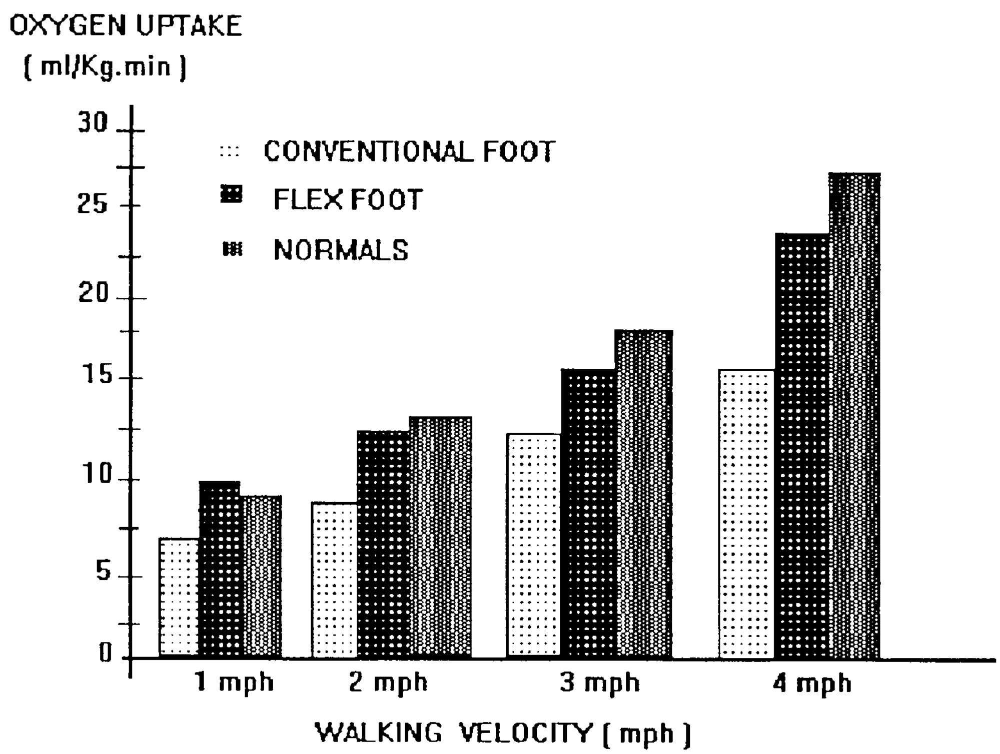


Figure 3.7 Energy cost of walking at different velocities

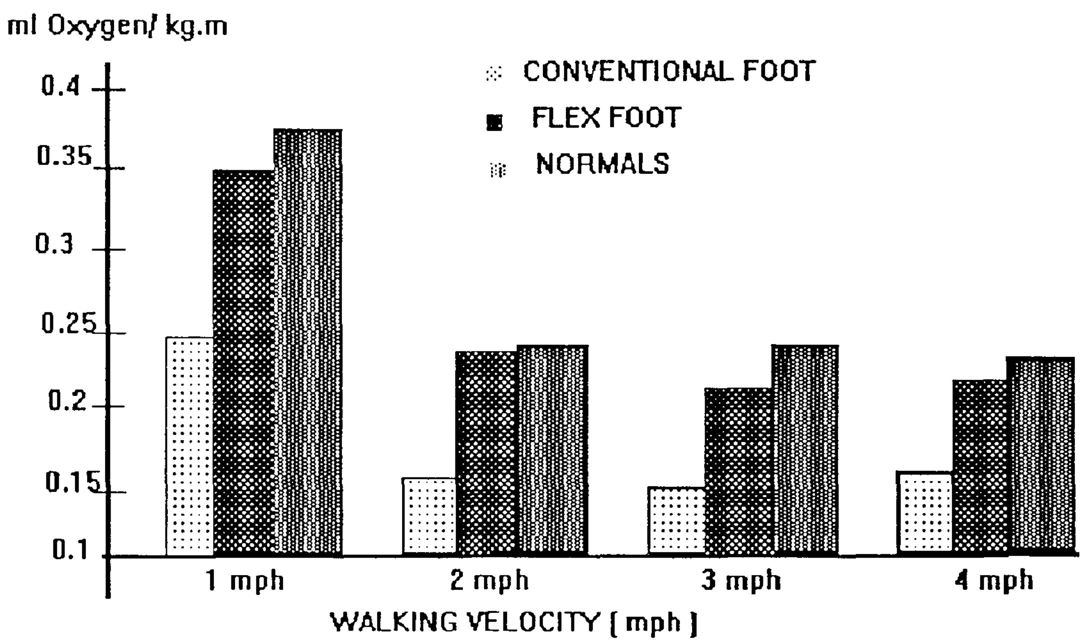


Figure 3.8 Gait efficiency at different walking velocities

An easily measured criterion of gait efficiency is the energy cost per distance travelled. It can be calculated as the ratio of oxygen uptake to walking velocity. Improved gait efficiency was observed for Flex foot walking at higher velocities (Figure 3.8). Optimal gait deficiency for walking with SACH foot was 0.24 ml oxygen/Kg.m at 3 mph and for the Flex foot, the optimal value was 0.21 ml oxygen/Kg.m at 3.2 mph. So we can say that for below-knee amputees, walking with the Flex foot conserves energy at higher walking velocities, resulting in lower levels of exercise intensity and enhanced gait efficiency.

In a study done by Meier *et al*. (27), SACH, Seattle and Flex foot were tested to evaluate energy expenditure. At a treadmill speed of 2 mph and 3.5 % grade oxygen consumption was 17.7 ml/Kg/min for the SACH foot, 19.10 ml/Kg/min for the Seattle foot and 15.89 ml/Kg/min for the Flex foot. These results indicate that the Flex foot may be the most energy efficient foot in treadmill ambulation. The results also suggest that the non energy-storing SACH foot may be more energy efficient than the energy-storing Seattle foot.

In the recent study done by Casillas *et al.* (28), a new energy-storing foot Proteor was tested in ten below-knee amputees and compared with SACH foot at various walking speeds. Oxygen uptake, heart rate and blood pressure were collected. While wearing the Proteor, the subjects had a higher self-selected walking velocity (5.19 Km/hr) than those with the SACH foot (4.95 Km/hr). The decrease in oxygen uptake in those with the energy storing foot compared to those with the SACH foot was not significant at the speed of 2 Km/hr. However this decrease in oxygen uptake at 4Km/hr is significant (14.91 ml/Kg/min with the energy storing foot compared to 19.65 ml/Kg/min with the SACH foot). During the inclined walking test, the reduction of energy cost with energy storing foot was also significant (18.22 ml/Kg/min compared to 22.59 ml/Kg/min with SACH). These results confirmed the benefits obtained from energy storing feet but could not explain the biomechanical reasons.

3.3.3 Subjective Preferences

Requirements for prosthetic feet vary because of different physical characteristics and activity objectives of patients. Subjective ratings by the amputees help in determining which physical characteristics and activity objectives influence the selection of a prosthetic foot. A measure of how easy or difficult the subjects find using the prostheses over a functional range of walking conditions is also an important measure and could be a factor to consider when prescribing a prosthetic foot.

In a study done by Wirta *et al.* (22), ratings were related to age, body weight, length of residual limb, etc. SACH, SAFE and the Seattle ankle-foot devices for below-knee prostheses were tested for effects on gait. The subjects were asked to rate each ankle-foot device in terms of excellent, superior, good, fair and poor. The SAFE and Seattle foot drew most of the favorable comments such as flexible, springy, comfortable, energy-saving, good on slopes and inclines. The common complaints with the SACH foot was that it was too stiff.

The following was inferred when the ratings were related to age, body weight and length of residual limb. The Seattle foot is best if the amputee is young, light-weight and has a medium length residual limb. The SAFE foot is best when amputee is middle-aged, slightly overweight and has a long residual limb. The SACH is good for an old, overweight amputee with a medium to long residual limb.

Amputee's assessment of walking difficulty was the focus of a study done by Macfarlane *et al.* (29). They investigated the effects of walking grade and walking speed on the subjects perception of walking difficulty with the Flex foot compared to the conventional foot. The subjects used in this study were healthy, active, below-knee amputees. As expected these subjects found walking with the Flex foot easier than walking with the conventional foot. The Flex-foot was associated with less walking difficulty compared to the conventional foot over nine different walking conditions.

In an earlier study done by Nielsen *et al.* (13), subjective feedback from the amputees using the Flex foot indicated that it improved their balance and stability on uneven ground but that the conventional foot might be better for slow and down-hill walking.

Similar results were obtained by Alaranta *et al.* (30). A comparison was made between the use of the Flex foot and the Carbon Copy ll on the basis of subjective ratings for ten items of movement. The below-knee amputees who participated in the study gave the Flex foot higher ratings in all the ten items. They concluded that moderately active persons benefit from an energy storing prosthetic foot system and that the Flex foot is particularly benefical for uphill and fast walking as shown in Table 3.2.

In a recent study, a comparative analysis of the Seattle foot and the Flex foot was done by Menard *et al.* (15). They found that both the Seattle foot and the Flex foot users walked well, were happy with their prostheses and reported improved functional capability. However many amputees found the excessive late "kick" in the Flex foot troublesome during routine activities. They preferred to use the Seattle foot for most activities and Flex foot for more demanding activities such as sports involving jumping.

Table 3.2 Subjective ratings of walking disability (0 = normal walking, 1= mild, 2 = moderate, 3 = severe disability) for the Flex Foot and conventional prosthesis for ten different kinds of walking

ITEM OF WALKING	PROSTHESIS	RATING OF DISABILITY 0	1	2	3
Indoors	FF	20	9	2	0
	CP	12	16	3	0
Upstairs	FF	14	14	3	0
	CP	3	13	15	0
Downstairs	FF	8	18	5	0
	CP	2	12	7	0
Even street	FF	21	9	1	0
	CP	9	17	4	1
Uneven street	FF	13	15	3	0
(sand,snow)	CP	8	1	11	1
Forest	FF	7	16	4	4
	CP	3	6	16	6
Street Uphill	FF	11	16	4	0
	CP	3	8	19	1
Street downhill	FF	10	18	3	0
	CP	3	13	14	1
Swift Walking	FF	13	12	5	1
	CP	5	7	15	4
Running	FF	3	8	9	11
	CP	0	0	11	20

CHAPTER 4
FINITE ELEMENT ANALYSIS

4.1 Introduction to Finite Element Analysis

The Finite Element Method is an advanced computer technique of structural stress analysis. It was introduced in the mid-sixties as a process of solving structural problems in mechanics (33). This method was soon recognized as a way to find approximate solutions to all the physical problems that can be modeled by differential equations. These equations can be analyzed to determine the performance (e.g. deformation or stress) of the model.

When a structure is loaded, stresses are generated in its materials. The magnitude and distribution of these stresses depend on three major factors : the geometry of the structure, the material properties of the structure, and the loading conditions on the structure. Also, the stresses depend on the boundary and interface conditions.

Theoretically, stress distribution is determined by using a mathematical model. Such a model represents the real structure to a certain extent and the structural factors (loading, geometry, material properties, boundary and interface conditions) are described mathematically. These mathematical descriptions are usually based on experimental data. The structural descriptions are combined with mathematical equations of the model and the equations are solved to determine the stresses.

Various theories and solution methods are available for certain classes of structures. The FEM as a computer method of structural stress analysis, however, is suitable in principle for any structure. The powerful Finite Element Method is capable of evaluating stresses in structures of complex shape, loading and material behavior.

The first step in using this method is to geometrically define the model or structure. The model is then (mathematically) divided into a number of elements, connected at corner locations called nodal points or nodes, thereby forming a finite element mesh. The boundary and loading conditions are numerically defined as displacements and forces, respectively in the boundary nodes. Each element is assigned one or more material properties (e.g. Modulus of Elasticity).

The FEM computer program calculates the stiffness characteristics of each element and assembles the element mesh through forces and displacements in each node. Since the entire structure is divided into smaller elements, the FEM program solves a large number of equations that govern force equilibrium at element nodes.

The solution obtained with the Finite Element Method is approximate in the sense that it converges to the exact solution for the model when the mesh density approximates infinity. Thus, the accuracy of a Finite Element Method depends on the number of elements in the mesh. The larger the number of elements, the more accurate the solution will be.

A variety of element types are usually available for 3-D and 2-D structures in an FEM computer package, and they differ in their shapes and number of nodal points. The computer time required for 3-D elements is higher than the time required for 2-D elements. Mesh accuracy is easy to obtain in a 2-D model, because of cost efficiency.

While interpreting the FEM results, it is very important to differentiate between the validity of the model and the accuracy of the model. Validity of the model is the precision by which the mathematical descriptions of the structural factors (loading, geometry, material properties, boundary and interface conditions) mimic the actual structure. Accuracy of the model is the precision by which the FEM mesh approximates the exact solution for the model. Accuracy can be checked with a convergency test while validity can be determined by experimental verification.

4.2 The Finite Element Method in Orthopedic Biomechanics

The Finite Element Method was first introduced to orthopedic biomechanics in 1972 to evaluate stresses in human bones. The traditional mathematical tools available for stress analysis were not very suitable for the irregular structural properties of bone. The logical solution for this problem was the FEM, which can be used to evaluate stresses in structures of complex shape.

Since then, there has been a rapidly growing interest in artificial joint replacement and new methods of fracture fixation. New questions and new methods have created an environment for the use of the FEM in orthopedic biomechanics. Finite element analysis is by far the most exciting and promising of the structural analytic techniques that have been applied to biomechanics.

Finite element analysis is presently being used for stress analysis of bones and bone-prosthesis structures, artificial joint designs, fracture fixation devices and tissues such as articular cartilage and intervertebral discs. The aim of the stress analysis is to determine the mechanical behavior of the tissues and to test and optimize artificial joint designs and fracture fixation devices.

The Finite Element Method is now well established as a tool for basic research and design analysis in orthopedic biomechanics. However, significant findings and useful concepts generated by this method are limited. One reason for this is the complexity of biological structures. The true behavior of biological materials (as opposed to engineering materials) has not yet been fully described in terms of mechanical behavior. Another limitation is the lack of a clear understanding of the nature of the clinical problem. The model should be designed in such a way that it fits the objectives of the finite element analysis. The validity of the model should be determined by methods such as experimental verification.

There are three possible approaches that can be followed in the solution of an orthopedics-related problem (33). The approaches and their advantages and disadvantages are listed in Table 4.1. The coordinated approach is the best approach in order to overcome some of the uncertainties inherent in biological material properties. Figure 4.1 illustrates the interactions between the finite element analysis, laboratory experiments and clinical evaluations in the coordinated approach. Such an approach provides a framework for an efficient method for using finite element analysis in orthopedic biomechanics.

In this thesis, a finite element stress analysis model was developed in order to provide insight into the mechanical behavior of a below-knee prosthesis. A structural analysis using the finite element method was performed in order to identify the relative importance of geometry, material properties and loading conditions in the design of this prosthesis.

The finite element method has two major advantages over other techniques in the design of a prosthesis : 1) This method allows us to represent various geometric and material properties in one particular design, which is difficult to do in most other methods. 2) It is possible to analyze multiple loading conditions with a single finite element model. 3) A particular design of a prosthesis can be tested to failure only once if we use experimental techniques. However, a finite element model can be analyzed

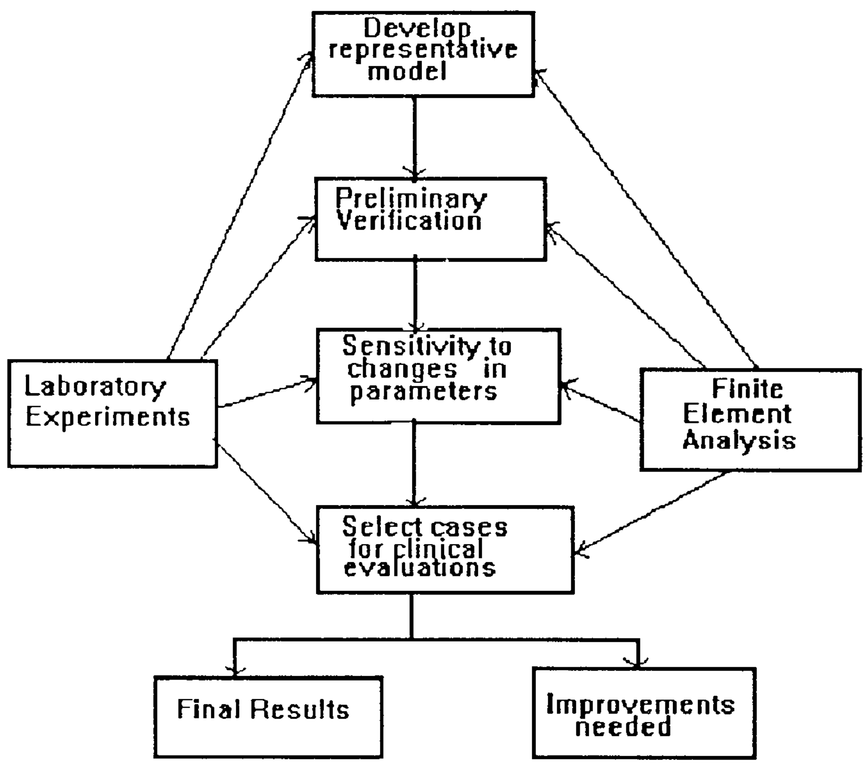


Figure 4.1 Interactions between finite element analysis, experiments and clinical evaluations in orthopedics problems

Table 4.1 Three approaches to orthopedics problems and their advantages and disadvantages

Approach	Advantages	Disadvantages
Computational model	▪ 'Low' costs ▪ 'Short' time	▪ Need to verify model with measurement
Experimental model	▪ Measurement for stresses rather than computations	▪ Time and cost required ▪ Difficult to study a range of parameters ▪ Difficult to extrapolate to new conditions
Coordinated computational and experimental	▪ Have measurements and a degree of confidence in computations ▪ Reduced time and costs ▪ Can extrapolate to other conditions	▪ Requires more thought in planning and coordinating programs

Presently the capabilities of the Finite Element Method have increased due to new developments in engineering mechanics. This, along with sophisticated computers, offer exciting possibilities for the future. Progress in this area of biomechanics, however, requires a sound understanding of engineering mechanics on the one hand, and an appreciation of biological complexities on the other.

CHAPTER 5
STRUCTURAL OPTIMIZATION

5.1 Definition of Optimization

The term optimization is defined to be the process that a designer uses to achieve an improved solution (32). Although it is desirable to have the best or optimum solution to a problem, the designer usually has to settle for improvement rather than perfection in design. Optimization provides a logical method for the selection of the best choice from among all the possible designs that are available.

5.2 The Optimization Problem

The mathematical formulation of the optimization problem is as follows :

$$\begin{aligned} &\text{Minimize } F(x_i), && i = 1, \ldots\ldots n \\ &\text{Subject to } \; c_j(x_i) = 0, && j = 1, \ldots\ldots p \\ &\qquad r_k(x_i) \leq R_k, && k = 1, \ldots\ldots q \\ &\qquad x^l_i \leq x_i \leq x^u_i, && i = 1, \ldots\ldots n \end{aligned}$$

where F : Objective function to be minimized

n : Number of design variables x_i

Cj : Functional constraints specified by the designer

p : Number of functional constraints

r_k : Regional constraints (Inequality constraints)

q : Number of regional constraints

x_i : Design variables

x^l_i : Lower limit for the design variables

x^u_i : Upper limit for the design variables

The objective function or merit function is an equation or expression that has to be maximized or minimized. It provides a way to evaluate and compare two different designs. Mathematically, the objective function defines an (n+1) - dimensional surface. The value of this function will depend on the values of the design variables. Functional constraints are functional relationships of the design variables that must be satisfied in the design solution. Examples of common objective functions to be maximized or minimized are cost, weight, strength, size and efficiency.

If there is only one design variable, the objective function can be plotted as shown in Figure 5.1 (32). If there are two design variables, the objective function can be plotted as a 3-D surface, as shown in Figure 5.2 (33). The physical and mathematical characteristics of the objective function are of great importance in the optimization process. Some types of optimization problems can be formulated in terms of more than one measure of merit. For example, it may be desired to maximize strength, minimize weight, and minimize cost. In such a case, the designer must establish priorities and assign weighting values to each measure of merit. This process results in a trade-off function and provides a single composite merit value to be used in the optimization process.

The optimization problem exhibits some characteristics that make it complicated when practical design applications are considered. The main difficulties arise from the size and the number of design variables. The number of design variables should be reduced by means of an equality constraint. Whenever possible, the optimization problem should be scaled so that the design variables are of the same relative magnitude. The form of the objective function should be as simple as possible. The process of

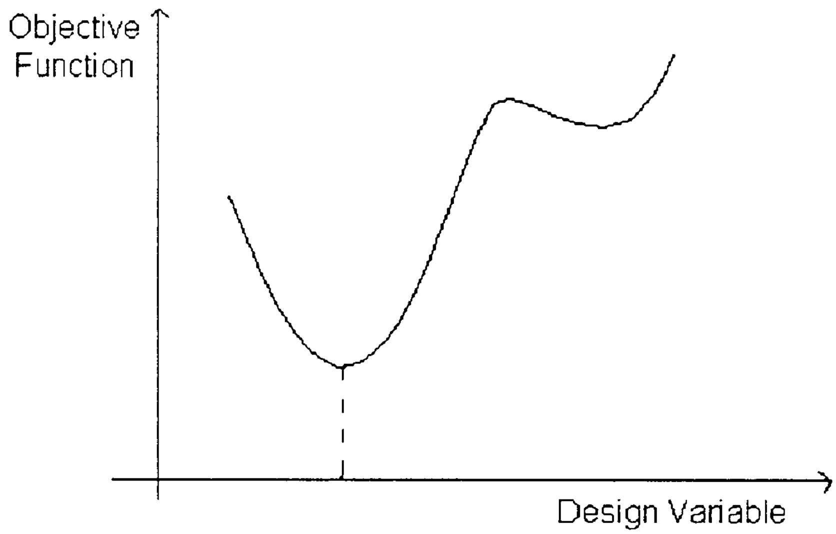


Figure 5.1 A one-dimensional objective function

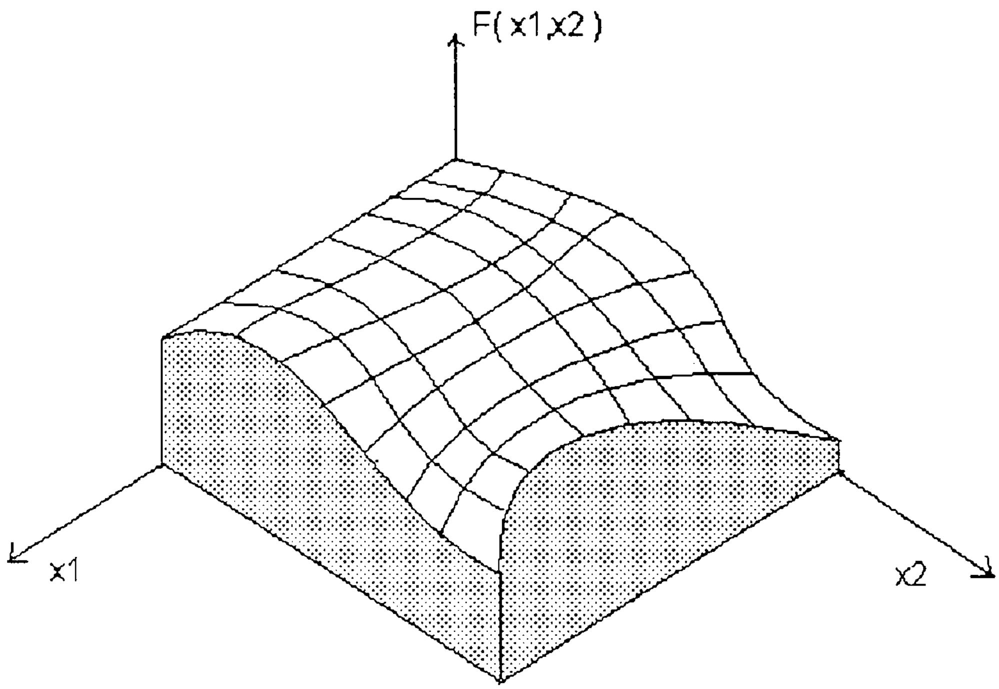


Figure 5.2 A two-dimensional objective function

optimization is an iterative one and each iteration consists of a complete finite element analysis. The above mentioned difficulties, however, may increase the cost of obtaining an optimal design of a structure.

5.3 Solution of the Optimization Problem

There are a number of numerical optimization techniques available for the solution of an optimization problem. Most of these problems utilize iterative and numerical methods of solution. The numerical methods, frequently used for the solution of optimization problems, are termed as optimality criterion methods. In these methods, the optimality conditions are contained in discrete sets of governing equations.

The selection of an optimization technique depends on the number of design variables and on the nature of the design space. For complex shapes, the state equations, constraint equations and the optimization conditions form a large set of equations. In such a case, it is better to use a purely numerical optimization technique.

Optimization methods in multidimensional space are classified in two broad categories :

5.3.1 Direct methods

Direct methods utilize a comparison of functional evaluation. These methods try to use a strategy that approaches the optimum value. Some direct methods are the Fletcher-Reeves method, Hooke and Jeeves method and Simplex method.

5.3.2 Indirect methods

Indirect methods use the mathematical principles of maximization and minimization. These methods try to satisfy the conditions of the problems without examining non-optimal points. Some indirect methods are the sectioning method, the area elimination method and the random method.

5.4 Structural Optimization

The aim of structural optimization is to improve the performance of the structure. It is very difficult to achieve this aim manually by making a number of design changes in an attempt to improve several aspects of a structure's performance. The behavior of the structure is highly dependent on local design changes. Hence, it is important to limit the number of design changes or to reject some combinations of design changes. This often results in a great deal of time being spent on achieving a relatively small improvement in design.

5.4.1 The Structural Optimization Problem

The process of optimization implies producing the best design for a structure under the defined loading conditions (37). The relative merit of alternative designs is generally evaluated with reference to :

1. Satisfactory performance - The designer may specify upper and lower limits on the structural response. In optimization, for linear static structural response, there are displacement (stiffness) constraints and stress (strength) constraints.
2. Structural mass - Of all the possible designs satisfying the performance requirements, the one with the minimum mass is defined to be the best. Thus optimization is designed to minimize mass subject to the specified

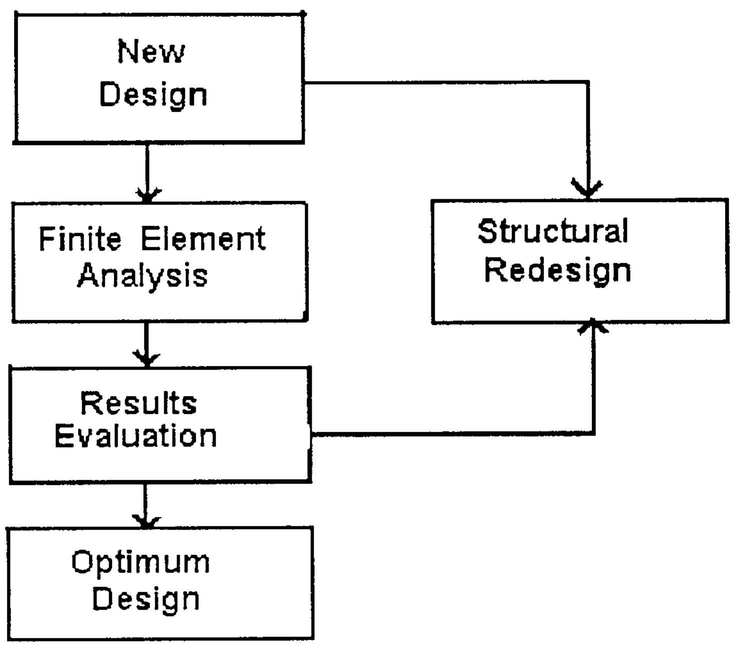


Figure 5.3 The design process using structural optimization

performance constraints. In optimization terminology, mass becomes the objective function. Other objective functions can be used but structural mass is generally used because minimizing mass results in the lightest and often the cheapest structure.

3. Analysis variables - Analysis variables are structural parameters that are fixed at the beginning of optimization. In optimization, material properties and outline geometry are considered fixed. In addition, some elements within the model of the structure may be assigned to be frozen and may not be allowed to change. Analysis variables are crucial to the function or fabrication of the structure and therefore cannot be changed during the process.

4. Design variables - Design variables are parameters, defining a structural system, that are allowed to vary. For example, in a finite element model composed of shell elements, the shell thickness is a design variable that is allowed to vary.

5. Gauge constraints - In addition to constraints on structural response, constraints may also be specified for design variables.

The upper and lower bounds on the design variables are called gauge constraints. These constraints may be due to the material sizes available, manufacturability, etc.

Optimization can make multiple simultaneous structural modifications to minimize the mass subject to a given set of stress, displacement, and gauge constraints. With only a few iterations, the designer can gain insight into alternative design studies and load paths. Optimization takes finite element analysis a step further by evaluating many design alternatives and selecting the best one. Structural optimization helps to minimize the number of iterations in the design (Figure 5.3).

5.4.2 Solution of the Structural Optimization Problem

A number of strategies have been adopted to obtain optimum solutions to structural problems. Iterative algorithms such as stress redesign and displacement /stress redesign are frequently used. In both the methods, the user specified constraints are evaluated. This supplies enough information for the stress redesign algorithm to resize the structure. However, the displacement/stress redesign algorithm requires further information that should be calculated for the resizing.

Convergence testing is used in both the algorithms to decide whether the optimum solution has been obtained. If not, the elements in the structure to be optimized are resized further and another finite element analysis is performed.

In this thesis, the design of a below-knee prosthesis was optimized for reduced weight. Such an optimization may result in improvements in amputee gait without a significant increase in amputee effort. Optimization was repeated while varying the material properties to determine if the weight of the prosthesis could be reduced by the use of different materials, without violating any of the functional constraints. The dimensions of the prosthesis were also optimized for reduced weight and therefore improved function.

Interest in Structural Optimization has increased greatly during the last decade because of the availability of reliable numerical analysis methods and the computer power necessary to use them efficiently. Therefore a number of commercial optimization systems, based on finite element analysis, such as ANSYS, I-DEAS, CAOS, etc. have been introduced. Most of these systems can be integrated into a computer-aided design environment to provide a valuable tool to the designer in the design, analysis and optimization process.

CHAPTER 6
COMPUTER AIDED ENGINEERING PACKAGE I-DEAS

6.1 Introduction to I-DEAS

I-DEAS (Integrated Design Engineering Analysis Software) is a comprehensive and integrated package of mechanical engineering software tools. The purpose of this software is to provide a concurrent engineering approach to product design and analysis. I-DEAS is made up of a number of " Families " of software modules. The main families are :

1. Solid Modeling
2. Finite Element Modeling and Analysis
3. System Dynamics
4. Test Data Analysis
5. Drafting
6. Manufacturing

I-DEAS integrates these families into one package with a common user interface and a shared application database. Each family is further subdivided into "Tasks" and each task has its own subdivision. All the tasks are executed from a common menu and share a common database. I-DEAS is a highly interactive, graphics-oriented and menu-driven package. Here we will restrict ourselves to the study of Solid Modeling and Finite Element Analysis families because only these two families are utilized in the design and analysis of our prosthesis model.

6.2 Solid Modeling

I-DEAS Solid Modeling family offers a number of software tools to develop and design geometry for any mechanical system. The tasks that were used in our design are as follows :

6.2.1 Object Modeling Task

The Object Modeling task within the Solid Modeling family can be used to create solid objects in the following three ways:

1. Primitives - geometry from familiar solid shapes.
2. Profiles - extruding or revolving a 2-D outline.
3. Skinning - using a set of profile cross-sections to define a solid.

After solid models are created, they can be modified by various construction operations such as cutting, joining, and intersecting with different objects.

6.2.2 Construction Geometry Task

The Construction Geometry Task can be used to create three kinds of geometry : wireframe, profiles, and skin groups. Solids can then be created from this geometry in the Object Modeling Task. A wireframe is a collection of points and curves in three dimensional space. It can be used to create profiles and skin groups for more complicated solid geometry. Profiles are two dimensional points and curves connected to define 2-D outline. They can be used to create solid objects by extruding or revolving one profile, or by skinning between multiple profiles. Skin groups are collections of profiles representing cross-sections through a part, or collections of wireframe curves defining a surface.

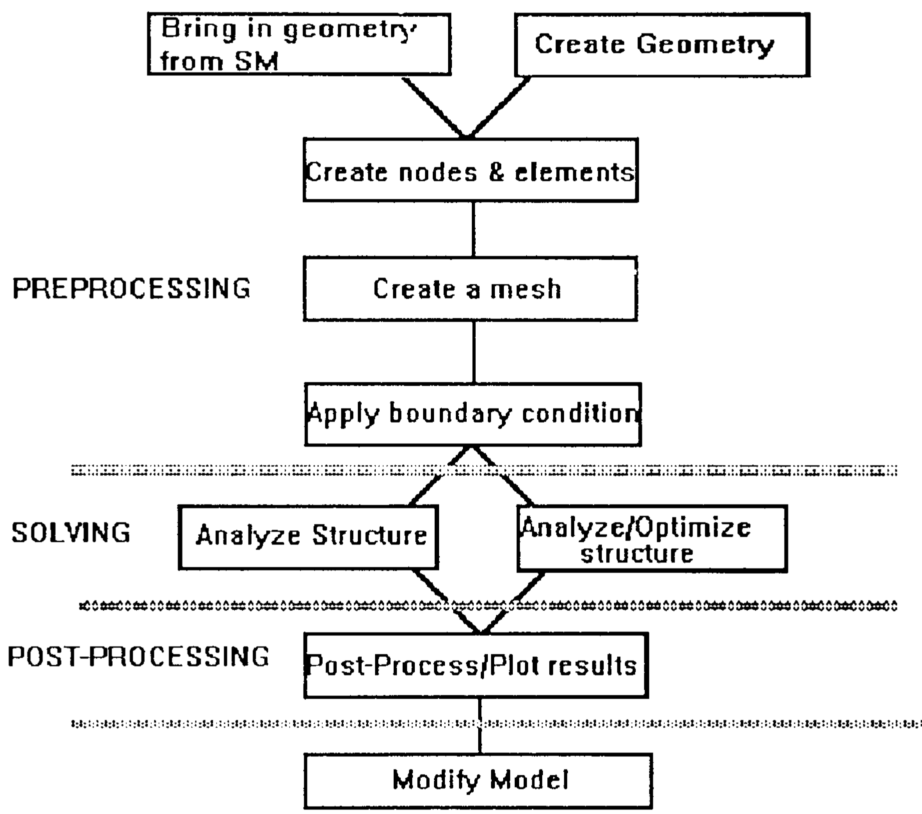


Figure 6.1 The steps of finite element modeling and analysis

6.3 Finite Element Modeling and Analysis

The Finite Element Modeling and Analysis family is used to calculate deflections and stresses due to loads on the model. The model is subdivided into a mesh of elements which are used to calculate the stiffness of the structure and solve for the deflections given the loads and boundary conditions. From the deflections, the stresses in each element can be calculated.

A Finite Element Model is the complete idealization of the entire structural problem, including the node locations, elements, physical and material properties, loads and boundary conditions. The FEM is constructed to mathematically model the deflection of the structure, not to look like it. The accuracy of the solution depends on how well the structure was modeled, the assumptions made for loads and boundary conditions, and the accuracy of the elements used for the given problem.

Finite Element Modeling and Analysis consists of the following steps : Pre-processing, Solution/Optimization and Post-processing (Figure 6.1). The basic tasks used for these steps of Finite Element Modeling and Analysis are :

1. Pre-processing
    - Geometry Modeling Task
    - Mesh Creation Task
    - Boundary Conditions Task
2. Solution
    - Model Solution Task
3. Optimization
    - Optimization Task
4. Post-processing
    - Post-processing Task

A Finite Element Model is the complete idealization of the entire structural problem, including the node locations, elements, physical and material properties, loads and boundary conditions. The FEM is constructed to mathematically model the deflection of the structure, not to look like it.

The accuracy of the solution depends on how well the structure was modeled, the assumptions made for loads and boundary conditions, and the accuracy of the elements used for the given problem.

6.3.1 Pre-processing

Pre-processing includes developing the geometry of a finite element model, entering physical and material properties, describing the boundary conditions and loads, and checking the model. The tasks used in pre-processing are as follows :

a) Geometry Modeling Task : The Geometry Modeling Task may be used to create or modify the geometry of the finite element model, which will then be used for defining mesh-areas. This geometry can either be created in the Geometry Modeling task or it can be transferred from an object created in Object Modeling. Surfaces for finite element modeling are also created and modified in this task.

b) Mesh Creation Task : The Mesh Creation Task is used to define mesh areas and volumes, to generate nodes and elements on mesh areas, to enter the material and physical properties and to check the model for errors.

(i) Mesh Area Definition - The geometry of the finite element model created in the Geometry Modeling task is used to define mesh areas. Mesh areas are used for generating nodes and elements and for defining mesh volumes. Mesh areas can be either mapped or free-mesh. Mapped meshing requires the same number of elements on opposite sides of the mesh area and requires that mesh areas be defined by at least three edges. Free or automatic meshing allows much more flexibility in defining mesh areas. An algorithm available in I-DEAS uses the element size in order to automatically create the free mesh. Mesh areas can be created by using the Auto_Create command.

(ii) Mesh Volume Definition - Mesh volumes are defined by closed regions bounded by mesh areas. They can be either mapped or free meshed. Mesh volumes are defined for a 3-D finite element model.

(iii) Nodes and Elements - " Nodes " are coordinate points in 3-D space, where " Elements " will be connected, loads applied, boundary restraints imposed, and displacement information determined. Nodes and elements can be created by mesh generation. Nodes can also be created manually by keying in their coordinates and elements can be created manually from two or more nodes. Nodes and elements can be modified after they have been generated.

(iv) Mesh Generation - Nodes and elements are generated on mesh areas and mesh volumes by using the Generate command. They can also be generated directly on an object that was created in Object Modeling and transferred to Finite Element Modeling. In such a case, we do not have to define mesh areas and volumes and determine the mesh size in order to generate the mesh on the solid object.

(v) Material and Physical Properties - Material and physical properties are entered in the Mesh Creation task. Each element in a finite element mesh is required to reference a material property table and a physical property table. Physical property tables are different for different types of elements, and contain data that define some physical aspect of the elements. Shell thicknesses, for example, are stored in physical property tables.

A material property table contains the data used to characterize the mechanical behavior of materials. Three types of material tables (isotropic, orthotropic and anisotropic) are supported by Model Solution Linear Statics analysis. The structural behavior of each of these types of materials is characterized by constants such as Elastic Modulus, Poisson's ratio, Shear Modulus, etc., which are specified in the material property tables. In order to represent the model correctly, it is important that the property tables are understood and that the entries are correct.

(vi) Model Checking - Model checking helps in identifying modeling errors in the finite element model. Some of the typical errors are duplicate nodes, duplicate or missing elements, and highly distorted elements.

c) Boundary Conditions Task : The Boundary conditions task in I-DEAS is used to define the loads and other boundary conditions that have to be applied to the model. The boundary conditions are collected into five generalized menus and the individual boundary conditions are found under these menus. These boundary conditions are :

(i) DOF Sets

(ii) Constraints

(iii) Restraints

(iv) Structural loads

(v) Heat Transfer loads

Case sets are used to collect different kinds of boundary conditions that belong together for an analysis. The purpose here is twofold : To bring together loads, restraints, kinematics and constraints that are pertinent to an analysis task. And to be able to do the analysis over again with a different set of boundary conditions by defining multiple case sets. The boundary conditions for most structural problems are restricted to loads and restraints.

6.3.2 Solution

The finite element model can be solved in the Model Solution task of I-DEAS Finite Element Modeling and Analysis. I-DEAS Model Solution can solve for linear statics, linear dynamics, conduction heat transfer, and potential flow analysis on models created in the Pre-processing module.

a) Model Solution Task : The steps to be performed in the Model Solution task are as follows :

(i) Select the solution type such as linear statics, linear dynamics, etc.

(ii) Select the execution options such as batch mode, interactive mode, etc.

(iii) Select the case set to use for the analysis.

(iv) Select the method of solution such as Verification_Only, or Solution_No_Restart.

(v) Select the type of outputs such as stresses, displacements, strain energies, etc.

(vi) Solve.

6.3.3 Optimization

Optimization automatically optimizes the physical properties of models created in Pre-processing. This optimization process is based on defined loads, and considers stresses, displacements and restraints that have been defined. For example, Optimization can be used to run several iterations to determine the optimum thickness parameters to minimize weight.

a) Optimization Task : This task puts Model Solution in a loop to run iterations and to optimize the model by using the previous results. The following steps are followed in the Optimization task :

(i) Create a design - A design is created under the Manage_Designs menu. This design is created from the finite element model used for the solution.

(ii) Setup Optimization - This is done under the Setup_Optimization menu. The following selections are available under this menu :

1. Node groups : If shape optimization is to be performed, node groups are selected. The shape of the object can be optimized by changing the location of the nodes, and the nodes that have to be moved are selected into groups.
2. Element groups : The elements that have to be modified are selected into an element group. Element groups are created if cross-section, physical, or material properties have to be modified.
3. Optimization variables : The variables that have to be modified are defined, for example, thickness of a group of shell elements, node locations, etc.
4. Optimization constraints : The constraints on the optimization problem are defined or the limits on certain variables are defined. For example, maximum deflection allowed for a particular node, maximum allowable stress, maximum allowable mass, etc.

(iii) Solution Control - This is done under the Control_Solution menu. First the method to be used is selected such as linear statics, linear dynamics, etc. Then the number of iterations is selected. Then the type of output is selected such as stresses, displacements, etc. And finally, the execution options are selected such as interactive or batch.

(iv) Solve - The optimization problem is given to the optimization solver to be solved.

(v) Display the results - Once the solution is finished, the history of mass, constraint values, or optimization variables as a function of iteration number can be plotted. The results from the optimization including the mass and stress history can be displayed and monitored.

(vi) Update finite element model - After displaying and analyzing the optimized model, it is returned to the original finite element model so as to update the old finite element model.

6.3.4 Post-processing

Post-processing involves plotting deflections and stresses, and comparing these results with the failure criteria imposed on the design such as maximum deflection allowed, material static and fatigue strengths, etc. Post-processing also includes checking for errors that were not detected while building the model.

Once the solution is finished, the results are displayed, interpreted and processed in Post-processing. First the analysis dataset that has to be displayed is selected. Then the display type is selected such as deformed geometry or contour. For stress data, the data component to be displayed is selected. Then the display option is selected such as continuous tone, free face, etc. for stress data; hidden line, shaded image, etc. for deflection data.

CHAPTER 7
STRUCTURAL OPTIMIZATION OF A PROSTHESIS USING I-DEAS

Since an actual prosthesis was unavailable, the dimensions of the below-knee prosthesis were obtained by means of geometrical approximation of the shank and foot. The two modules of I-DEAS used in this study were Solid Modeling and Finite Element Modeling and Analysis. The steps followed in the finite element analysis of this model are explained as follows :

7.1 Solid Modeling

The solid model of the prosthesis was created in the Object_Modeling task of the I-DEAS Solid Modeling family. Initially the geometry was created using profiles in the Construction_Geometry task. Profile points for seven sections were created by using the obtained measurements. Splines were fit through the points of each section to create seven profiles. The path method was used to create a skin group from these profiles. This method uses another profile to define a path and the section profiles are "hung" on it. A path profile was therefore created to provide a path for skinning betweeen the seven profiles. The seven profile cross-sections and the path profile used are shown in Appendix A.

The profiles were positioned at predefined points on the path and a skin group was created through the profiles. After the skin group was created, it was used in the Object_Modeling task to create a closed solid object as shown in Appendix B.1. The Skin command in the Create menu of Object_Modeling was used to create this solid model of the prosthesis. Since this model is symmetric about the XY plane (i.e. at $Z = 0$), the loading on it will also be symmetric. In such a case, it is possible to model only half of the problem. Therefore, the solid model is cut with the XY plane using the Cut command in the Construction menu. The part of the object on the positive side of the plane was saved into a file.

7.2 Finite Element Modeling and Analysis

The solid model created in Object_Modeling was transferred to FE_Modeling_&_Analysis to create a finite element model of the prosthesis. A mesh was generated on the object using the Generate_Mesh command in the Mesh_Creation task. The global element length was selected as 0.36cm. The model contained 535 nodes and 235 elements. The finite element model of the prosthesis is shown in Appendix B.2.

7.2.1 Physical properties

Thin triangular and quadrilateral shell elements were used to model the prosthetic structure. Thin shell elements can be effectively used for structures with relatively thin walls such as molded plastic where bending and in-plain forces are important. Using thin shell elements assumes that the stress can only vary linearly through the thickness, which is a limitation. However, a model made up of thin shell elements takes much less time to solve. The thickness of the thin shell elements was assumed to be 0.5 cm which was entered as a physical property in the Mesh_Creation task.

Table 7.1 Material properties of the below-knee prosthesis (Reference 35)

Material	Elastic Modulus (MPa)	Specific Gravity	Poisson's Ratio
Fiberglass reinforced polyster resin	1.4 * 10000	1.9	0.13
Fiberglass reinforced epoxy	3.9 * 10000	1.84	0.13
Polystyrene	1.4 * 1000	1.06	0.13

7.2.2 Material properties

The elements were assumed to have isotropic material properties. It was decided to utilize qualities of certain fiber reinforced composites to provide some form of assistance in initiating heel rise at toe-off and in moving the prosthesis forward. Some of these composites can be designed thin enough to flex and thick enough to maintain structural rigidity. The advantages of these composites include increased flexibility and stress absorption an reduced weight.

Table 7.1 lists the material properties that were used in the analysis. Material properties include Youngs Modulus of elasticity, Poisson's ratio, yield strength, etc. Each of these sets of material properties were applied to the finite element model in order to determine the importance of different material properties in the design of the prosthesis.

7.2.3 Boundary Conditions

The Boundary_Conditions task was used to build analysis case sets containing loads and restraint boundary conditions to be applied to the model. Most structural problems only need structural loads and restraints. In this case, two different loading conditions were analyzed.

A resultant force of 1000N (for a person weighing 70 kg) was used as one of the loading conditions. This force was applied at a nodes located in the heel to simulate heel strike. The second loading condition was used to simulate the stance phase of the gait cycle. This load case consisted of a distributed load (1000N) applied to the knee and directed parallel to the axis of the shank.

Since only half of the model was being analyzed, symmetric boundary conditions were created on the XY plane by using the Restraints command in the Boundary_Conditions task (Appendix B.3). To balance the applied load for the stance phase, all the nodes at the base of the foot were restrained. To balance the force during heel strike, the nodes at the knee were rigidly restrained. Analysis case sets were created for the two loading conditions.

7.2.4 Model Solution

Both the case sets that were created in the Boundary_Conditions task were used in the Model_Solution task to obtain the solution. The linear statics solution type and the interactive execution option were selected. Analysis data sets were formed which included displacement and stress output datasets. Finally, the finite element model was solved.

7.2.5 Post Processing

In this task the results obtained after the solution were displayed and interpreted. First the stored Analysis datasets were selected. The displacements data sets were used to determine the deflection of the prosthesis as the result of the applied load. The stress datasets were used to display maximum principal stresses, normal stresses, shear stresses and Von Mises stresses. Von Mises stress is generally considered as the yield indicator for most materials. It was therefore used to evaluate the general distribution of the stresses on the prosthesis. The contour menu was selected to display the stresses in various formats. The deformed geometry menu was selected to display the deflection of the prosthesis. Various display options such as continous tone and free face for stress data and hidden line for deflection data were used.

CHAPTER 8
RESULTS

Maximum principal stresses and Von Mises stresses were obtained from finite element analysis of the model of the prosthesis for the loading condition of heel-strike. The stress distributions were compared for varying material properties of the prosthesis.

The Maximum Principal and Von Mises stress distributions for the composite material polystyrene, which has an Elastic Modulus of $1.4*10^3$ MPa, are presented in Appendix C.1 and C.2. The weight of the prosthesis using polystyrene was found to be 1.094 kg. Similarly, the Maximum Principal and Von Mises stress distributions for fiber glass reinforced polyester resin, which has an Elastic Modulus of $1.4*10^4$ MPa, are presented in Appendix C.3 and C.4. The weight of this prosthesis was found to be 1.9598 kg.

When the stress distributions of the two designs were compared the pattern of the stress distributions was not different but there was a large difference in the magnitudes of the stresses. For both the polyester resin and the polystyrene designs, the maximum principal stresses obtained from the analysis were highest (about 63 KPa & 65 KPa respectively) at the posterior knee region of the prosthesis. The Von Mises stresses obtained were highest (60 KPa & 52 KPa respectively) at the anterior and posterior tips of the knee region.

The Maximum Principal and Von Mises stress distributions for the loading condition for the foot-flat are presented in Appendix C.5 and C.6 for the composite material polystyrene. The stress distributions for glass reinforced polyester are presented in Appendix C.7 and C.8. The pattern of

Table 8.1 Weights of the solid and hollow models of the prosthesis for different materials

Material	Weight of solid	Weight of hollow
Fiberglass-reinforced polyester resin	1.9798 kg.	0.3713 kg.
Fiberglass-reinforced epoxy	1.9173 kg.	0.3596 kg.
Polystyrene	1.094 kg.	0.2052 kg.

the stress distributions for foot-flat was not significantly different but there was a significant difference in the stress magnitudes. For both the polyester resin and polystyrene designs, the maximum principal stresses obtained were highest (about 16 KPa and 18 KPa respectively) at the anterior ankle region of the prosthesis. The Von Mises stresses were highest (26 KPa and 25 KPa) at the posterior knee region of the prosthesis.

The resulting stress distributions indicated that higher stresses are present during the loading condition of heel-strike compared to the loading condition of foot-flat. Also, the weight of the polystyrene prosthesis (1.094 kg.) was less than the weight of the glass reinforced polyester prosthesis (1.9598 kg.). The disadvantage of the low weight polystyrene prosthesis is that the displacement (deformation) of this prosthesis under both the loading conditions is large as presented in Appendix C.9 and C.10. This results in problems such as instability and loss of balance during walking on the part of the amputee. In such a case, a trade-off between weight and stress distribution should be made in order to design an optimal prosthesis.

A hollow model of the prosthesis was constructed in order to determine the effect of the geometry of the prosthesis on its weight. The weight of the initial model was 1.9598 kg. for the fiber glass reinforced polyester (Appendix C.11). The hollow model was constructed by scaling the initial model by a factor of 0.9. Then the initial model was cut by this scaled model, which resulted in the hollow model. The hollow model had a weight of 0.3713 kg. which is a reduction of about 85-90% from the initial model (Appendix C.12). The weights of the solid and hollow models of the prosthesis obtained for varying material properties are shown in Table 8.1.

CHAPTER 9

CONCLUSIONS AND RECOMMENDATIONS

9.1 Conclusions

The study of energy storing prosthetic feet in this thesis have led to the conclusion that energy storing prosthetic feet approach normal running gait pattern more closely than traditional prosthetic feet. Energy storing prosthetic feet store energy during stance and release this energy during toe-off to passively propel the limb. These feet allow the amputee to walk and run smoothly and conserve more energy.

The Flex Foot provides the greatest energy storing potential and can be used for a wide range of activities such as running and vigorous sports. For the geriatric amputee, energy storing prosthetic feet provide a decrease in the effort required for walking due to low weight and increased responsiveness. For the young and athletic amputee, the flexibility of energy storing prosthetic feet facilitate participation in sports.

While prescribing a prosthetic foot for an amputee, the design characteristics, materials, advantages and disadvantages of energy storing prosthetic feet should be considered. The age, weight, financial resources and activity level of the amputee should also be considered in order to provide the amputee with the most optimal prosthesis.

Research studies, comparing gait in amputees using different prosthetic feet, provide quantitative information to help in the selection of a prosthetic foot. Most of the research studies reviewed here confirmed to a certain extent the energy conserving capabilities of energy storing prosthetic feet compared to conventional feet.

In this thesis, a finite element model of a prosthesis was used to compare stress distributions for the loading conditions of heel-strike and foot-flat. Various geometric and material properties were represented in this design. Optimization was repeated while varying the geometry and material properties of the prosthesis to determine if the weight of the prosthesis could be reduced by the use of different geometries and materials.

From this study, we can conclude that finite element modeling and analysis is very useful in identifying the importance of geometry, material properties and loading conditions in the design of a prosthesis. A particular design of a prosthesis can be evaluated for weight and stress distribution, without actually fabricating and clinically testing a prosthesis, which is expensive and time-consuming.

A prosthesis composed of a material having a low modulus of elasticity had a reduced weight compared to the prosthesis composed of the high modulus material. The results indicate that fabricating a prosthesis, with a material having an elastic modulus 10 times smaller, generally results in reduction of stresses within the prosthesis. Therefore, in terms of material choice, polystyrene is preferred over glass reinforced polyester resin because it produces more favorable conditions.

A lower modulus model results in a more flexible prosthesis, which is definitely an advantage for a young and athletic amputee. However, the low modulus material may result in instability in walking for the geriatric amputee, due to the increased deformation of the prosthesis. Also, the fatigue or deformation limit influence the extent to which the modulus of elasticity can be reduced in the design.

The stress distributions for the loading condition of heel-strike were generally higher compared to those for the loading condition of foot-flat. A hollow model resulted in a reduction of weight (about 90%) compared to the solid model.

These results showed a significant dependence of stress within the prosthesis on the loading conditions, geometry and material properties of the prosthesis. Also, the weight of the prosthesis was found to be dependent upon the material properties and the geometry of the prosthesis. Therefore, from these results a better understanding of the behavior and the requirements of a below-knee prosthesis was obtained.

9.2 Recommendations

More clinical research studies, comparing gait in amputees using different prosthetic feet should be carried out order to provide objective data regarding gait characteristics, energy cost and subjective preferences. Only quantitative and scientific studies, such as those reviewed in this thesis, will be able to confirm or deny the functional advantages obtained from energy storing prosthetic feet.

One limitation of this finite element study is the approximation of the prosthetic structure. In the solid model, the profiles used may not completely model the actual shape of the prosthesis. Most of the current shape measurement methods rely on artisan techniques. Such techniques depend on approximate measurements and the artisans perception of the amputees normal limb.

To overcome such limitations, a CAD/CAM system has been developed at the Kessler Institute of Rehabilitation (38) for producing the outer, cosmetic shape of the prosthesis. The system uses a 3D shape sensing instrument to measure the amputees contralateral limb. This shape can be easily quantified and reproduced. A PC-based CAD program accepts the shape data from the shape sensing instrument and creates a computer model of the prosthesis. This program is similar to the Solid Modeling family of I-DEAS, which can also be used to develop a solid model of the prosthesis.

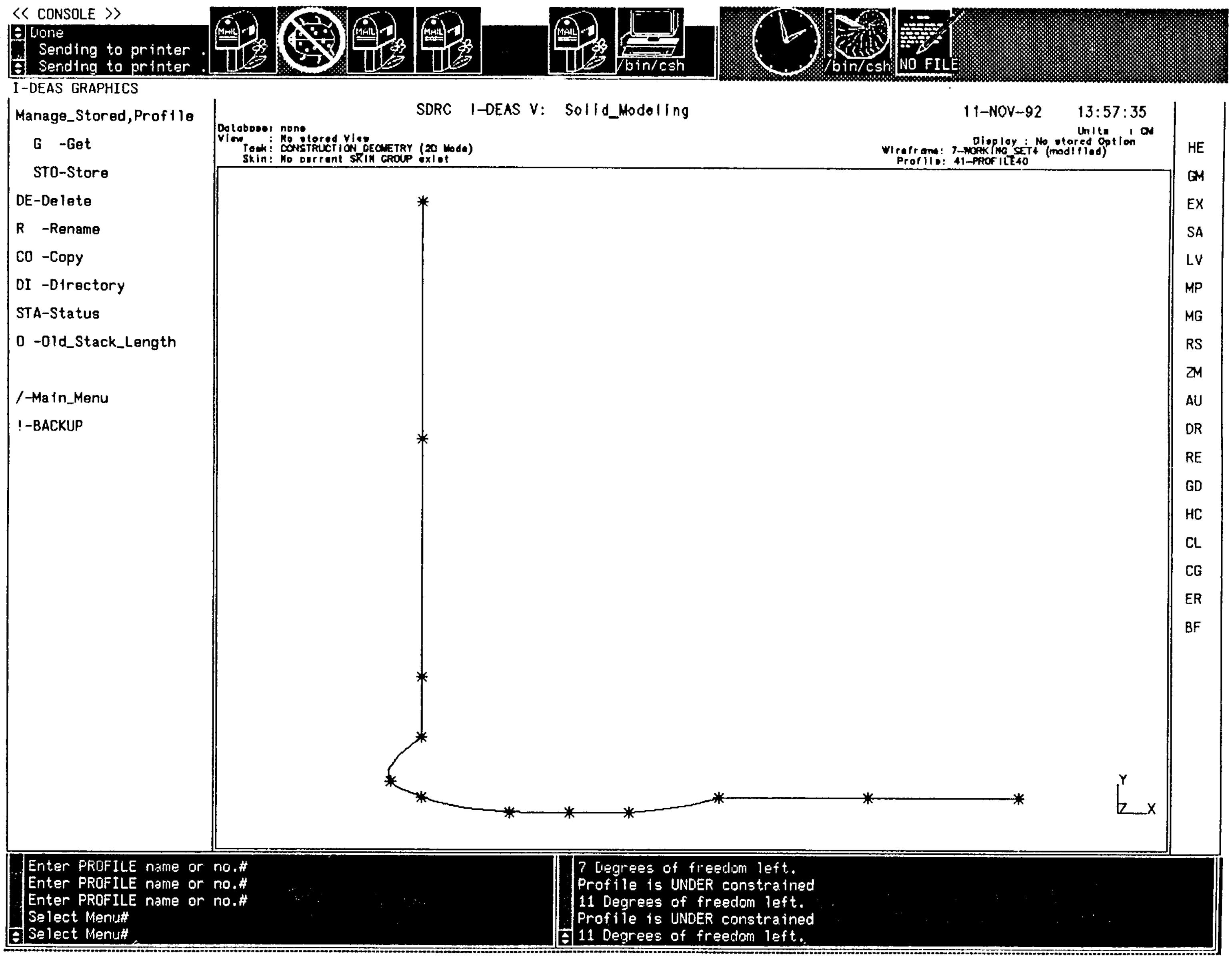
This shape can be modified if desired to obtain optimal weight and stress distributions and the modified shape is then output to a computer-numerically-controlled (CNC) carving machine which carves the desired shape from the prosthetic material.

The advantage of such a CAD/CAM system is that the system may be used to fabricate several types of prosthesis for use in the evaluation of a design. The variable parameters in each design can be monitored. Such a technique provides a scientific and quantitative base for developments in prosthetic design. Several different designs can be evaluated prior to use by using finite element analysis. Changes can be made based on the finite element analysis and the design that provides the most desirable weight and stresss distributions can be selected.

This study assumed the model of the prosthesis to be composed of a material having a linear modulus of elasticity. Further studies in the design of a below-knee prosthesis could take into consideration a material having a non-linear modulus of elasticity.

In the future, it is very important to use scientific principles in prosthetic design and to test and evaluate new designs effectively. Experimental studies should be carried out to verify the results obtained and to obtain data which can be applied to the computer models. Finally, any future progress in this area of biomechanics requires a detailed knowledge of both applied mechanics and biological factors.

APPENDIX A



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DE-Delete
R -Rename
CO -Copy
DI -Directory
STA-Status
O -Old_Stack_Length
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EX
SA
LV
MP
MG
RS
ZM
AU
DR
RE
GD
HC
CL
CG
ER
BF
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Enter PROFILE name or no.#
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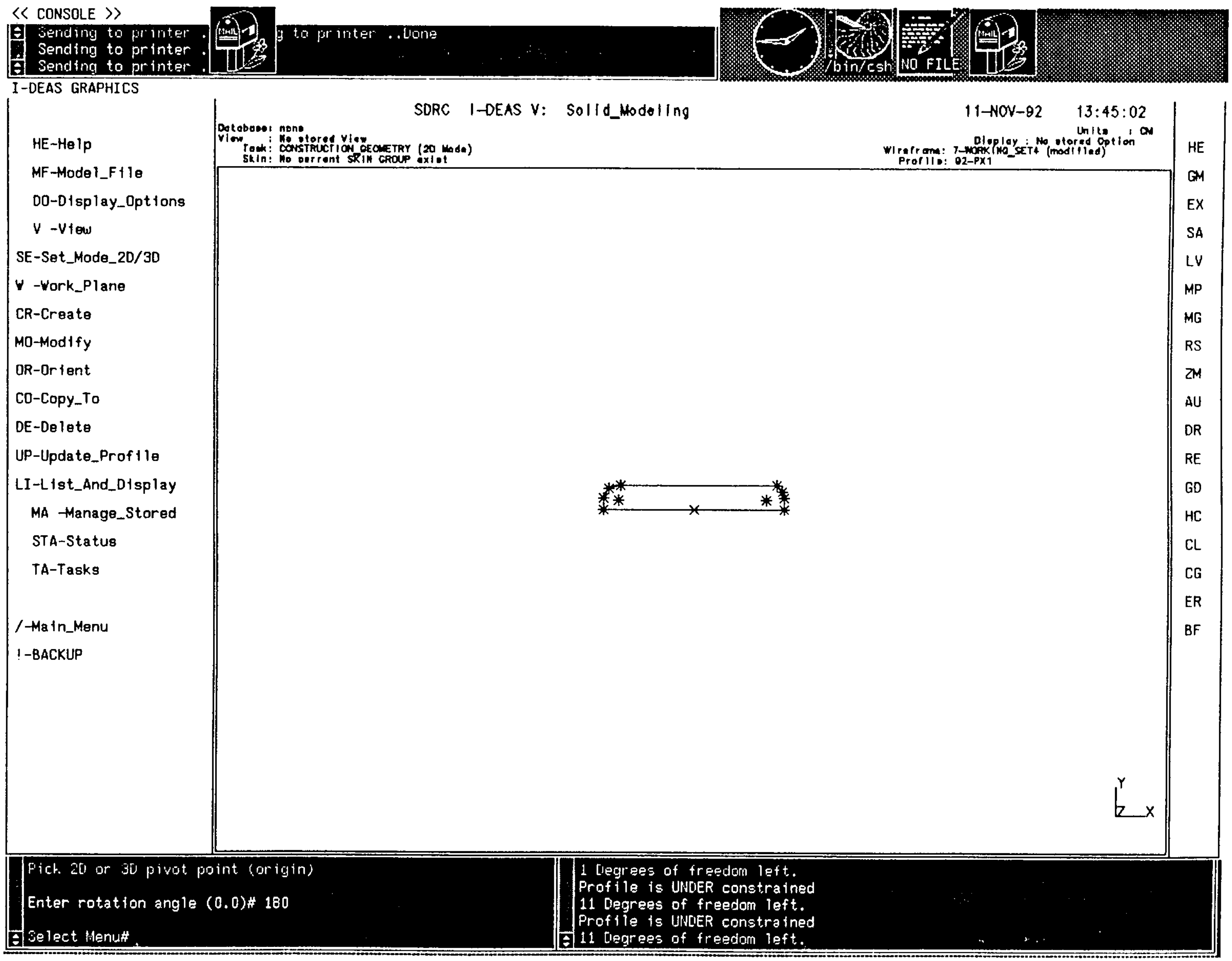
Profile is UNDER constrained
11 Degrees of freedom left.

A.1



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Display : No stored Option
Wireframe: 7-WORK(NO_SET4 (modified)
Profile: 92-PX1
HE-Help
MF-Model_File
DO-Display_Options
V -View
SE-Set_Mode_2D/3D
W -Work_Plane
CR-Create
MO-Modify
OR-Orient
CO-Copy_To
DE-Delete
UP-Update_Profile
LI-List_And_Display
MA -Manage_Stored
STA-Status
TA-Tasks
/-Main_Menu
!-BACKUP
HE
GM
EX
SA
LV
MP
MG
RS
ZM
AU
DR
RE
GD
HC
CL
CG
ER
BF
Y
Z
X
Pick 2D or 3D pivot point (origin)
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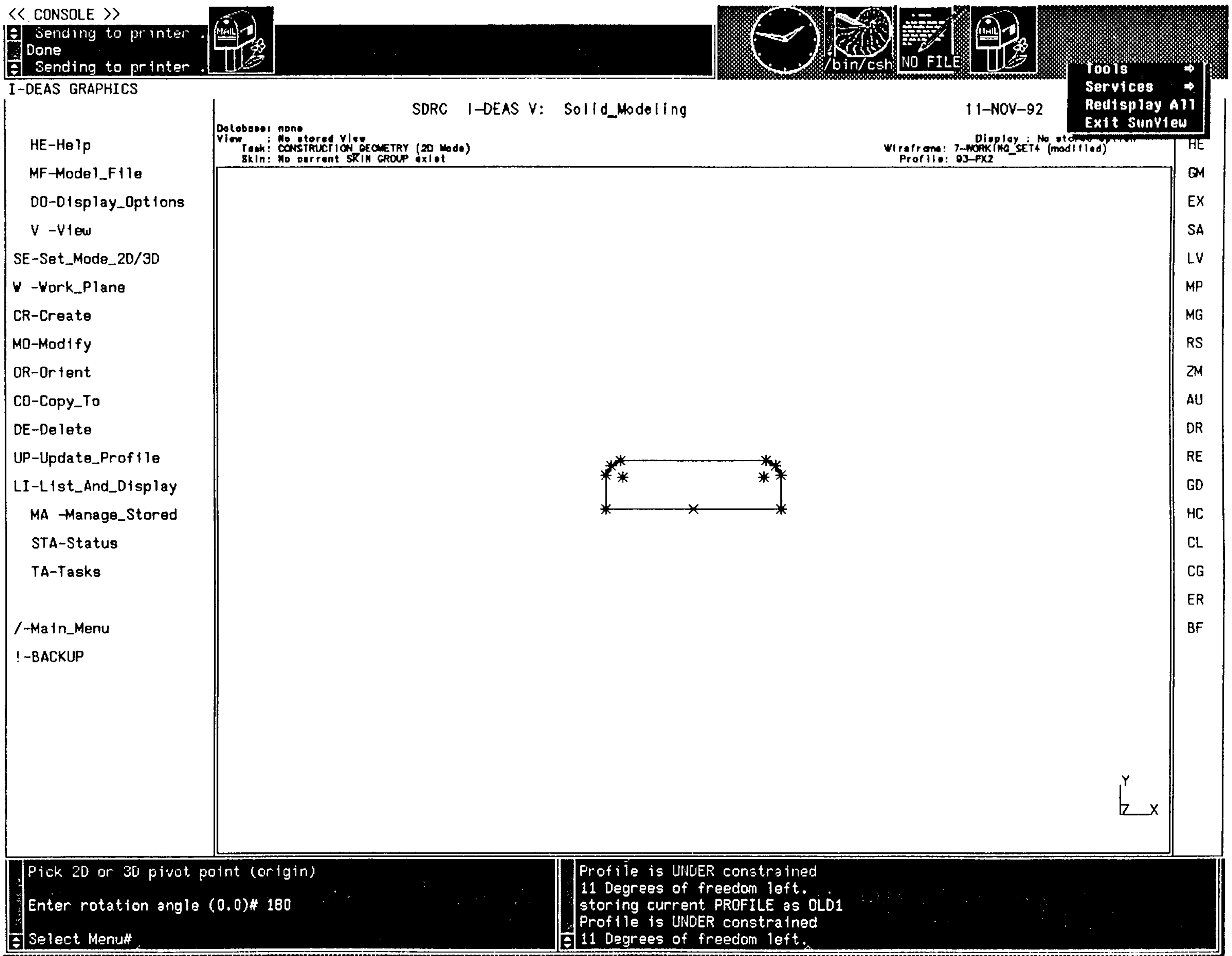
11 Degrees of freedom left.
Profile is UNDER constrained
11 Degrees of freedom left.

A.2



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SDRC I-DEAS V: Solid_Modeling
11-NOV-92
Database: none
View : No stored View
Task: CONSTRUCTION GEOMETRY (2D Mode)
Skin: No current SKIN GROUP exist
Wireframe: 7-WORK(NO_SET4 (modified)
Profile: 93-PX2
HE-Help
MF-Model_File
DO-Display_Options
V -View
SE-Set_Mode_2D/3D
W -Work_Plane
CR-Create
MO-Modify
OR-Orient
CO-Copy_To
DE-Delete
UP-Update_Profile
LI-List_And_Display
MA -Manage_Stored
STA-Status
TA-Tasks
/-Main_Menu
!-BACKUP
HE
GM
EX
SA
LV
MP
MG
RS
ZM
AU
DR
RE
GD
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CG
ER
BF
Y
Z
X
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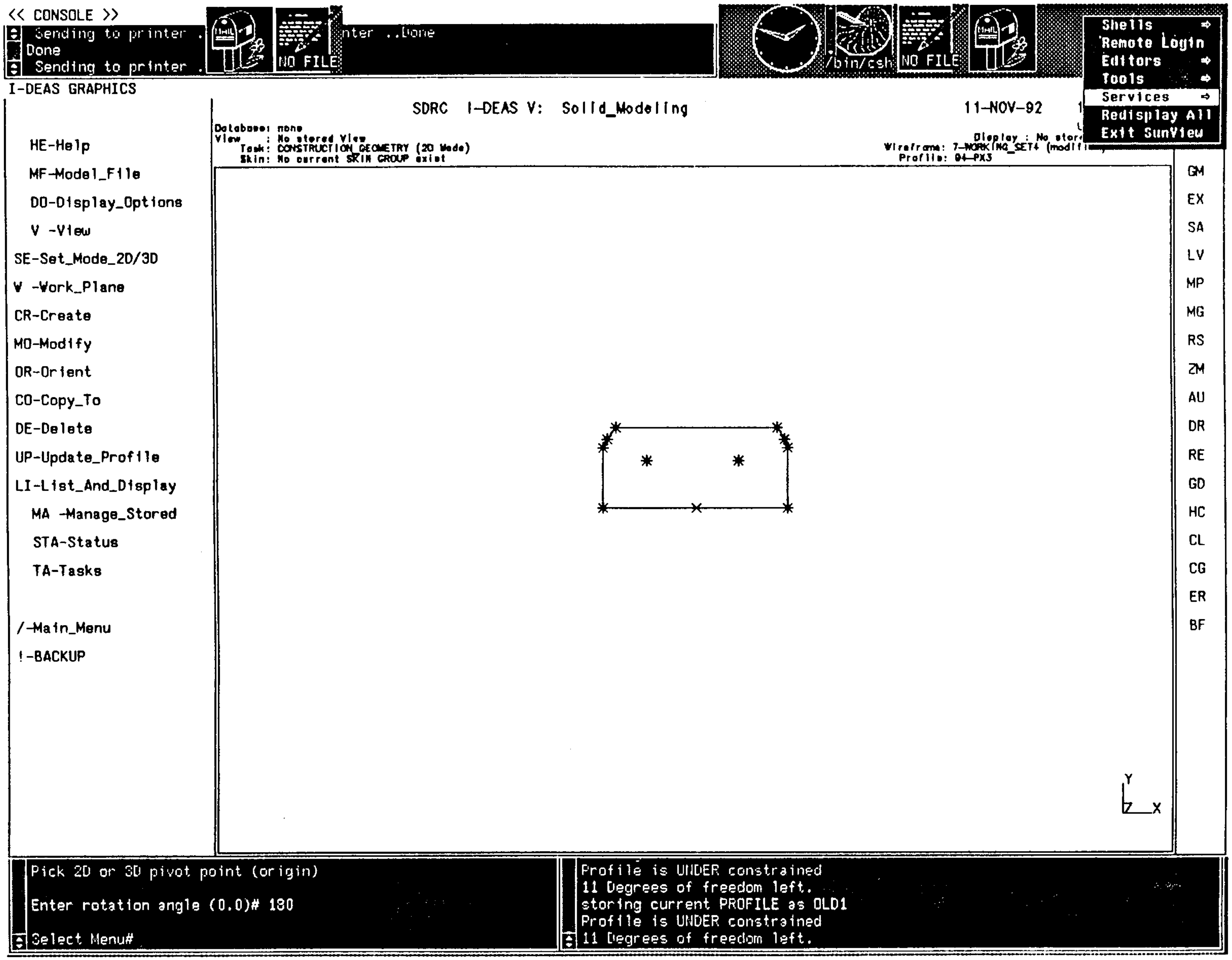
Profile is UNDER constrained
11 Degrees of freedom left.

A.3



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Remote Login
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Exit SunView
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SDRC I-DEAS V: Solid_Modeling
11-NOV-92
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DO-Display_Options
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SE-Set_Mode_2D/3D
W -Work_Plane
CR-Create
MO-Modify
OR-Orient
CO-Copy_To
DE-Delete
UP-Update_Profile
LI-List_And_Display
MA -Manage_Stored
STA-Status
TA-Tasks
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GM
EX
SA
LV
MP
MG
RS
ZM
AU
DR
RE
GD
HC
CL
CG
ER
BF
Y
Z
X
Pick 2D or 3D pivot point (origin)
Enter rotation angle (0.0)# 180
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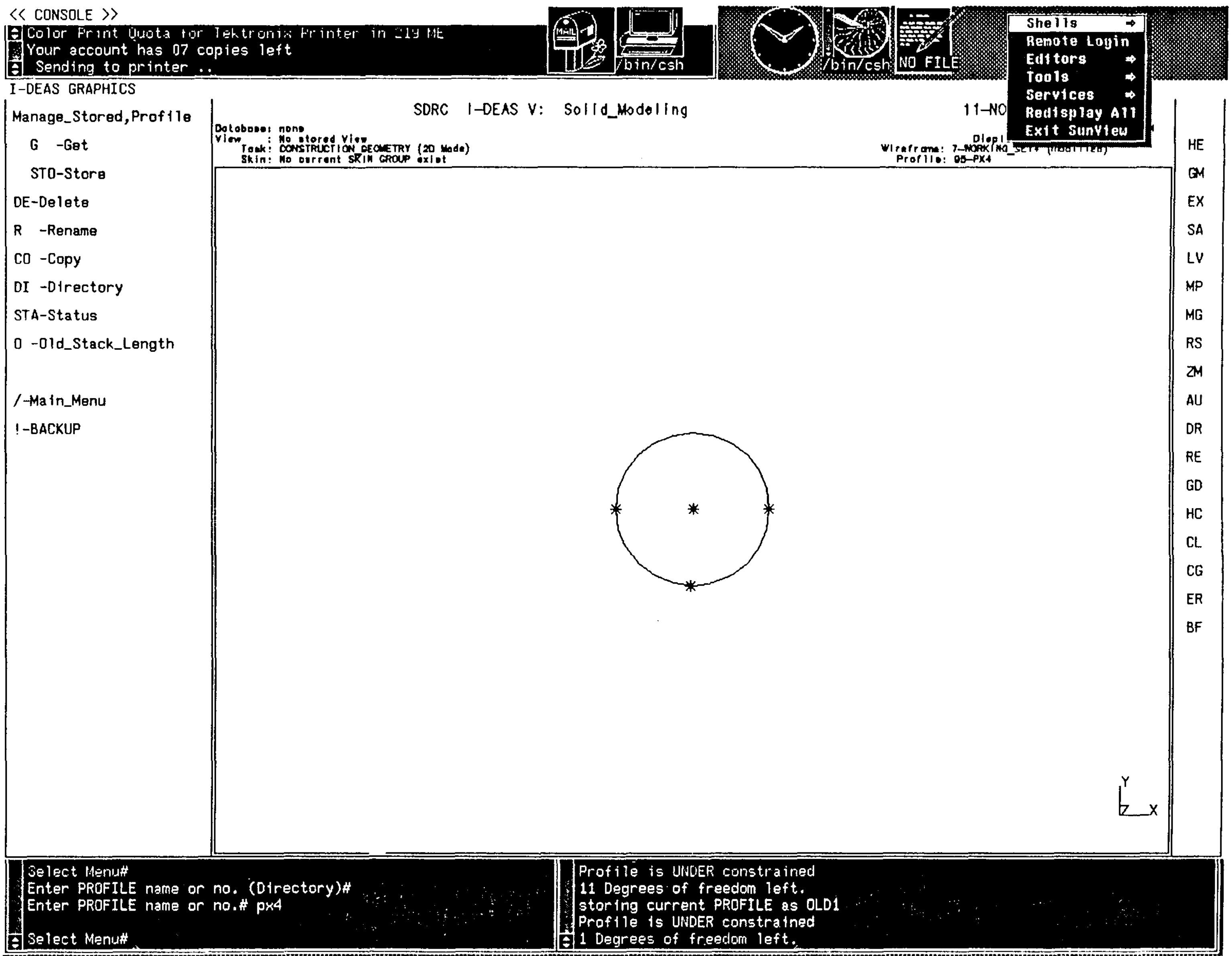
11 Degrees of freedom left.
storing current PROFILE as OLD1
Profile is UNDER constrained
11 Degrees of freedom left.

A.4



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Color Print Quota for Tektronix Printer in 219 ME
Your account has 07 copies left
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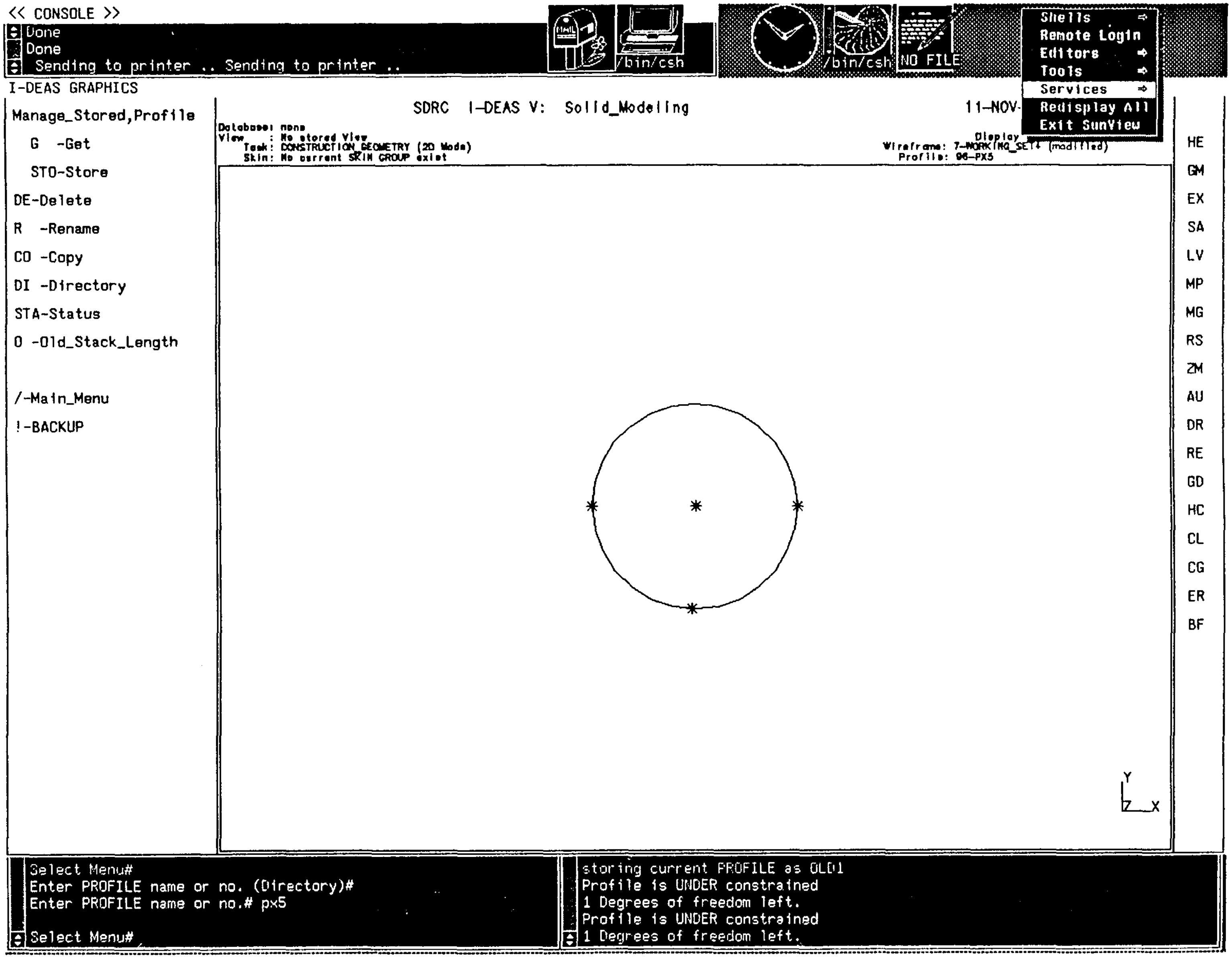
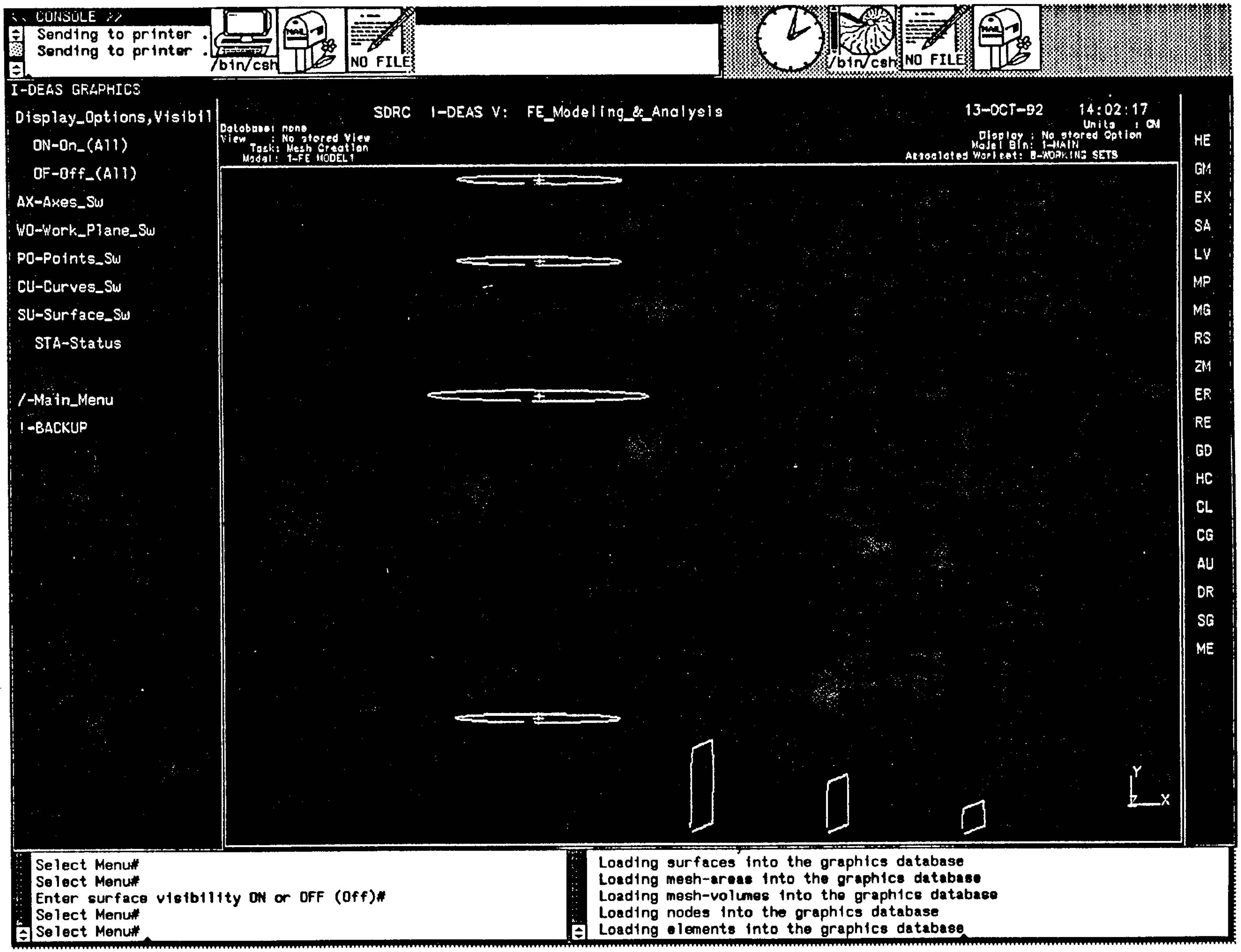
Shells
Remote Login
Editors
Tools
Services
Redisplay All
Exit SunView
I-DEAS GRAPHICS
Manage_Stored,Profile
G -Get
STO-Store
DE-Delete
R -Rename
CO -Copy
DI -Directory
STA-Status
O -Old_Stack_Length
/-Main_Menu
!-BACKUP
SDRC I-DEAS V: Solid_Modeling
Database: none
View : No stored View
Task: CONSTRUCTION GEOMETRY (2D Mode)
Skin: No current SKIN GROUP exist
Profile: 95-PX4
HE
GM
EX
SA
LV
MP
MG
RS
ZM
AU
DR
RE
GD
HC
CL
CG
ER
BF
Y
Z X
Select Menu#
Enter PROFILE name or no. (Directory)#
Enter PROFILE name or no.# px4
Select Menu#
Profile is UNDER constrained
11 Degrees of freedom left.
storing current PROFILE as OLD1
Profile is UNDER constrained
1 Degrees of freedom left.

A.5

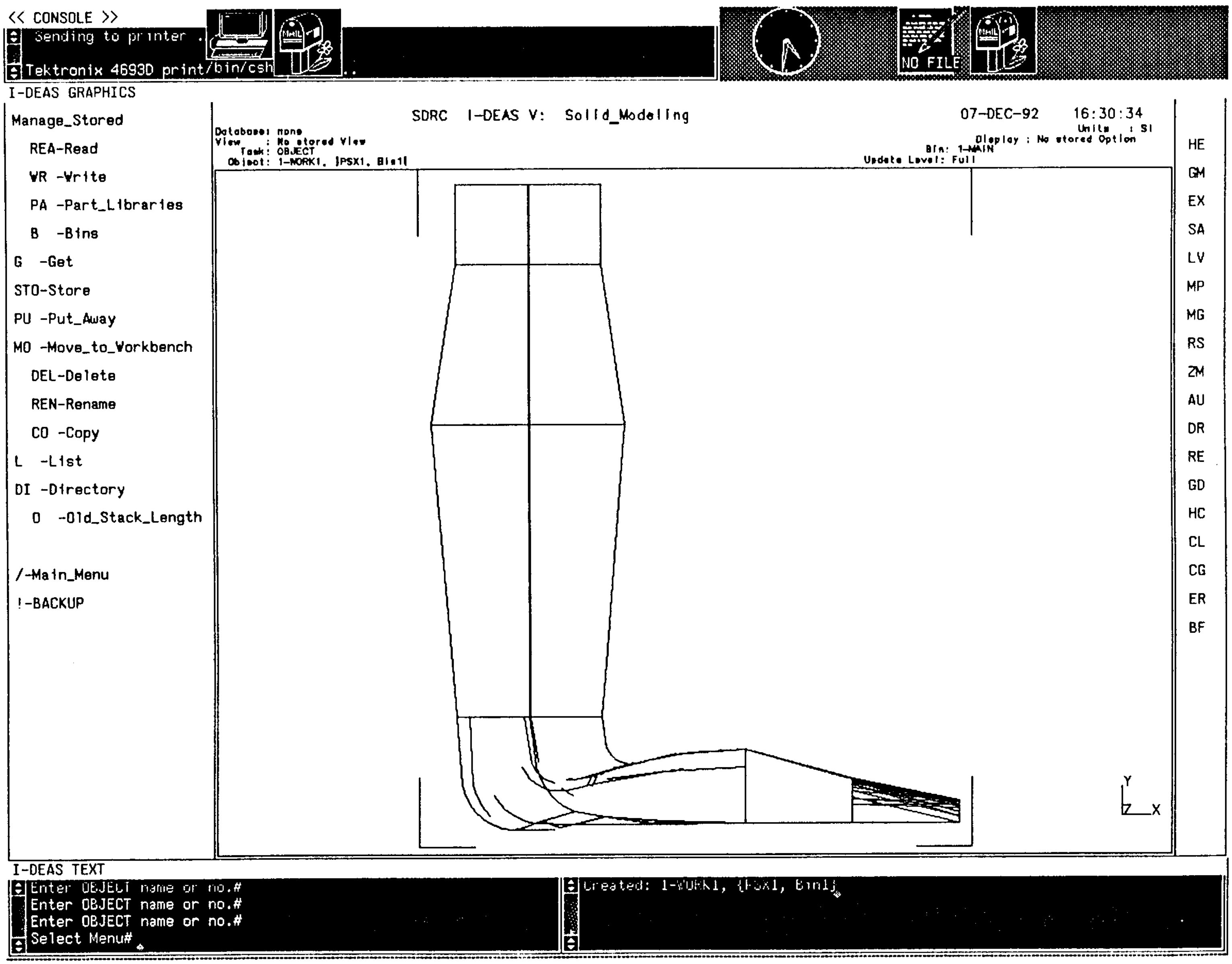

<< CONSOLE >>
Done
Done
Sending to printer .. Sending to printer ..
/bin/csh
/bin/csh
NO FILE
Shells
Remote Login
Editors
Tools
Services
Redisplay All
Exit SunView
I-DEAS GRAPHICS
Manage_Stored,Profile
G -Get
STO-Store
DE-Delete
R -Rename
CO -Copy
DI -Directory
STA-Status
O -Old_Stack_Length
/-Main_Menu
!-BACKUP
SDRC I-DEAS V: Solid_Modeling
Task: CONSTRUCTION GEOMETRY (2D Mode)
HE
GM
EX
SA
LV
MP
MG
RS
ZM
AU
DR
RE
GD
HC
CL
CG
ER
BF
Y
Z
X
Select Menu#
Enter PROFILE name or no. (Directory)#
Enter PROFILE name or no.# px5
Select Menu#
storing current PROFILE as OLD1
Profile is UNDER constrained
1 Degrees of freedom left.
Profile is UNDER constrained
1 Degrees of freedom left.

A.6


CONSOLE
Sending to printer .
Sending to printer .
/bin/csh
NO FILE
/bin/csh
NO FILE
I-DEAS GRAPHICS
Display_Options,Visibil
ON-On_(All)
OF-Off_(All)
AX-Axes_Sw
WO-Work_Plane_Sw
PO-Points_Sw
CU-Curves_Sw
SU-Surface_Sw
STA-Status
/-Main_Menu
!-BACKUP
SDRC I-DEAS V: FE_Modeling_&_Analysis
13-OCT-92 14:02:17
Units : CM
Database: none
View : No stored View
Task: Mesh Creation
Model: 1-FE MODEL1
Display : No stored Option
Model Bin: 1-MAIN
Associated Workset: 8-WORKING SETS
HE
GM
EX
SA
LV
MP
MG
RS
ZM
ER
RE
GD
HC
CL
CG
AU
DR
SG
ME
Y
X
Select Menu#
Select Menu#
Enter surface visibility ON or OFF (Off)#
Select Menu#
Select Menu#
Loading surfaces into the graphics database
Loading mesh-areas into the graphics database
Loading mesh-volumes into the graphics database
Loading nodes into the graphics database
Loading elements into the graphics database

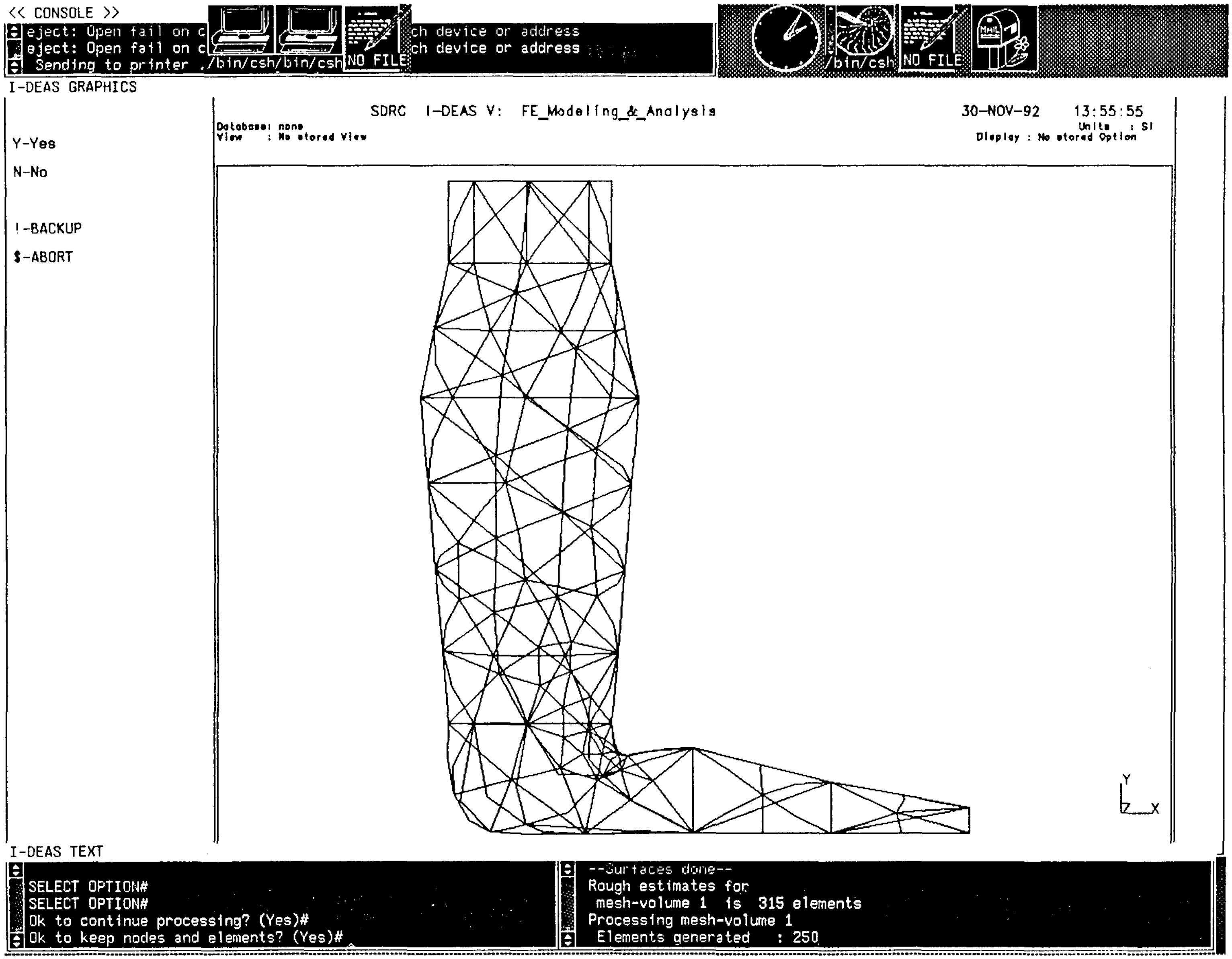
A.7

APPENDIX B

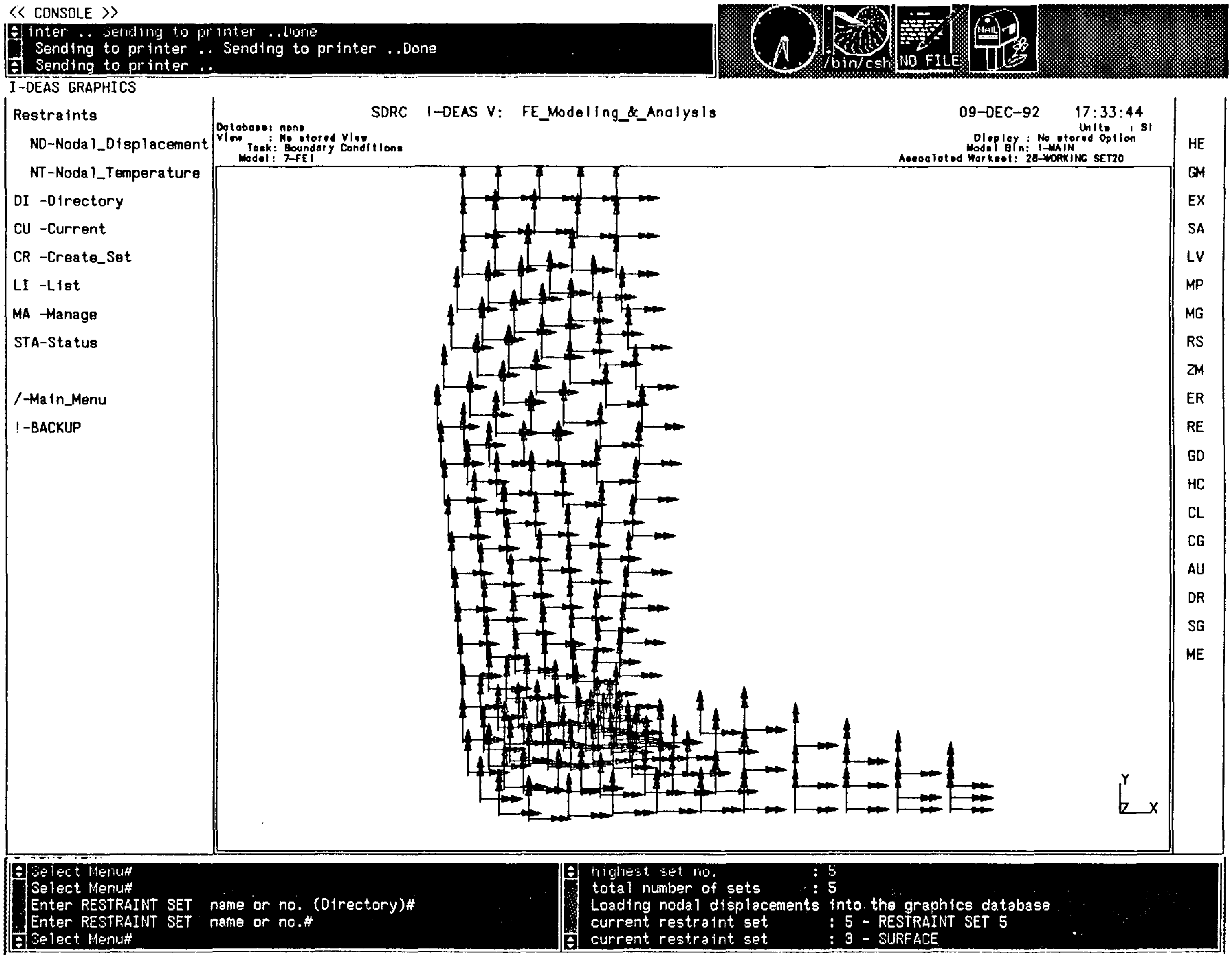


<< CONSOLE >>
Sending to printer .
Tektronix 4693D print/bin/csh
NO FILE
I-DEAS GRAPHICS
Manage_Stored
REA-Read
WR -Write
PA -Part_Libraries
B -Bins
G -Get
STO-Store
PU -Put_Away
MO -Move_to_Workbench
DEL-Delete
REN-Rename
CO -Copy
L -List
DI -Directory
O -Old_Stack_Length
/-Main_Menu
!-BACKUP
SDRC I-DEAS V: Solid_Modeling
07-DEC-92 16:30:34
Database: none
View : No stored View
Task: OBJECT
Units : SI
Display : No stored Option
Bin: 1-MAIN
Update Level: Full
HE
GM
EX
SA
LV
MP
MG
RS
ZM
AU
DR
RE
GD
HC
CL
CG
ER
BF
Y
Z
X
I-DEAS TEXT
Enter OBJECT name or no.#
Enter OBJECT name or no.#
Enter OBJECT name or no.#
Select Menu#

B.1


<< CONSOLE >>
eject: Open fail on c
eject: Open fail on c
Sending to printer
./bin/csh/bin/csh
NO FILE
ch device or address
ch device or address
/bin/csh
NO FILE
I-DEAS GRAPHICS
Y-Yes
N-No
!-BACKUP
$-ABORT
SDRC I-DEAS V: FE_Modeling_&_Analysis
30-NOV-92 13:55:55
Database: none
View : No stored View
Units : SI
Display : No stored Option
Y
Z
X
I-DEAS TEXT
SELECT OPTION#
SELECT OPTION#
Ok to continue processing? (Yes)#
Ok to keep nodes and elements? (Yes)#
--Surfaces done--
Rough estimates for
mesh-volume 1 is 315 elements
Processing mesh-volume 1
Elements generated : 250

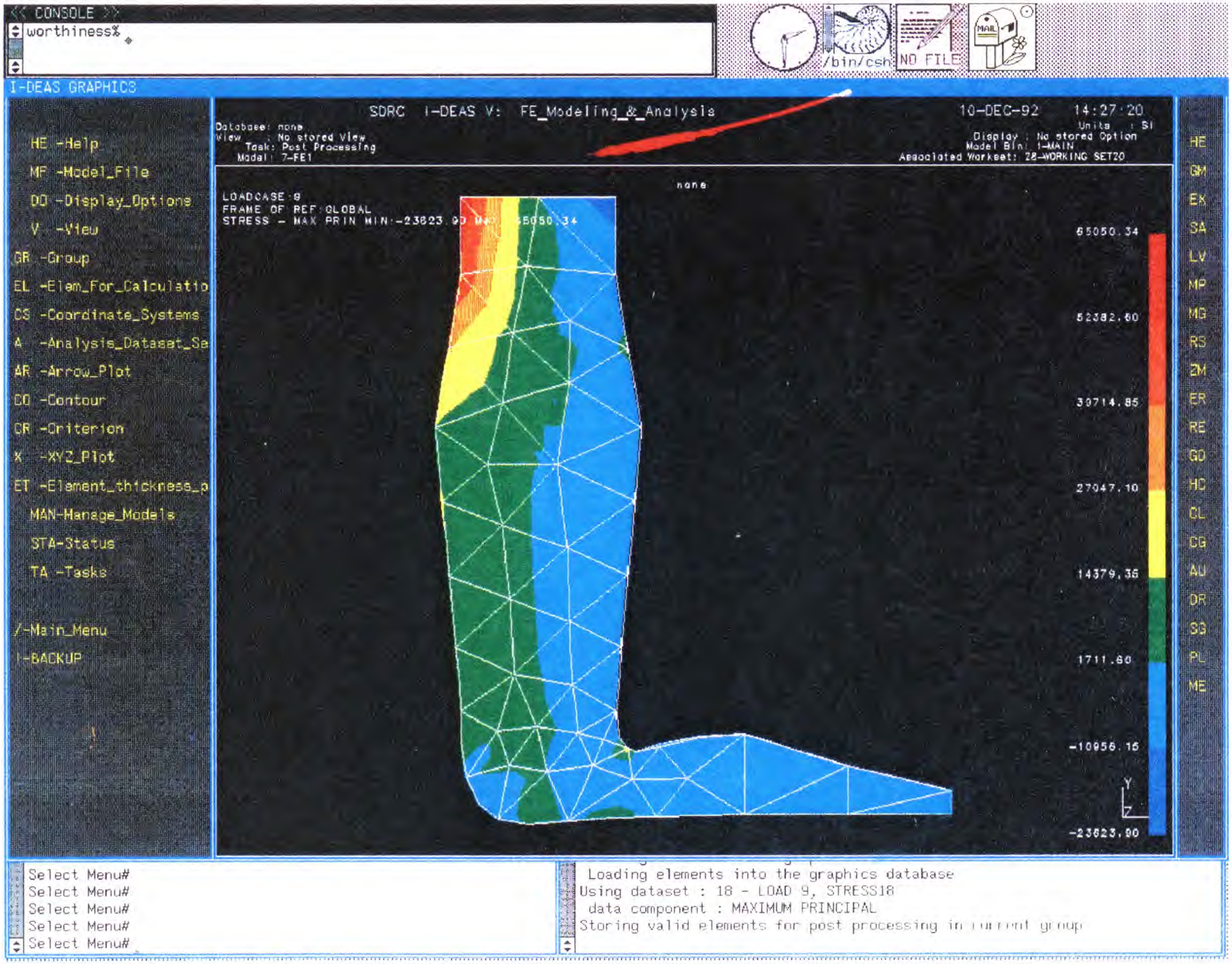
B.2



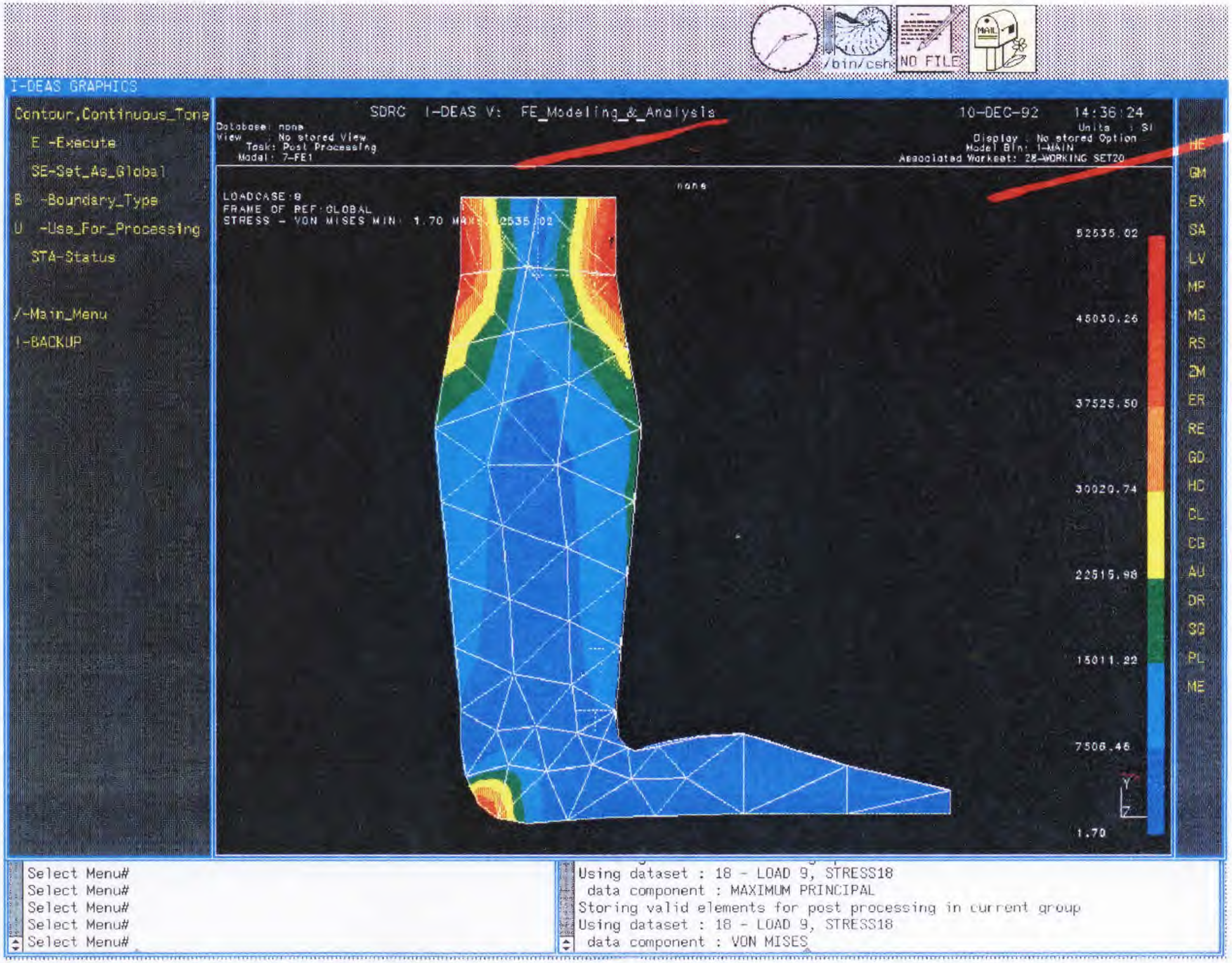
<< CONSOLE >>
inter .. Sending to printer ..Done
Sending to printer .. Sending to printer ..Done
Sending to printer ..
/bin/csh
NO FILE
I-DEAS GRAPHICS
Restraints
ND-Nodal_Displacement
NT-Nodal_Temperature
DI -Directory
CU -Current
CR -Create_Set
LI -List
MA -Manage
STA-Status
/-Main_Menu
!-BACKUP
SDRC I-DEAS V: FE_Modeling_&_Analysis
09-DEC-92 17:33:44
Database: none
View : No stored View
Task: Boundary Conditions
Model: 7-FE1
Units : SI
Display : No stored Option
Model Bin: 1-MAIN
Associated Workset: 28-WORKING SET20
HE
GM
EX
SA
LV
MP
MG
RS
ZM
ER
RE
GD
HC
CL
CG
AU
DR
SG
ME
Y
Z X
Select Menu#
Select Menu#
Enter RESTRAINT SET name or no. (Directory)#
Enter RESTRAINT SET name or no.#
Select Menu#
highest set no. : 5
total number of sets : 5
Loading nodal displacements into the graphics database
current restraint set : 5 - RESTRAINT SET 5
current restraint set : 3 - SURFACE

B.3

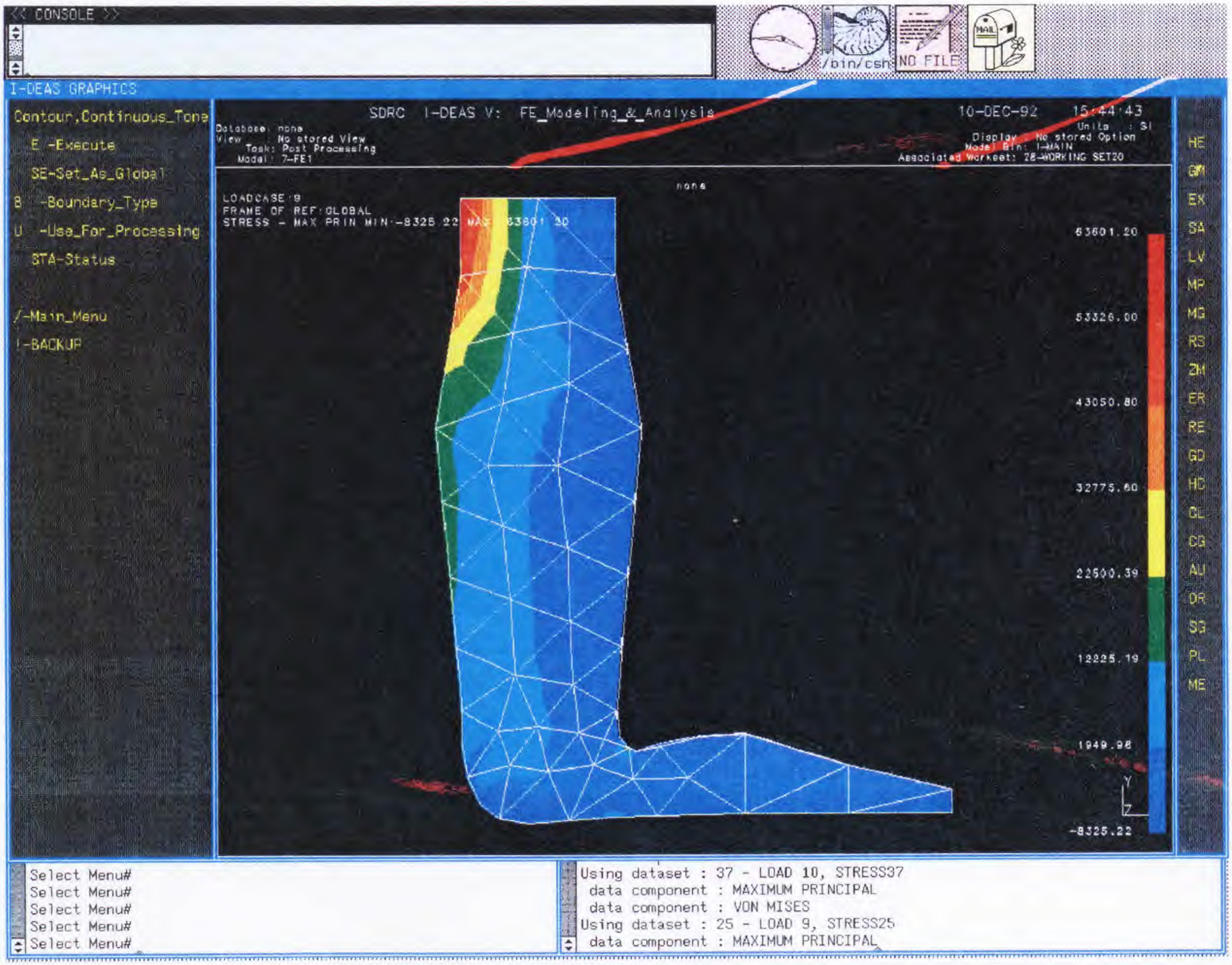
APPENDIX C


CONSOLE
worthiness%
/bin/csh
NO FILE
I-DEAS GRAPHICS
SDRC I-DEAS V: FE_Modeling_&_Analysis
10-DEC-92 14:27:20
HE -Help
MF -Model_File
DO -Display_Options
V -View
GR -Group
EL -Elem_For_Calculatio
CS -Coordinate_Systems
A -Analysis_Dataset_Se
AR -Arrow_Plot
CO -Contour
CR -Criterion
X -XYZ_Plot
ET -Element_thickness_p
MAN-Manage_Models
STA-Status
TA -Tasks
/-Main_Menu
!-BACKUP
LOADCASE 9
FRAME OF REF:GLOBAL
65050.34
52382.60
39714.85
27047.10
14379.35
1711.60
-10956.15
-23623.90
Select Menu#
Select Menu#
Select Menu#
Select Menu#
Select Menu#
Loading elements into the graphics database
Using dataset : 18 - LOAD 9, STRESS18
data component : MAXIMUM PRINCIPAL

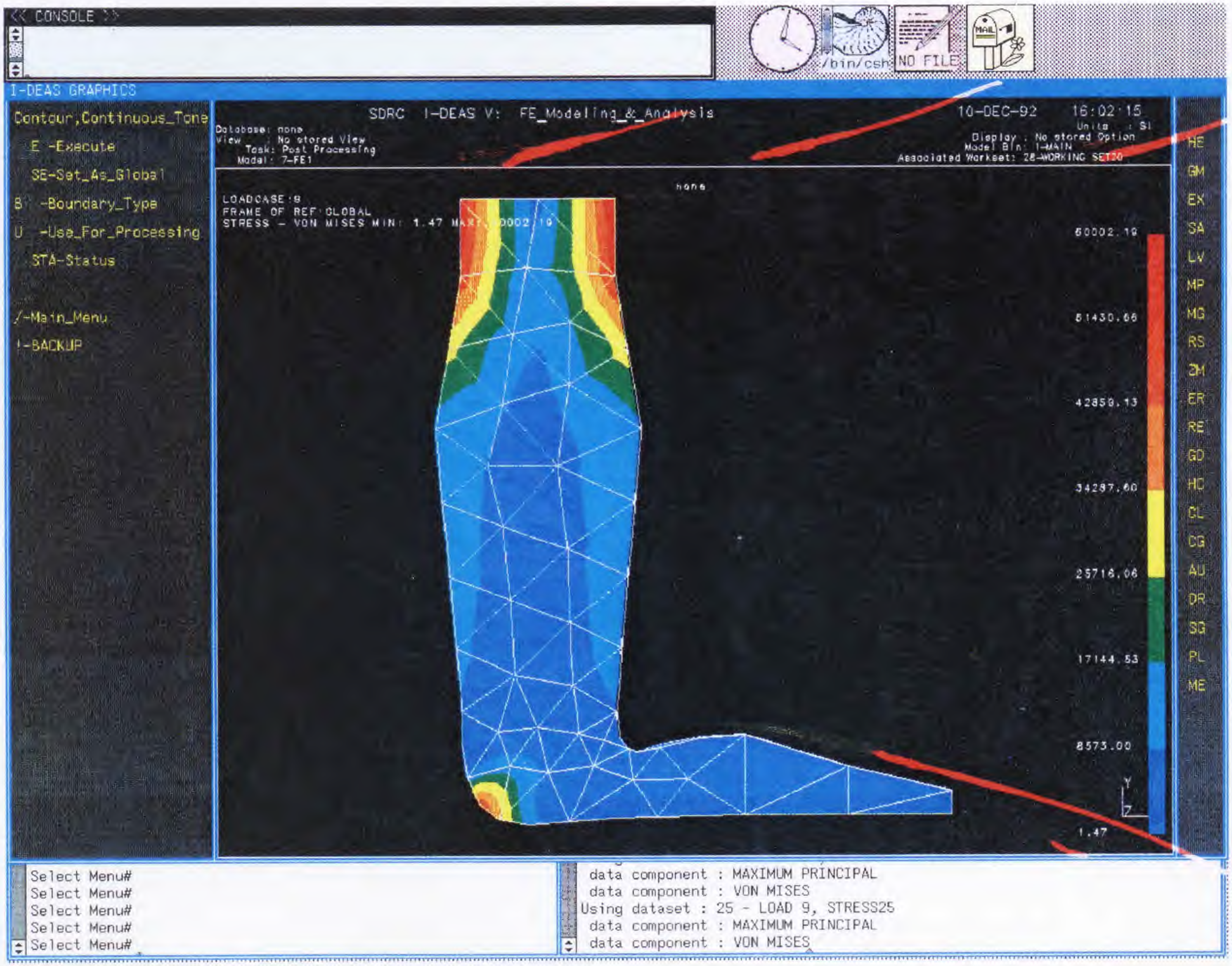
C.1


I-DEAS GRAPHICS
/bin/csh
NO FILE
Contour,Continuous_Tone
E -Execute
SE-Set_As_Global
B -Boundary_Type
U -Use_For_Processing
STA-Status
/-Main_Menu
!-BACKUP
SDRC I-DEAS V: FE_Modeling_&_Analysis
10-DEC-92 14:36:24
LOADCASE:8
FRAME OF REF:GLOBAL
52535.02
45030.26
37525.50
30020.74
22515.98
15011.22
7506.46
1.70
HE
GM
EX
SA
LV
MP
MG
RS
2M
ER
RE
GD
HC
CL
CG
AU
DR
SG
PL
ME
Select Menu#
Select Menu#
Select Menu#
Select Menu#
Select Menu#
Using dataset : 18 - LOAD 9, STRESS18
data component : MAXIMUM PRINCIPAL
Storing valid elements for post processing in current group
Using dataset : 18 - LOAD 9, STRESS18
data component : VON MISES

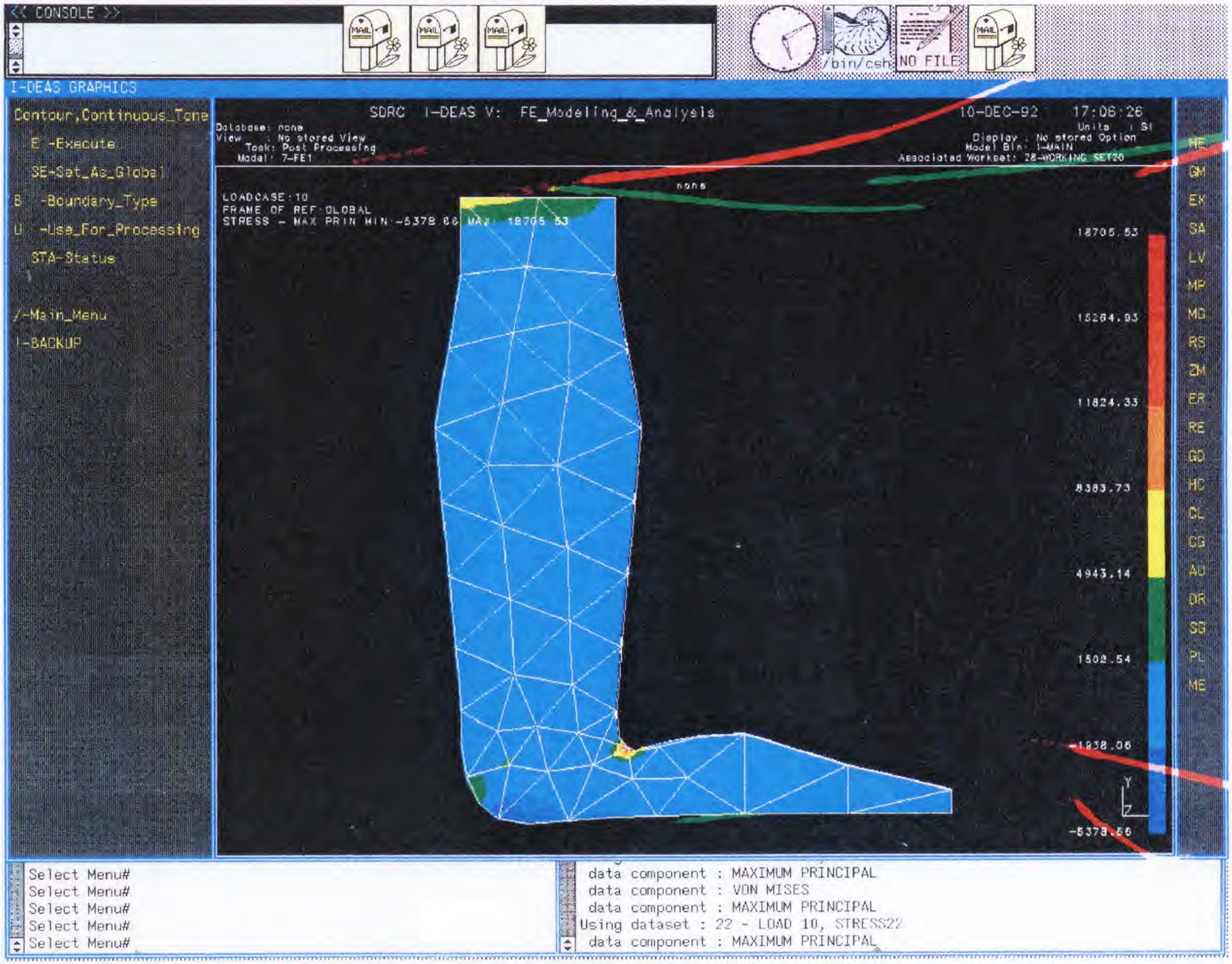
C.2


<< CONSOLE >>
/bin/csh
NO FILE
I-DEAS GRAPHICS
Contour,Continuous_Tone
E -Execute
SE-Set_As_Global
B -Boundary_Type
U -Use_For_Processing
STA-Status
/-Main_Menu
!-BACKUP
SDRC I-DEAS V: FE_Modeling_&_Analysis
10-DEC-92
LOADCASE:9
FRAME OF REF:GLOBAL
63601.20
53326.00
43050.80
32775.60
22500.39
12225.19
1949.98
-8325.22
HE
GM
EX
SA
LV
MP
MG
RS
ZM
ER
RE
GD
HC
CL
CG
AU
DR
SG
PL
ME
Select Menu#
Select Menu#
Select Menu#
Select Menu#
Select Menu#
Using dataset : 37 - LOAD 10, STRESS37
data component : MAXIMUM PRINCIPAL
data component : VON MISES
Using dataset : 25 - LOAD 9, STRESS25
data component : MAXIMUM PRINCIPAL

C.3

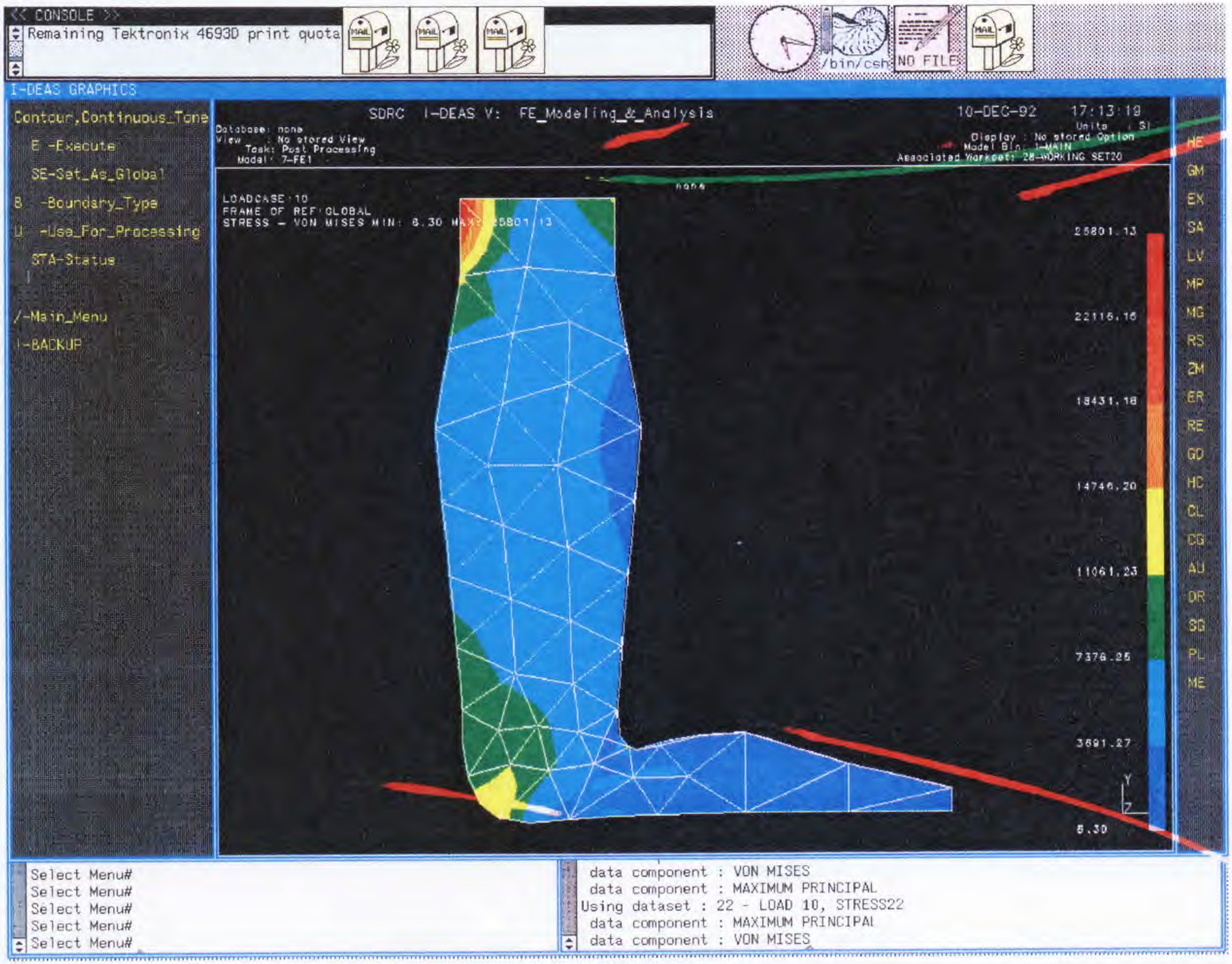

<< CONSOLE >>
/bin/csh
NO FILE
I-DEAS GRAPHICS
Contour,Continuous_Tone
E -Execute
SE-Set_As_Global
B -Boundary_Type
U -Use_For_Processing
STA-Status
/-Main_Menu
!-BACKUP
SDRC I-DEAS V: FE_Modeling_&_Analysis
10-DEC-92 16:02:15
Database: none
View : No stored View
Task: Post Processing
Model: 7-FE1
Units : SI
Display : No stored Option
Model Bin: 1-MAIN
Associated Workset: 28-WORKING SET30
none
LOADCASE:9
FRAME OF REF:GLOBAL
STRESS - VON MISES MIN: 1.47
60002.19
51430.66
42859.13
34287.60
25716.06
17144.53
8573.00
1.47
HE
GM
EX
SA
LV
MP
MG
RS
ER
RE
GD
HC
CL
CG
AU
DR
SG
PL
ME
Select Menu#
Select Menu#
Select Menu#
Select Menu#
Select Menu#
data component : MAXIMUM PRINCIPAL
data component : VON MISES
Using dataset : 25 - LOAD 9, STRESS25
data component : MAXIMUM PRINCIPAL
data component : VON MISES

C.4



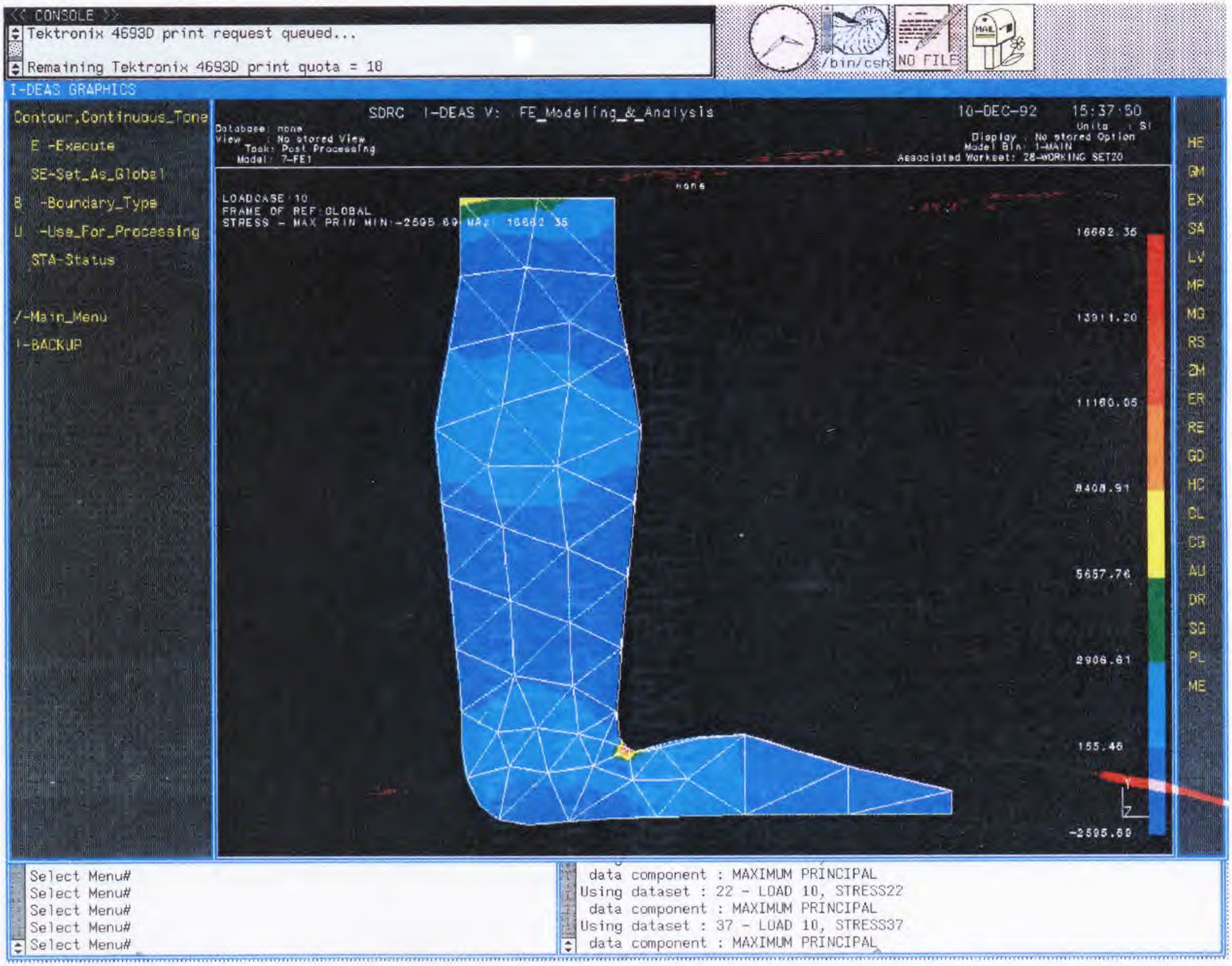
<< CONSOLE >>
/bin/csh
NO FILE
I-DEAS GRAPHICS
Contour,Continuous_Tone
E -Execute
SE-Set_As_Global
B -Boundary_Type
U -Use_For_Processing
STA-Status
/-Main_Menu
!-BACKUP
SDRC I-DEAS V: FE_Modeling_&_Analysis
10-DEC-92 17:06:26
LOADCASE:10
FRAME OF REF:GLOBAL
STRESS - MAX PRIN MIN:-5378.66 MAX: 18705.53
18705.53
15264.93
11824.33
8383.73
4943.14
1502.54
-1938.06
-5378.66
Select Menu#
Select Menu#
Select Menu#
Select Menu#
Select Menu#
data component : MAXIMUM PRINCIPAL
data component : VON MISES
data component : MAXIMUM PRINCIPAL
Using dataset : 22 - LOAD 10, STRESS22
data component : MAXIMUM PRINCIPAL

C.5

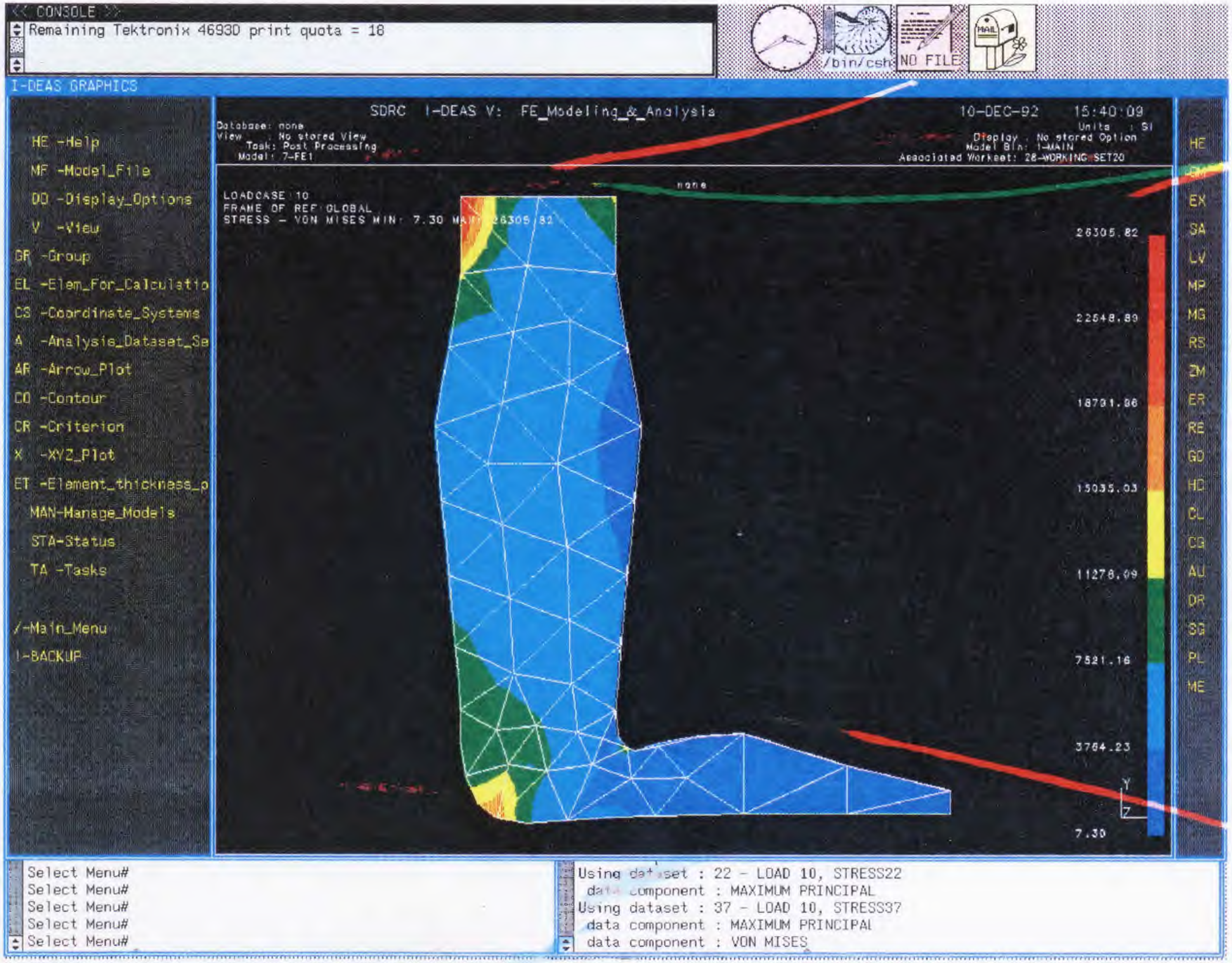


<< CONSOLE >>
Remaining Tektronix 4693D print quota
/bin/csh
NO FILE
I-DEAS GRAPHICS
Contour,Continuous_Tone
E -Execute
SE-Set_As_Global
B -Boundary_Type
U -Use_For_Processing
STA-Status
/-Main_Menu
!-BACKUP
SDRC I-DEAS V: FE_Modeling_&_Analysis
10-DEC-92 17:13:19
Database: none
View : No stored View
Task: Post Processing
Model: 7-FE1
Display : No stored Option
Associated Workset: 28-WORKING SET20
LOADCASE 10
FRAME OF REF GLOBAL
STRESS - VON MISES MIN: 8.30 MAX: 25801.13
25801.13
22116.16
18431.18
14746.20
11061.23
7376.25
3691.27
8.30
HE
GM
EX
SA
LV
MP
MG
RS
ZM
ER
RE
GD
HC
CL
CG
AU
DR
SG
PL
ME
Select Menu#
Select Menu#
Select Menu#
Select Menu#
Select Menu#
data component : VON MISES
data component : MAXIMUM PRINCIPAL
Using dataset : 22 - LOAD 10, STRESS22
data component : MAXIMUM PRINCIPAL
data component : VON MISES

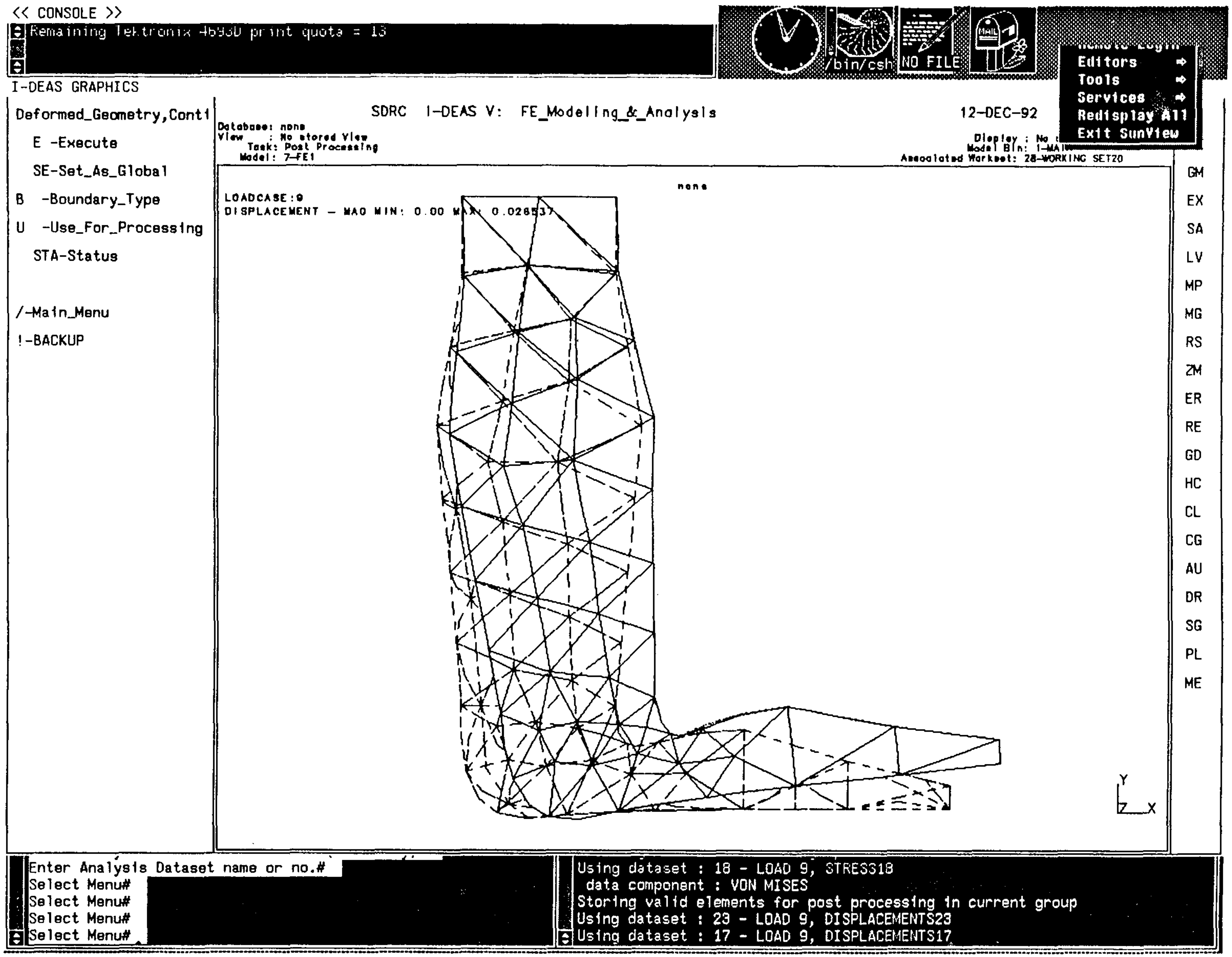
C.6


<< CONSOLE >>
Tektronix 4693D print request queued...
Remaining Tektronix 4693D print quota = 18
/bin/csh
NO FILE
MAIL
I-DEAS GRAPHICS
Contour,Continuous_Tone
E -Execute
SE-Set_As_Global
B -Boundary_Type
U -Use_For_Processing
STA-Status
/-Main_Menu
!-BACKUP
SDRC I-DEAS V: FE_Modeling_&_Analysis
10-DEC-92 15:37:50
Units : SI
Database: none
View : No stored View
Task: Post Processing
Model: 7-FE1
Display : No stored Option
Model Bin: 1-MAIN
Associated Workset: 28-WORKING SET20
LOADCASE:10
FRAME OF REF:GLOBAL
STRESS - MAX PRIN MIN:-2595.69 MAX: 16662.36
16662.36
13911.20
11160.05
8408.91
5657.76
2906.61
155.46
-2595.69
HE
GM
EX
SA
LV
MP
MG
RS
2M
ER
RE
GD
HC
CL
CG
AU
DR
SG
PL
ME
Select Menu#
Select Menu#
Select Menu#
Select Menu#
Select Menu#
data component : MAXIMUM PRINCIPAL
Using dataset : 22 - LOAD 10, STRESS22
data component : MAXIMUM PRINCIPAL
Using dataset : 37 - LOAD 10, STRESS37
data component : MAXIMUM PRINCIPAL

C.7

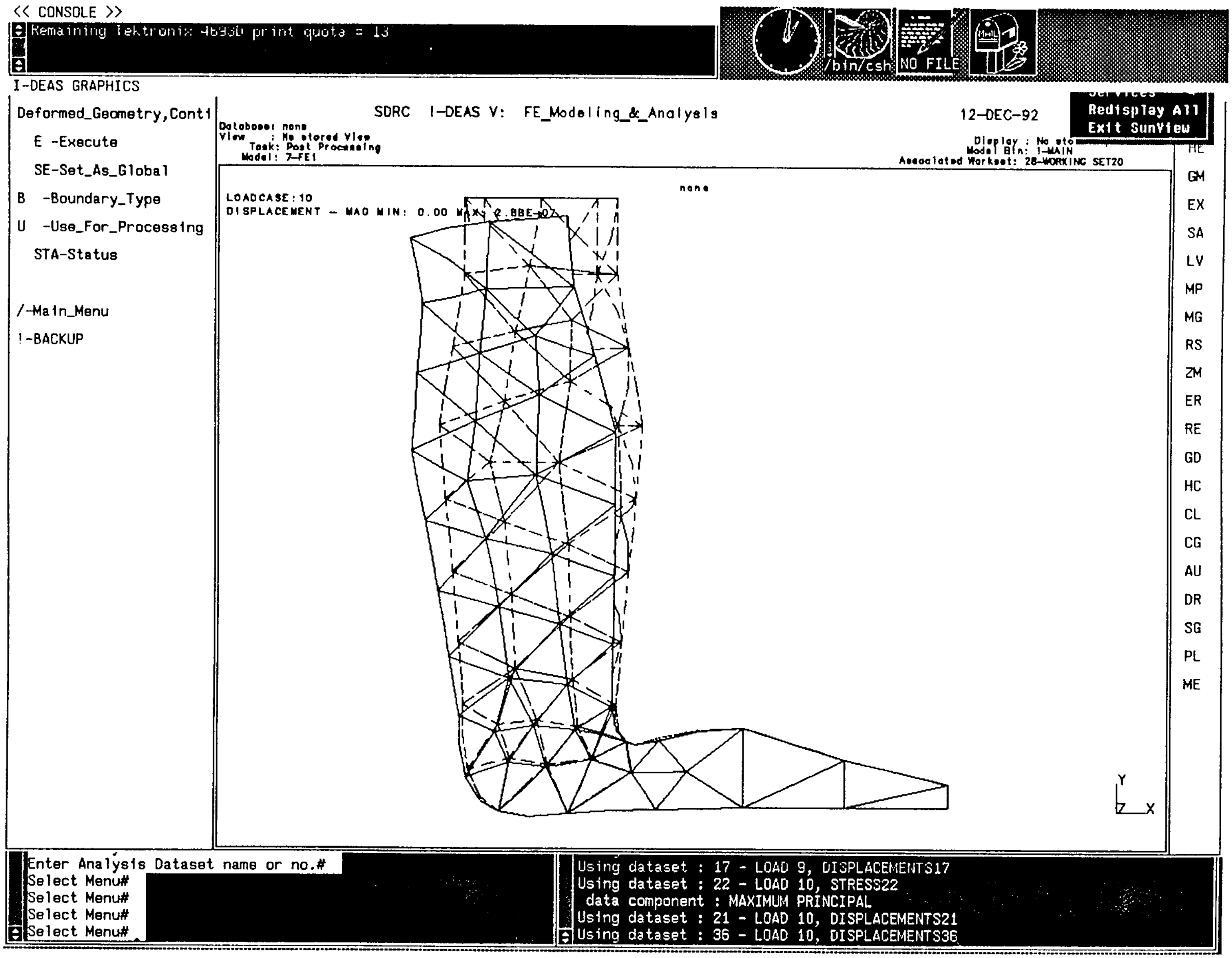

<< CONSOLE >>
Remaining Tektronix 4693D print quota = 18
/bin/csh
NO FILE
MAIL
I-DEAS GRAPHICS
SDRC I-DEAS V: FE_Modeling_&_Analysis
10-DEC-92 15:40:09
Database: none
View : No stored View
Task: Post Processing
Model: 7-FE1
Units : SI
Display : No stored Option
Model Bin: 1-MAIN
Associated Workset: 28-WORKING SET20
LOADCASE 10
FRAME OF REF GLOBAL
STRESS - VON MISES MIN: 7.30 MAX: 26305.82
HE -Help
MF -Model_File
DO -Display_Options
V -View
GR -Group
EL -Elem_For_Calculatio
CS -Coordinate_Systems
A -Analysis_Dataset_Se
AR -Arrow_Plot
CO -Contour
CR -Criterion
X -XYZ_Plot
ET -Element_thickness_p
MAN-Manage_Models
STA-Status
TA -Tasks
/-Main_Menu
!-BACKUP
26305.82
22548.89
18791.96
15035.03
11278.09
7521.16
3764.23
7.30
HE
EX
SA
LV
MP
MG
RS
ZM
ER
RE
GO
HD
CL
CG
AU
DR
SG
PL
ME
Select Menu#
Select Menu#
Select Menu#
Select Menu#
Select Menu#
Using dataset : 22 - LOAD 10, STRESS22
data component : MAXIMUM PRINCIPAL
Using dataset : 37 - LOAD 10, STRESS37
data component : MAXIMUM PRINCIPAL
data component : VON MISES

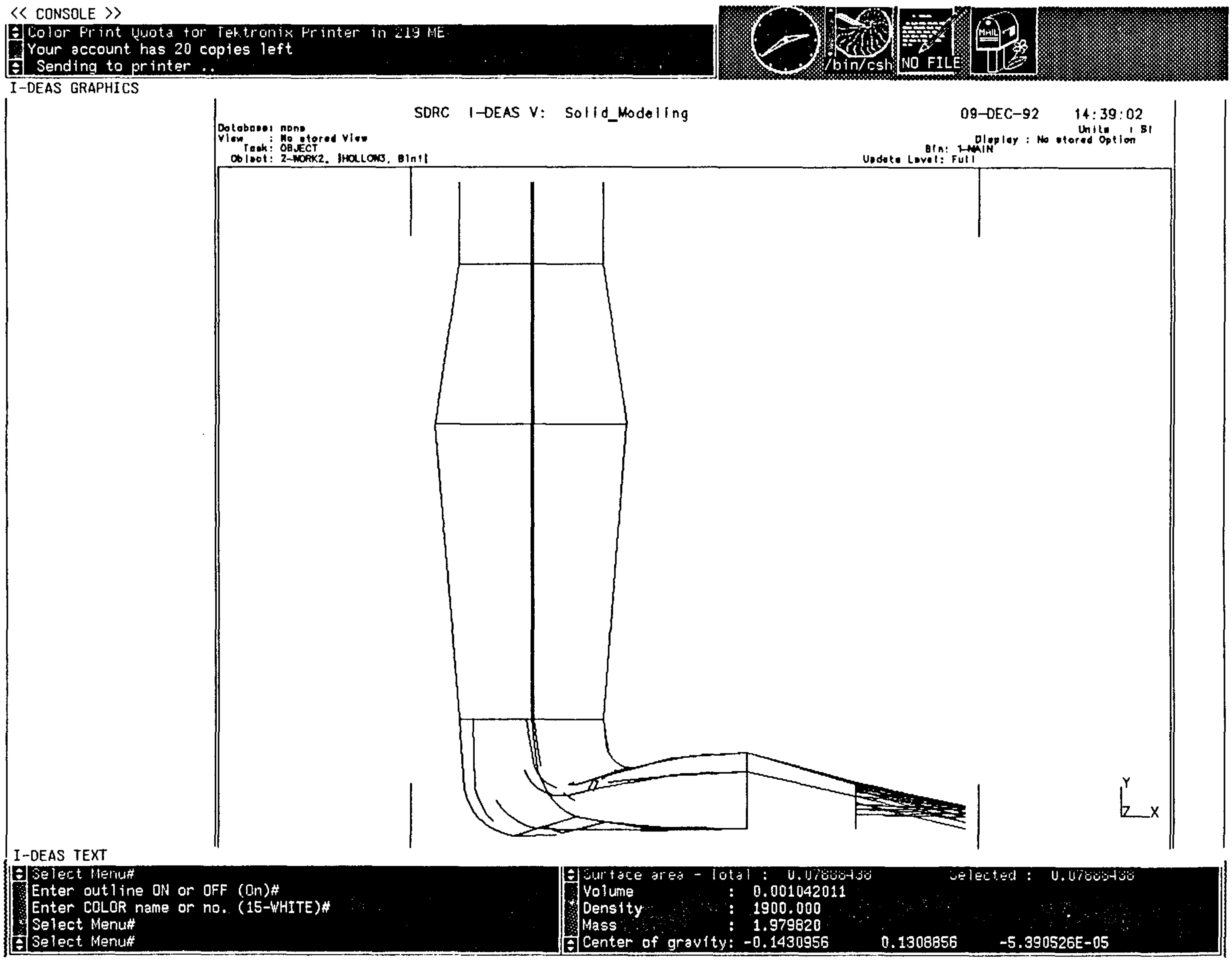
C.8



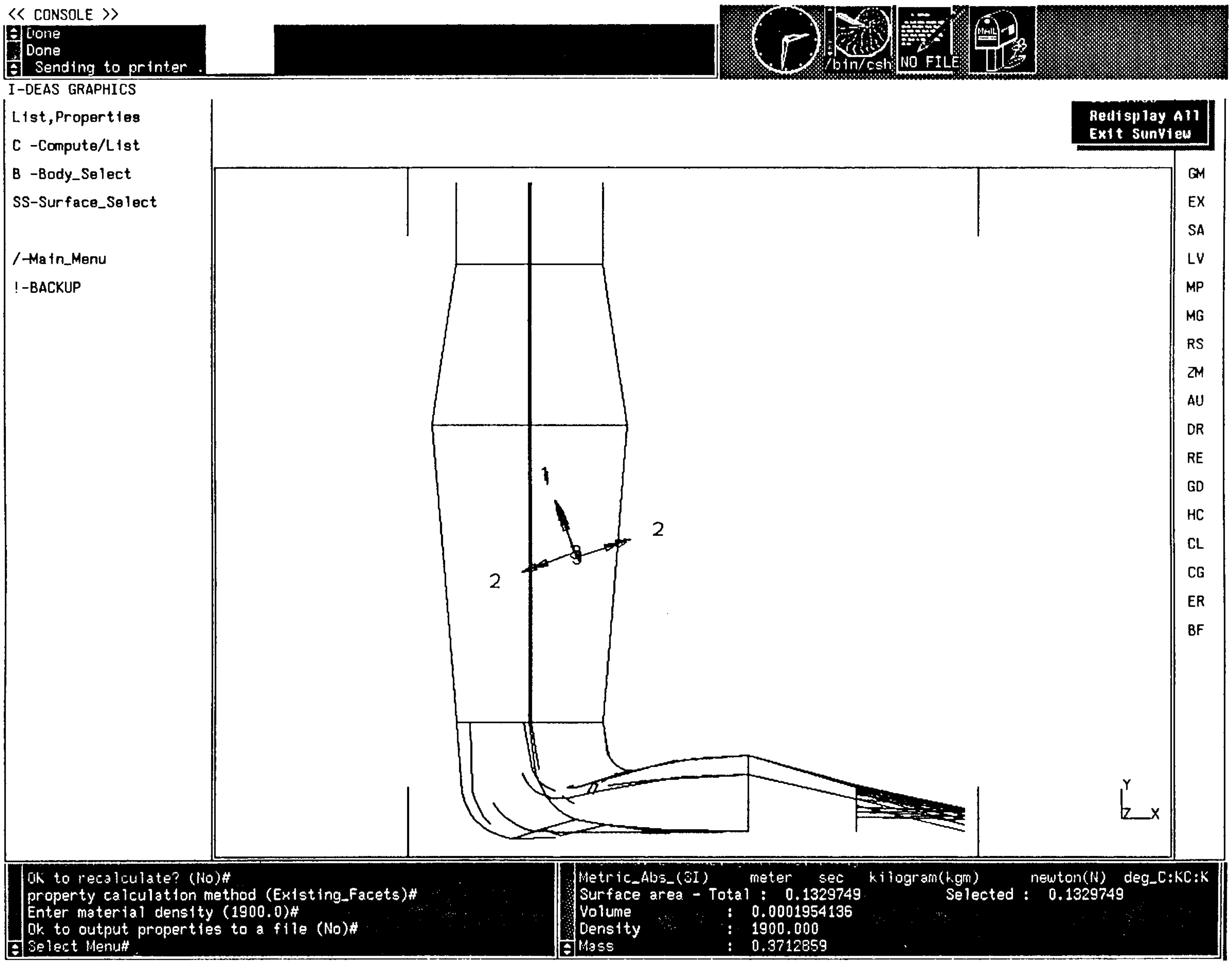
<< CONSOLE >>
Remaining Tektronix 4693D print quota = 13
/bin/csh
NO FILE
Editors
Tools
Services
Redisplay All
Exit SunView
I-DEAS GRAPHICS
Deformed_Geometry,Conti
E -Execute
SE-Set_As_Global
B -Boundary_Type
U -Use_For_Processing
STA-Status
/-Main_Menu
!-BACKUP
SDRC I-DEAS V: FE_Modeling_&_Analysis
12-DEC-92
Database: none
View : No stored View
Task: Post Processing
Model: 7-FE1
Associated Workset: 28-WORKING SET20
none
LOADCASE:9
DISPLACEMENT - MAG MIN: 0.00 MAX: 0.026537
GM
EX
SA
LV
MP
MG
RS
ZM
ER
RE
GD
HC
CL
CG
AU
DR
SG
PL
ME
Y
Z X
Enter Analysis Dataset name or no.#
Select Menu#
Select Menu#
Select Menu#
Select Menu#
Using dataset : 18 - LOAD 9, STRESS18
data component : VON MISES
Storing valid elements for post processing in current group
Using dataset : 23 - LOAD 9, DISPLACEMENTS23
Using dataset : 17 - LOAD 9, DISPLACEMENTS17

C.9


<< CONSOLE >>
Remaining Tektronix 4693D print quota = 13
/bin/csh
NO FILE
I-DEAS GRAPHICS
Deformed_Geometry,Conti
E -Execute
SE-Set_As_Global
B -Boundary_Type
U -Use_For_Processing
STA-Status
/-Main_Menu
!-BACKUP
SDRC I-DEAS V: FE_Modeling_&_Analysis
12-DEC-92
Redisplay All
Exit SunView
Database: none
Task: Post Processing
Model: 7-FE1
Model Bin: 1-MAIN
Associated Workset: 28-WORKING SET20
none
LOADCASE:10
DISPLACEMENT - MAG MIN: 0.00 MAX: 2.88E-07
GM
EX
SA
LV
MP
MG
RS
ZM
ER
RE
GD
HC
CL
CG
AU
DR
SG
PL
ME
Y
Z
X
Enter Analysis Dataset name or no.#
Select Menu#
Select Menu#
Select Menu#
Select Menu#
Using dataset : 17 - LOAD 9, DISPLACEMENTS17
Using dataset : 22 - LOAD 10, STRESS22
data component : MAXIMUM PRINCIPAL
Using dataset : 21 - LOAD 10, DISPLACEMENTS21
Using dataset : 36 - LOAD 10, DISPLACEMENTS36


<< CONSOLE >>
Color Print Quota for Tektronix Printer in 219 ME
Your account has 20 copies left
Sending to printer ..
/bin/csh
NO FILE
I-DEAS GRAPHICS
SDRC I-DEAS V: Solid_Modeling
09-DEC-92 14:39:02
Units : SI
Display : No stored Option
Bin: 1-MAIN
Update Level: Full
I-DEAS TEXT
Select Menu#
Enter outline ON or OFF (On)#
Enter COLOR name or no. (15-WHITE)#
Select Menu#
Select Menu#
Volume : 0.001042011
Density : 1900.000
Mass : 1.979820
Center of gravity: -0.1430956 0.1308856 -5.390526E-05

C.11



<< CONSOLE >>
Done
Done
Sending to printer .
/bin/csh
NO FILE
I-DEAS GRAPHICS
List,Properties
C -Compute/List
B -Body_Select
SS-Surface_Select
/-Main_Menu
!-BACKUP
Redisplay All
Exit SunView
GM
EX
SA
LV
MP
MG
RS
ZM
AU
DR
RE
GD
HC
CL
CG
ER
BF
1
2
2
Y
Z X
OK to recalculate? (No)#
property calculation method (Existing_Facets)#
Enter material density (1900.0)#
Ok to output properties to a file (No)#
Select Menu#
Metric_Abs_(SI) meter sec kilogram(kgm) newton(N) deg_C:KC:K
Surface area - Total : 0.1329749 Selected : 0.1329749
Volume : 0.0001954136
Density : 1900.000
Mass : 0.3712859

C.12

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