

OPTIMAL DESIGN OF COMPOSITE HIP IMPLANTS
USING NASA TECHNOLOGY

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N93-22183
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ABSTRACT

Using an adaptation of NASA software, we have investigated the use of numerical optimization techniques for the shape and material optimization of fiber composite hip implants. The original NASA in-house codes, were originally developed for the optimization of aerospace structures. The adapted code, which was called OPORIM, couples numerical optimization algorithms with finite element analysis and composite laminate theory to perform design optimization using both shape and material design variables.

The external and internal geometry of the implant and the surrounding bone is described with quintic spline curves. This geometric representation is then used to create an equivalent 2-D finite element model of the structure. Using laminate theory and the 3-D geometric information, equivalent stiffnesses are generated for each element of the 2-D finite element model, so that the 3-D stiffness of the structure can be approximated. The geometric information to construct the model of the femur was obtained from a CT scan. A variety of test cases were examined, incorporating several implant constructions and design variable sets.

Typically the code was able to produce optimized shape and/or material parameters which substantially reduced stress concentrations in the bone adjacent to the implant. The results indicate that this technology can provide meaningful insight into the design of fiber composite hip implants.

INTRODUCTION

In orthopaedics, one of the most successful and widely used procedures to treat joint disease is total joint replacement (TJR). Total joint replacement involves the use of prosthetic components to replace the biological joint surfaces. The objective of total joint replacement is to provide an artificial joint that is capable of reproducing "normal" joint kinematics, and that is able to withstand the stresses induced by everyday activities for as long as possible. Although there exists today a plethora of orthopaedic implants that are capable of dramatically improving the performance of pathological joints, their service life is limited. Joint replacement procedures are now being performed on young, active patients. It is evident that the service lives of implants must be extended in order to prevent implant failures and revision surgeries.

One of the most common modes of failure of orthopaedic implants, especially hip and knee prostheses, is loosening of the components. The mechanisms by which this loosening takes place are complex, and most likely involve the interface between the bone and the implant. Clinical studies have suggested that interfacial failure is one of the most common initiators of aseptic loosening in total hip replacements [9,17]. Studies such as these have prompted many investigators and designers of orthopaedic implants to concentrate on minimizing the stresses and stress concentrations in the interface, since these high stresses could ultimately lead to component loosening. Critical regions with regard to implant loosening are the bone adjacent to the implant and the bone/implant interface itself.

Another factor that has been proven to contribute to aseptic loosening of orthopaedic implants is the mismatch between the stiffness of cortical bone and the stiffnesses of traditional implant materials. In the load sharing that takes place between implant and bone, the stress carried by each material is proportional to its stiffness [7]. As a result, an overly stiff implant tends to carry nearly all of the stress in the bone/implant composite structure. This phenomenon is known as stress shielding. The response of bone to changes in stress, as predicted by Wolff's hypothesis is known as remodeling [21]. In many cases, when bone is insufficiently loaded, its adaptive remodeling leads to atrophy with subsequent thinning and increased porosity [13]. Bone degradation at the bone/implant interface usually results in aseptic loosening and implant failure.

Attempts have been made in recent years to reduce the effects of stress shielding by creating new, low stiffness materials for orthopaedic applications. Fiber composite materials seem to have great potential because of their high strength, their biocompatibility, and because of the possibility of tailoring their material properties. In theory, fiber composite implants can be manufactured that meet the demanding strength requirements while closely matching the stiffness of the adjacent bone. Implants such as these would minimize stress shielding effects, which may help prevent implant failure due to aseptic loosening.

Designers of fiber composite implants are faced with a virtually infinite number of combinations of shapes and material properties that can be used for a given implant design. Therefore, the process of searching for optimum choices of shapes and material properties can be both complex and time-consuming. Work by several authors has demonstrated the utility of numerical optimization techniques for the solution of similar problems [8,11,12,15,23]. Some investigators have incorporated Finite Element algorithms into iterative numerical optimization schemes, and attempted to optimize the composite structure of total joint replacements for various parameters [8,11,12,23]. Other investigators have sought to optimize the material properties of orthopaedic implants [22]. The design variables included the Young's moduli of the implant and the cement layer. Good agreement was found to exist between the stresses predicted by the design sensitivity analysis and those obtained from the finite element model [22].

In previous work, our project has adapted computational procedures for shape and material tailoring of aerospace structures to the shape optimization of a total knee and/or hip replacement (TKR) or (THR) component [2,16]. The computational procedures were originally developed in NASA programs for composites analysis and structural optimization [3,4,5]. In the present work, we have extended the procedures to include both shape and material optimization. Here we describe the application of the procedure to a femoral component of a total hip replacement.

METHODS

The OPORIM program has evolved from an analysis code called STAT (Structural Tailoring of Advanced Turboprops), which was originally developed by Pratt & Whitney under a NASA contract [3,4,5]. OPORIM was adapted and enhanced to perform shape and material optimization of fiber composite hip endoprostheses.

The basic structure of the OPORIM program is shown in Fig. 1. Information supplied by the user includes geometric information, material properties, choice of design variables, and information related to the optimization scheme. First, a three-dimensional geometric model is constructed. Then, a dimensionless mesh of gridpoints is created and mapped onto the midplane of the 3-D model. The gridpoints are then used to construct a 2-D finite element mesh. Using laminate theory and the 3-D geometric information, equivalent stiffnesses are generated for each element of the 2-D finite element mesh. An objective function is formulated, usually based on some stress criterion. External loads and boundary conditions are applied to the mesh, and a finite element analysis is performed. The optimizer then changes the design variables in attempt to minimize the objective function. A remesh scheme is performed in response to the change in design variables, and another F.E.A. analysis is conducted. Design variable changes and F.E.A. analyses are performed until convergence on an optimal design has been achieved.

Geometry generation within OPORIM is accomplished by means of several design curves. Some of

the design curves describe the external geometry of the model, and some describe the internal geometry of each material region within the model. The discrete geometric information contained in the input file is interpolated to produce piecewise quintic polynomial splines. The spline curves are used in the design optimization process, and are updated as design changes are made to the model [3].

The composite analysis section of OPORIM is based upon ICAN (Integrated Composite Analyzer), developed at NASA Lewis Research Center's Structural Mechanics Branch [14]. ICAN was designed for the analysis of multilayered fiber composites using micromechanics equations and laminate theory. Laminate theory provides OPORIM with the capability to represent complex composite 3-D structures with a relatively simple mesh of 2-D finite elements. In addition, it allows for the specification of composite properties as design variables in the optimization routine.

Each element of the finite element mesh can be thought of as a laminated plate, composed of material layers through the thickness of the plate. Each material layer is composed of a number of plies, whose thicknesses can be controlled by the input file of OPORIM. Using laminate theory, an equivalent stiffness can be calculated for the laminated plate to create a 2-D plane stress finite element [1,2].

OPORIM uses the relationships derived for composite plate stiffnesses to construct element stiffness matrices based on the three-dimensional geometry (thicknesses), and the material properties of each material region. Once a finite element solution has been obtained, the strains in the individual plies are back-calculated. From these, the laminate model is used to calculate the individual ply stresses [19].

The finite element analysis in OPORIM is based on the commercial code NASTRAN. OPORIM supports elements that are very similar to the NASTRAN TRIA3 element, a 3-node combined membrane-bending triangular plate element. Each node has six degrees of freedom.

The computational efficiency of the 2-D element makes it better suited to optimization algorithms than other more computationally costly elements, such as three-dimensional elements. To confirm the validity of the optimization results, it was necessary to compare the optimal 2-D designs to equivalent 3-D finite element models. This insured that design improvements were actually made.

The optimization module of OPORIM is ADS, a commercial optimization code written by Dr. G.N. Vanderplaats [18]. ADS separates the solution of the problem into three basic levels: the optimization strategy, the optimizer, and the one-dimensional search. Several choices of specific algorithms for the iterative optimization procedure are available in the code.

APPLICATION

The OPORIM code was adapted to perform shape and material optimization of a total hip prosthesis. Reference [1] covers in detail the changes that were made to the OPORIM source code, and the additional code that was developed for these studies.

A typical manufactured total hip prosthesis is illustrated in Fig. 2. It consists of an acetabular component (A), a femoral component (B) and an acetabular insert (C). The models that were considered in this study consisted of the femoral component of the prosthesis and the proximal eight inches of the femur. As a starting point for the model, a CT scan of the proximal twelve inches of a femur was obtained, with cross sections taken at every 5 mm. Measurements were taken from the CT scan of the width of the femur (in the coronal plane), the thickness of the femur (in the sagittal plane), and the thickness of the cortical shell.

The material properties for the bone of the proximal femur were taken from the literature [6,20,24]. All bone considered in this work was assumed to be isotropic. The proximal femur was broken down into three material regions: a cortical shell region, a high-density cancellous region, and a low-density cancellous region.

Two different implant materials were considered in these studies: a titanium alloy (Ti-6Al-4V), and carbon fiber-reinforced polyetheretherketone (PEEK) composite. The properties of titanium, carbon fiber and PEEK were obtained from available literature. As the shape of the implant is changed during the optimization, it becomes necessary to move the nodal lines to accurately represent the new shape of the

implant. This is done by the remeshing scheme in OPORIM, which locates the design curves (splines) that define the edges of the implant, and which moves the nodal lines so that the shape of the implant and the core are defined. In addition, the remesh scheme redistributes the rest of the nodal lines to maintain favorable aspect ratios of the mesh.

The objective function for the design optimization was chosen to be a measure of the stresses in the region of the bone near the implant interface. This reflects the conviction that controlling interface stresses is one of the more important issues with regard to bone adaptation and passive loosening.

Several detailed objective functions were considered including the maximum Von Mises equivalent stresses in the bone elements adjacent to the implant. Others included the sum of the squares of the Von Mises stresses and the sum of the squares of the maximum shear stresses in these same elements.

The same loading conditions were applied to all of the models for all of the objective functions considered. The nodes at the distal (lower) end of the models were fixed.

One of the loading conditions was a force couple producing a bending moment of 1 lb-in applied to two nodes on the end of the "neck" portion of the implant. This loading scheme certainly does not accurately represent physiological loading. However, pure bending is attractive as a test case, since it does not depend on the orientation or the point of application of the load [10]. Results for this loading condition are described below.

RESULTS

Numerous results were generated in the study. To illustrate typical results, we cite the following 2 cases. For more information and additional results, the reader is referred to references [1] and [2].

Case 1

Case 1 involved shape optimization of an implant composed of one material region. The implant was constructed entirely of titanium.

The initial and optimum shapes of the implant in Case 1 are shown in Fig. 3. The optimization produced a proximal widening of the implant and a slight amount of distal narrowing. The initial and optimal distributions of von Mises stress along the normalized length of the implant are shown in Figures 4a and 4b. Note that the solid and dashed lines represent stresses along the lateral and medial edges of the implant, respectively. The stresses are presented in p.s.i., and appear to be extremely low, but the applied loading was a unit bending moment of 1 lb-in, and these stresses can be scaled up for higher applied loads. The peak cancellous von Mises stresses in the initial model were reduced by 77%, and the objective function was reduced by 65%.

Case 2

Case 2 involved shape optimization of an implant with two material regions. The outer region was composed of carbon fiber reinforced PEEK and the inner "core" region was composed of titanium. The design variables included width variables of both the inner and outer regions.

The optimization produced proximal widening in the carbon fiber/PEEK outer region. The optimization of the inner titanium region produced proximal widening, slightly distal to where the titanium region intersects the "neck" of the implant. A slight amount of distal narrowing was observed. The optimization reduced the peak cancellous von Mises stresses by 65% and the objective function by 64%.

DISCUSSION

All of the optimization cases produced significant reductions of the peak cancellous bone stresses in the vicinity of the bone/implant interface. The magnitudes of these reductions ranged between 45% and 84%. In addition, all of the cases resulted in a more uniform transfer of load at the interface, as illustrated by the

plots of stress vs. length.

For all of the cases subjected to the same loading, certain similarities in the stress distribution were observed regardless of the implant construction. The peak stresses in the initial models always occurred in elements at the proximal end of the implants. These stresses seemed to dominate the optimizations, since they contributed the most to the objective functions. In every case, the high stresses at the proximal end of the implant were reduced so that the proximal stresses were closer in magnitude to the more distal stresses.

As expected, the choice of objective function had an impact on the outcome of the optimization cases. In some cases, dramatic differences in the optimal shape and material properties were observed when a different objective function was used for the same test case. In spite of the different outcomes, all of the objective functions produced comparable reductions in peak stresses. There were certain trends, however, that seemed to occur regardless of the type of objective function that was used. For instance, each one of the shape optimizations resulted in some type of proximal widening of the implant. The shapes that were obtained were similar to those of typical currently manufactured implants, which have fairly massive proximal ends.

Another trend seemed to occur when material optimization was performed on implants constructed of an outer and inner region of fiber composite. In all of these cases, the optimal fiber/volume ratio of the outer region was greater than that of the inner region.

The most dramatic results were always obtained when shape optimization was combined with material optimization. If shape or material optimization was performed alone, the reductions in peak stresses were always less than in the combined optimizations. These results are logical, since the additional design variables in the combined optimization increased the size of the design space by adding new ways for the designs to change.

The optimization results demonstrate that the choice of objective function and loading scheme significantly affected the outcome of the optimizations. However, certain characteristics, such as stiff outer material regions and proximal to distal tapers, were produced for all of the cases, no matter which loading scheme or objective function was used. Although the results of optimization studies such as these are not comprehensive enough to be used as the basis for implant designs, they can provide some meaningful information that can help better define the sensitivities of implant designs to various parameters. As computational capabilities improve in years to come, more realistic anatomical models incorporating interface behavior and bone remodeling will be possible. In closing, at its present state the developed software may be a significant computer aided design tool for the improvement and possible customization of orthopaedic implants.

ACKNOWLEDGEMENT

This research was sponsored by the Technology Utilization Office at the NASA Lewis Research Center, Cleveland, OH under NASA Grant NAG 3-1027.

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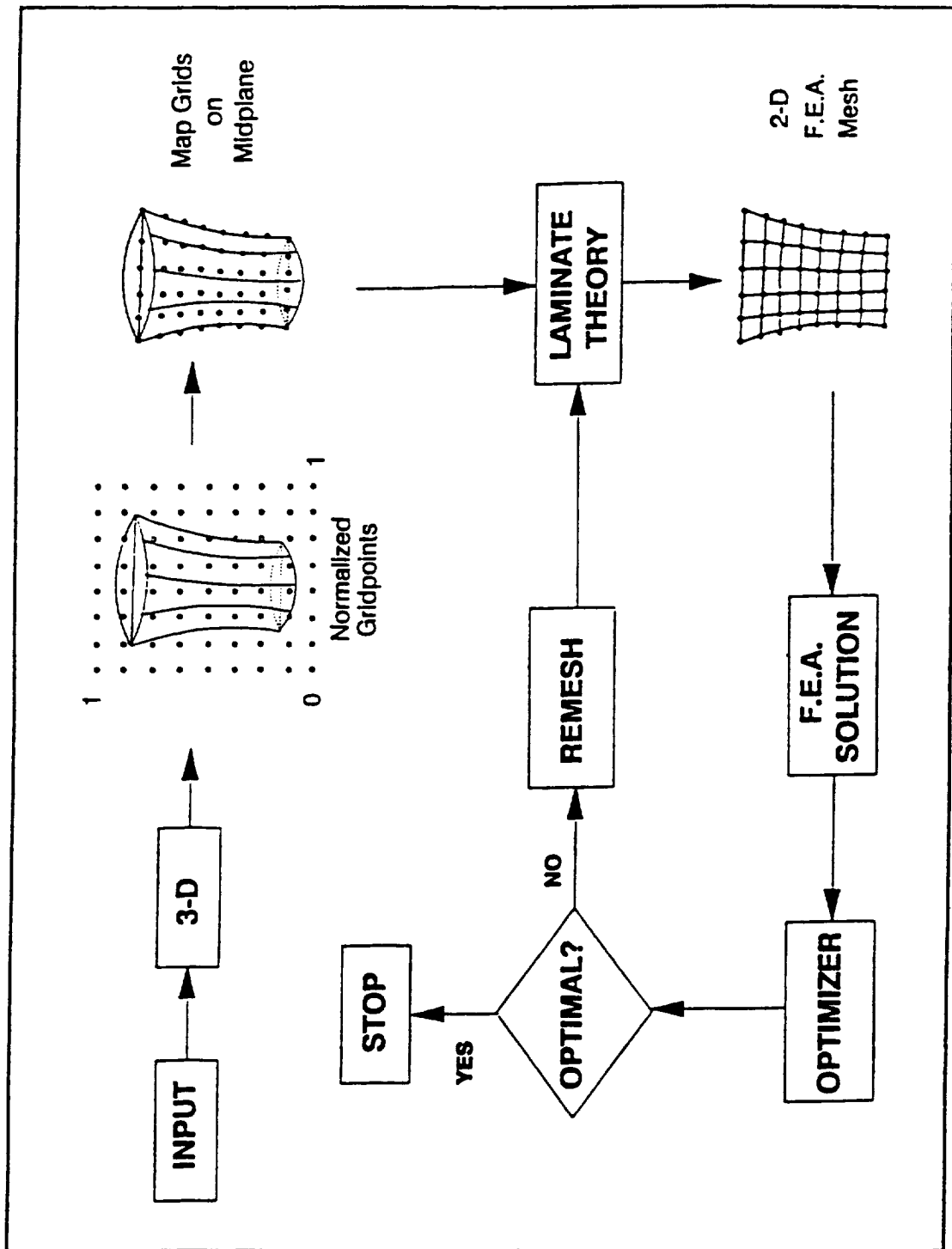


Figure 1: Flowchart of OPORIM

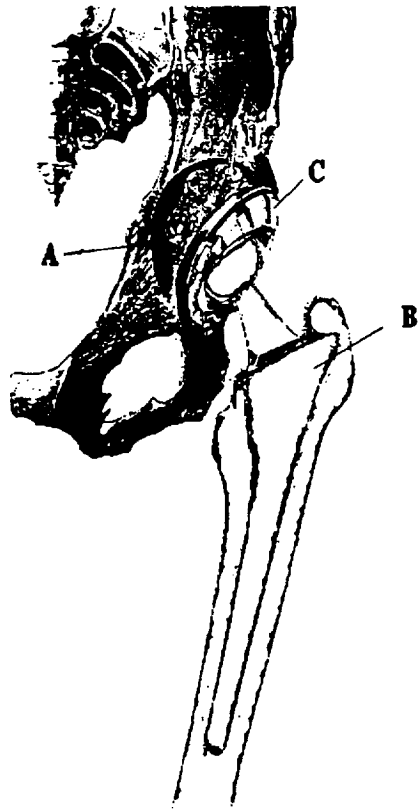


Figure 2: Typical Total Hip Replacement Joint

Initial

Optimal

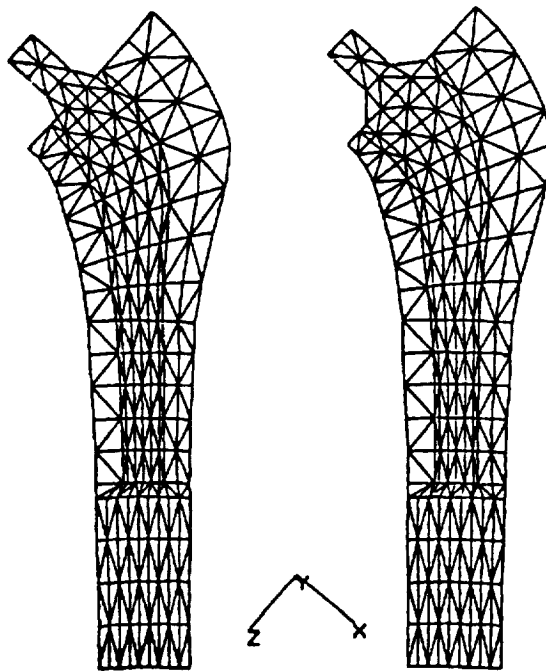


Figure 3: Initial and Optimal Implant Shapes

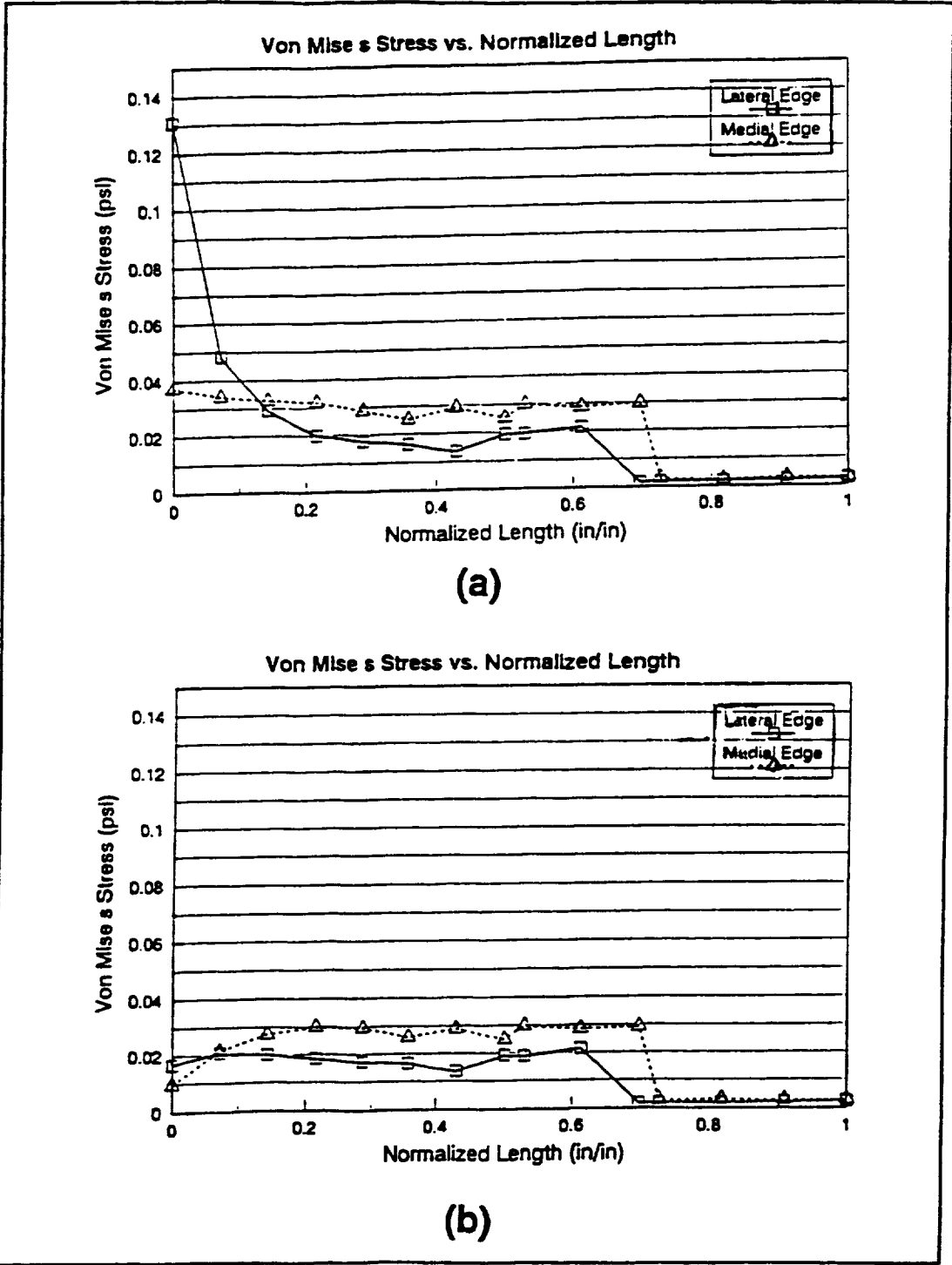


Figure 4: von Mises stresses, Case 1 (a) Initial Design (b) Optimal Design