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Numerical optimization of the design of a coated, cementless hip prosthesis

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A numerical optimization technique was used to improve the design of a coated, cementless hip prosthesis. The prosthesis was represented by a simple one-dimensional finite element model, and its diameter and coating thickness at various points were altered so as to minimize stress shielding while keeping implant–bone interface stresses within realistic limits. The resulting design showed a very large reduction in both stress shielding and interface stresses compared to conventional designs.

1. Introduction

The interposition of a low-modulus interlayer between an intramedullary prosthesis and the surrounding bone can completely change the pattern of load transfer between the prosthesis and the bone [1]. The present work set out to explore the best way of using such an interlayer in the form of a flexible coating [2] around a cementless hip prosthesis.

Numerical optimization techniques can be used to adjust a number of variables so as to optimize a function which is not explicitly known, and have previously been applied to the design of cemented hip prostheses by Huiskes and Boeklagen [3] and by Yoon *et al.* [4]. This technique was used to improve the design of the prosthesis by iteratively altering a simple finite element model.

2. Methods

For simplicity, the prosthesis and the surrounding bone were modelled by a one-dimensional finite element model, using 60 tapered beam elements representing the prosthesis and the cortical bone, coupled together by 60 spring elements representing the stiffness of the coating in the axial and radial directions. The trabecular bone within the upper part of the femur was represented by a further 15 beam elements. The spring elements were coupled to the beam elements using constraint equations so that the deflection of each spring node was equal to the average deflection of the corresponding beam element. The structure of the model is shown schematically in Fig. 1; note that the bone and prosthesis elements have been separated for clarity and that all the elements actually lie in a single line. The model was assumed to be axisymmetric for simplicity, and the axial and normal coating stiffnesses and forces were assumed to be axial and radial regardless of the tapers of the inner surface of the bone and the outer surface of the stem. Geometric and material properties for the bone were estimated from data in the literature (Noble *et al.* [5]); the flexural rigidity of the bone structure agreed well with that found by Bobyn *et al.* [6].

This model is limited by the assumptions of beam theory, most notably that plane sections remain plane during bending. This assumption is reasonable for most of the range of coating thickness studied, but breaks down as the coating thickness approaches zero. The representation of thin coatings is also limited by the mesh density.

The shape of the prosthesis was described by nine variables. The outer diameter of the coating was fixed at 15 mm over the distal and central parts of the stem, while at the proximal end there was a double taper described by two variables. A further seven variables specified the diameter of the core at various points along its length. Since the proximal end of the prosthesis is the most critical part of the design, the core diameter variables were concentrated at this end. Care was taken to ensure that the chosen variables and limits did not allow zero or negative dimensions at any point.

The model was formulated and solved using the ANSYS finite element system, and was then optimized using the optimization routine provided in ANSYS. This routine uses a database of up to 50 previous design sets to find a quadratic approximate function which is then minimized using an unconstrained search technique. The design variable set corresponding to this minimum is then used in the next iteration. Penalty functions are used to represent constraints on the design.

Since there is no explicitly known function in this type of optimization, there is no guarantee of convergence to a global minimum except in trivial cases. During optimization, the only evidence of convergence is provided by the magnitude of the change between successive iterations. The question of convergence will be discussed further in the light of the results of the present study.

The result of the optimization process clearly depends on the objective and constraints chosen for the design. It is important to note that many factors in the design are therefore neglected. Previous numerical optimization studies by Huiskes and Boeklagen [3]

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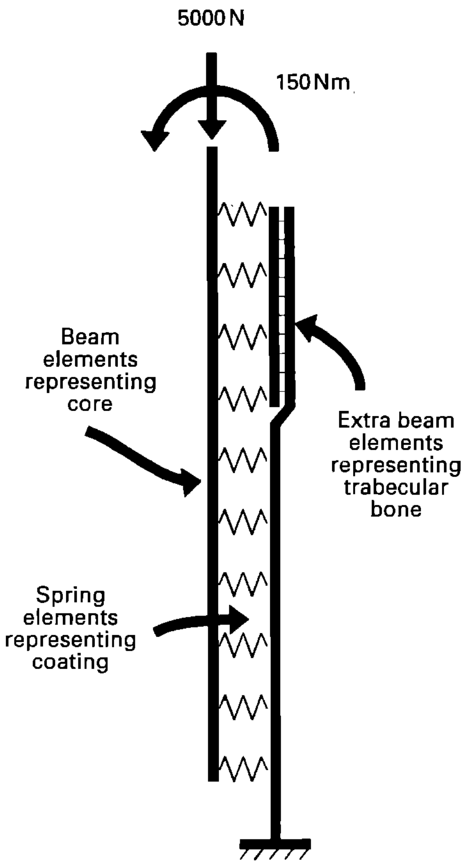

Figure 1 Schematic drawing of the beam-on-elastic foundations finite element model. Note that for clarity only some of the springs are shown.

and Yoon *et al.* [4] of the design of cemented prostheses have aimed to minimize stresses across the implant–bone interface, and have neglected the changes in the loading of the bone as a result of the prosthesis. In the present study, the objective of the optimization process was to minimize the difference in the total strain energy in the bone between the current design and that found in the absence of a prosthesis. This provides a simple method of summing the levels of loading in the bone elements in order to assess the difference between the loading pattern in the implanted femur and that found in the natural femur. Fyhrie and Carter [7] and Huiskes *et al.* [8] have used the strain energy in bone as a regulator of bone density in simulated remodelling studies and found that this produces realistic behaviour; in the present study the total strain energy was used so that the contribution of each element to the objective function was weighted in proportion to its volume.

The loading configuration was chosen to represent a large, but not unusual, load on the hip. An axial load of 5000 N was offset by 30 mm in the medial direction, resulting in an axial load of 5000 N and a bending moment of 150 Nm at the proximal end of the stem. The effect of muscle forces on the proximal femur was neglected; however, this results in a conservative design since the effect of such forces will be to reduce the bending moment in the femur.

The elastic moduli of the coating, bone and core were assumed to be 1, 10 and 100 GPa, respectively, and the Poisson's ratio of all three materials was assumed to be 0.3. A safety factor of four was used in calculating the maximum allowable stress in each component. The maximum stress in the core of the prosthesis was limited to 250 MPa, representing a practical design value for Ti–6Al–4V. The interfacial shear stress between the bone and the prosthesis was limited to 2 MPa, based on push-out strengths of around 8 MPa reported by Boone *et al.* [9] and Zimmerman *et al.* [10] for hydroxyapatite-coated polymers, and the normal interface stress was limited to 20 MPa, based on the compressive strength of cortical bone. Note that the coating was represented only by spring elements coupling the core and the bone, and that its strength and stiffness were neglected. This assumption was thought to be reasonable since the coating was assumed to have a modulus 100 times less than that of the stem, and its stiffness was therefore small.

The optimization procedure has no single starting point as each new design set is based on a database of previous designs. A series of 15 initial design sets were generated representing various design concepts and others with extreme values of the design variables. Since none of these provided a feasible solution, the routine proceeded by generating random design sets until a feasible solution was found.

3. Results

It was found that the optimization process proceeded slowly, and required several hundred iterations to converge. The process was complicated by the strong interdependence of the variables, and the small part of the possible variable range which allowed feasible solutions.

As discussed earlier, convergence cannot be proven in this procedure, and it is almost certain that the final solution does not represent the optimum. However, the final solution shown below represents a substantial improvement over conventional designs within the terms of the optimization study.

Fig. 2 shows the shape arrived at using nine variables in the optimization procedure. The distal part of the core (region A) is rapidly reduced to its minimum thickness, and effectively becomes redundant. The proximal–central part of the core (region B) becomes very thin, but is loaded to the maximum allowable stress. The proximal part of the core (region C) is much thicker, and there is a pronounced proximal taper.

Fig. 3 shows the bending moment distribution along the proximal femur for the optimized design, for a conventional cementless design and for the unimplanted femur. It is clear that the optimized design results in a loading pattern that is much closer to the homeostatic norm than is found with a conventional cementless prosthesis.

Fig. 4 shows the interfacial shear and normal stresses in the optimized design and in a conventional design with a 1 mm coating. The interfacial stresses are reduced to sustainable levels, as a result of improved load transfer. It should be noted that in an uncoated prosthesis the interfacial stresses would be very much larger.

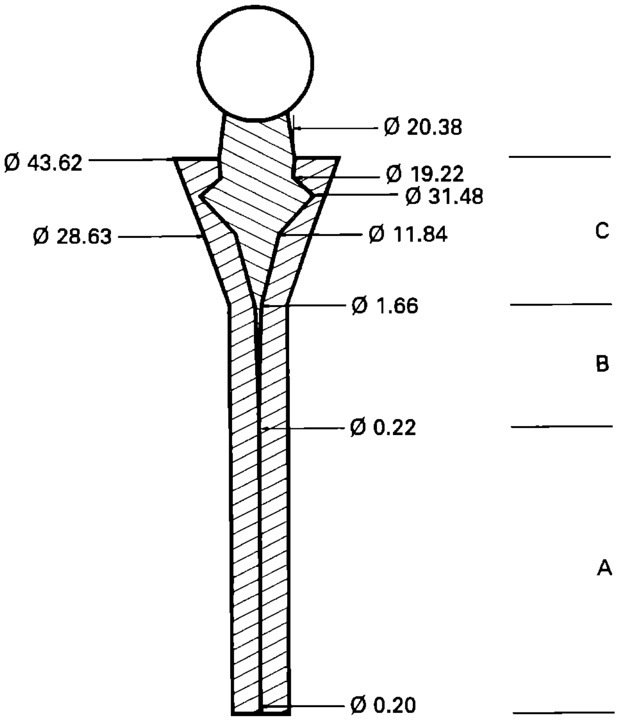

Figure 2 The shape of the prosthesis after optimization, showing the nine optimization variables and their final values in millimetres.

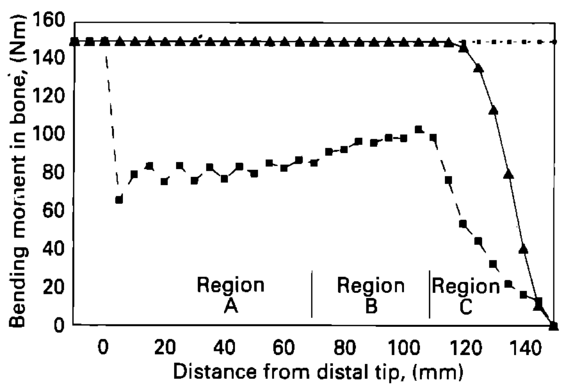

Figure 3 The bending moment distribution in the bone elements along the length of the prosthesis: ▪ bone only; ■ conventional design; ▲ optimal design.

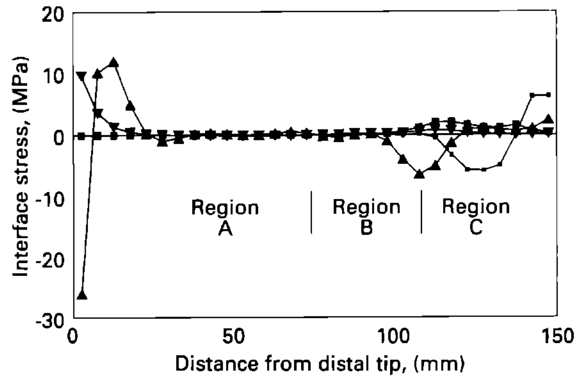

Figure 4 Normal and shear stresses across the implant–bone interface along the lateral edge of the optimal design and a conventional design with a 1 mm thick coating: ▲ normal (1 mm coating); ▼ shear (1 mm coating); ■ Normal (optimum); ◘ shear (optimum).

4. Discussion

The shape of the optimized design may be explained in terms of the load transfer mechanisms described by Huiskes [11]. The prosthesis–bone system may be conceptually divided into three regions, marked A, B and C in Figs 2–4: a proximal load transfer region (C) where some of the load is transferred from the prosthesis to the bone, a central region (B) where the bone and prosthesis act as a composite beam, and a distal load transfer region (A) where the remaining load is transferred from the prosthesis to the bone. In the case of the optimized design, the share of the load taken by the prosthesis in the central region is minimized, and so there is very little load transfer distally.

In region A, a reduction in the stiffness of the prosthesis and an increase in the coating thickness both act to reduce the interfacial stresses and to reduce the stress-shielding effect of the prosthesis. The distal half of the core is thus quickly reduced to its minimum diameter in the optimization procedure. An alternative view of this result is that load transfer should take place as far proximally as possible in order to minimize stress shielding, and so the distal part of the prosthesis becomes redundant and could be removed from the optimized design altogether without significantly affecting the results.

There is little load transfer in region B and it acts largely as a composite beam. The relative proportions of this part of the stem determine the proportion of the load that is transferred in the proximal and distal load transfer regions, and so its dimensions have a sensitive effect on the behaviour of the whole prosthesis. In the optimization study, the core diameter is reduced as far as possible, limited by its strength towards the proximal end and by the minimum allowed diameter distally. This ensures that as much of the load as possible is transferred in region C.

Region C behaves in a more complex manner than region A. Here a decrease in the core diameter has two conflicting effects. A reduction in the stiffness of the core concentrates load transfer proximally, increasing the interface stresses and reducing stress shielding, while an increase in the coating thickness spreads the load transfer region distally, reducing the proximal interface stresses and increasing stress shielding. The optimum shape of this part of the prosthesis is therefore a compromise between these conflicting factors. At the proximal end of the prosthesis, the effect of the coating predominates, and so there is a pronounced proximal taper. Slightly further distally, however, a thicker core is required to transfer the load further down the stem in order to keep the interface stresses within the allowed limits.

It is generally accepted that metallic materials do not have the most appropriate mechanical properties for use in bone replacement, and that the use of prostheses with mechanical properties more closely matched to those of the femur may be beneficial. However, a simple substitution of a lower modulus material will be of little or no benefit. The selection of materials and the mechanical design of the prosthesis are necessarily interrelated and must proceed together. In order to achieve a substantial improvement in mechanical compatibility, the mechanical properties of the prosthesis must be matched to those of the femur, so as to optimize the pattern of load transfer from the prosthesis to the bone. The present work indicates the way in which this may be done, and the possible benefits of this approach.

There are a number of limitations in this simple study. In particular, the strength limits on the design and the stress shielding calculations should be calculated in more detail from different loading configurations. The strength of the interfaces and the components should be based upon the maximum impact loading in a variety of design cases representing situations such as falls where prosthesis damage is likely. On the other hand, bone remodelling is thought to be controlled by the mean peak loading during exercise [12], and so walking or running loads might be more appropriate here. The effect of partial or total debonding of the interfaces should also be considered. Proper representation of these aspects of the design would require a three-dimensional model of much greater complexity than the simple model presented here, and numerical optimization of such a model would require a more efficient optimization algorithm to reduce the amount of computation to more manageable proportions.

Previous numerical optimization studies by Huiskes and Boeklagen [3] and Yoon *et al.* [4] showed that a significant reduction in interface stresses was possible by tailoring the stiffness of the prosthesis to improve load transfer to the bone. However, these studies considered a cemented prosthesis, and were thus of limited practical value because of the difficulty of producing a strong, well-bonded cement layer around the prosthesis, particularly where there was a pronounced proximal taper. Similarly, a simple study by Shirandami and Esat [13] showed that an improvement in the mechanical compatibility of a hip prosthesis could result from the use of an outer layer with elastic properties midway between those of the bone and the core; in the present study, it is shown that a much greater improvement is possible through the use of an outer layer which is much more flexible than the bone or the core. The use of adhesively bonded coatings developed by the authors [2] permits the construction of an implant with any appropriate stiffness characteristics; a structure of the type considered in the present study could therefore provide a practical improvement in cementless implant performance. A practical design would also have to take into account the three-dimensional shape and properties of the bone, the prosthesis and the loading, and would require consideration of many other design features.

This study demonstrates the way in which improved implant performance can be obtained by using structures with functionally gradient properties. A substantial improvement in performance is obtained by varying the stiffness of the prosthesis both through its thickness and along its length. It is suggested that more sophisticated structures could result in further performance improvements.

5. Conclusions

The present study demonstrates that very large improvements in the mechanical performance of hip prostheses are possible by utilizing structures with functionally gradient properties. By tailoring the stiffness of the prosthesis both along its length and through its thickness it is possible to change the pattern of load transfer between the prosthesis and the bone so as to achieve both a reduction in interface stresses to practically sustainable values, and a very substantial reduction in stress shielding.

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