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Citation for published version (APA):
Verdonschot, N. J. J., Dalstra, M., & Huiskes, H. W. J. (1990). The relevance of implant telemetry for mechanical analyses of total hip arthroplasties. In G. Bergmann, F. Graichen, & A. Rohlmann (Eds.), *Implantable telemetry in orthopaedics : workshop : papers* (pp. 249-257). Freie Universität Berlin.

Document status and date:
Published: 01/01/1990

Document Version:
Publisher's PDF, also known as Version of Record (includes final page, issue and volume numbers)

Please check the document version of this publication:

• A submitted manuscript is the version of the article upon submission and before peer-review. There can be important differences between the submitted version and the official published version of record. People interested in the research are advised to contact the author for the final version of the publication, or visit the DOI to the publisher's website.
• The final author version and the galley proof are versions of the publication after peer review.
• The final published version features the final layout of the paper including the volume, issue and page numbers.

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Download date: 04. Oct. 2023

The Relevance of Implant Telemetry for Mechanical Analyses of Total Hip Arthroplasties

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Abstract

Although not primarily developed for use in combination with the finite element method, implant telemetry does provide data which are very suitable for this purpose. In doing so, it is possible to monitor the output variables in time, corresponding to the changes of the joint force. In this study, telemetric data of the hip joint force during normal walking were used as input for a three dimensional finite element model of a non-cemented, HA-coated femoral stem. The stresses in the prosthesis, in the bone and at the interface were studied during one walking cycle. It was found that the variation of the hip joint force had considerable effects on the stresses and it was concluded that such a quasi dynamic analysis gives much better insight in the load-transfer mechanism of a prosthesis/bone system as can be obtained from an analysis with only a single load. It is concluded, that the combination of telemetry and finite element analysis can provide realistic pre-clinical testing methods for prosthetic designs.

Introduction

In recent years, a few studies on in vivo telemetric measurements of hip prosthesis loading have been reported in the literature. These studies have provided valueable information on the magnitude and direction of the hip joint force, values which formerly could only be estimated from mathematical models. Measurements with instrumented implants are performed predominantly with the femoral component of a THA. Examples are the studies of Rydell (1966), English et al. (1979), Davy et al. (1988) and Bergmann et al. (1989). In some cases, the acetabular component is instrumented with probes (Rushfeldt et al.; 1981, Bereiter et al., 1989). Apart from the obvious advantage that telemetry offers the possibility to dispose directly of the values of the hip joint force when analysing the forces around the hip, telemetry also provides essential data for stress analyses of THA. Depending on the objective, finite element stress analyses of THA are either performed with two dimensional (2-D) or 3-D models (Huiskes and

The materials used in the analyses are assumed to behave linear elastic, homogenous and isotropic. The elastic properties are listed below:

Cortical bone	: E= 17.0 GPa
Trabecular bone	: E= 1.0 GPa
Titanium	: E= 100.0 GPa

A Poisson's ratio of 0.3 is taken for all materials. To simulate the bonded interface of the osseous integrated prosthesis, all material interfaces were assumed to be connected in both shear and tension.

As load input for the analysis, telemetric data of the hip joint force during walking, as found by Bergmann et al. (1989), is used. Only one cycle is considered and within this cycle the curves of the three hip joint force components are linearized by taking some characteristic points on each curve and then interpolating linearly between them. In Fig. 2, the three components are shown as function of time. Furthermore, in accordance with the data of Bergmann et al. (1989), a bodyweight of 650 N was used. Although muscle forces do have a considerable influence on the overall stress distribution (Rohlmann et al., 1983), they were not taken into account in this study. At this stage, the objective was not so much to determine the actual values for the stresses, but rather to demonstrate the possibilities of quasi dynamic stress analysis. However, if corresponding dynamic data of muscle forces are available, they can easily be included in the analysis.

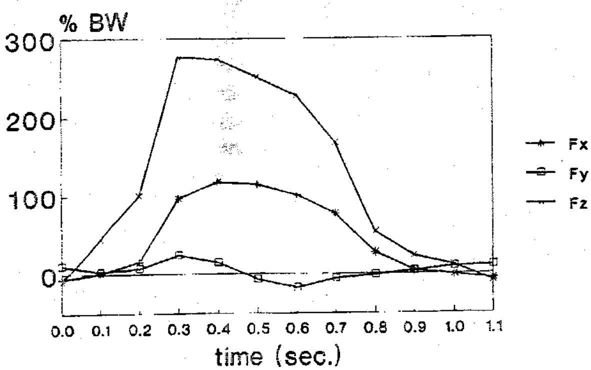

Fig. 2
Hip joint force replacement, one walking cycle. *The three components of the hip joint force during one walking cycle, based on in vivo telemetric data obtained by Bergmann et al. (1989).*

The purpose is now to calculate the stresses at several stages during the walking cycle. Instead of performing finite element calculations for each stage, only three calculations are done, one for each unit hip-force component. The output of these calculations is stored in separate files. The solution for each arbitrarily chosen stage in the gait cycle is then found by superimposing the three separate files, each multiplied with the corresponding magnitude of the force component concerned at that stage. This approach is justified only, when linearity of the model is assumed.

of the hip joint force becomes relatively large, resulting in a more laterally directed resultant force. At the distal side of the prosthesis, the resultant bending moment changes sign, resulting in tensile stresses at the medial side of the stem. The peak stress value of 37 MPa is not reached at 0.4 seconds, when the hip joint force reaches its maximal value of 1,900 N, but at 0.5 seconds, when the ratio Fx/Fz is slightly higher, although the magnitude of the hip joint force has already decreased again to 1,750 N. At the same time on the lateral side, the prosthetic bending stresses shows a compressive peak of 53 MPa. These results suggest that the medial/lateral x-component of the hip joint force has a strong influence on the bending stresses at the distal side of the prosthesis. This becomes clear, when we look at the bending stresses due to the unit loads. In **Fig. 4**, the curves of the bending stresses due to a unit load in the medial/lateral x-direction and a unit load in the axial z-direction, are shown. The bending moment in the prosthesis, due to **Fz**, is more or less constant, but the bending moment due to Fx varies along the length of the prosthesis, because of the increasing length of the moment arm. As is to be expected, the signs of the bending stresses are opposite to one another, and furthermore, peak values due to a unit Fx are about three times as high as the ones due to a unit Fz. This means that the medial/lateral x-component of the hip joint force does not need to be as large as the axial z-component to produce stresses of the same order of magnitude.

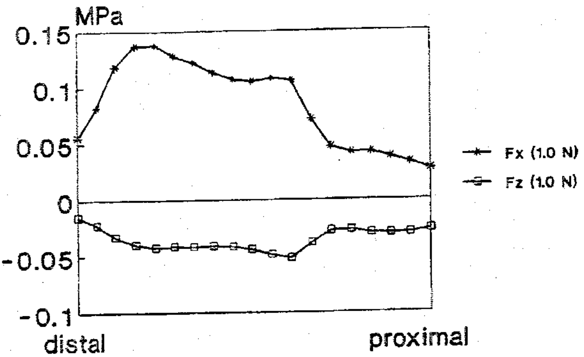

Fig. 4
Bending stresses, prosthesis, medial side.
The prosthetic bending stresses along the medial surface for unit loads in the x- and z-directions.

When interface stresses are considered, three important stress components can be distinguished. One normal stress component and two shear stress components (a longitudinal and a transversal shear stress component). **Figs. 5 a, b and c** show these interface stresses along the medial interface in the time interval from 0.0 to 0.5 seconds. Stress concentrations are found proximally and near the distal tip, indicating load-transfer in these regions predominantly. For both the interface normal stresses and longitudinal shear stresses a maximum value of about 0.9 MPa is found. Laterally, higher interface stresses are generated (maximal values are 5 MPa and 11 MPa for normal stress and longitudinal shear stress, respectively). Proximally, the normal and longitudinal shear stress curves show a maximum at 0.3 seconds, whereas distally maximal values are generated at 0.5 seconds. As stated above, this can be explained by the fact that the medial/lateral Fx component of the hip joint force gains influence more distally, while proximally the axial hip joint force is of more relative importance.

The transversal shear stresses are significantly lower than the longitudinal shear stresses (maximum values are about 0.2 MPa at the medial, and 1.8 MPa at the lateral interface). Obviously, the relatively small value of the anterior/posterior Fy component of the force plays an important role in this phenomenon. When **Figs. 2 and 5 c** are considered, one can see that the transversal shear stress is directly related to the sign and the magnitude of Fy. As the sign of Fy changes between 0.4 and 0.5 seconds, the transversal shear stresses do as well. As function of time, the Fy component of the hip joint force shows symmetric values with respect to the Fy=0.0 level. This results in transversal shear stresses which show the same symmetry, as can be seen in **Fig. 5 c**.

Discussion

In this study, telemetric results of a walking cycle have been applied to a 3-D finite element model of a prosthesis/bone system. It appears that both the magnitude and the direction of the hip joint force play an important role in the stress patterns, generated in the model.

Maximal bending stresses in the prosthesis (tensile at the medial side and compressive at the lateral side), generated during one walking cycle were found to be not only dependent on the magnitude of the hip joint force, but to have a strong relation with its direction as well. With increasing lateralisation of the hip joint force, higher bending stresses are found distally. In fact, peak values were not found when the hip joint force reached its maximal value of 1,900 N at an angle of 23 deg. relative to the longitudinal axis, but at an agle of 25 deg., when the hip joint force had already decreased again to 1,750 N. In standard fatigue tests of femoral stems, a load of 3,000 N at an angle of 10 deg. relative to the longitudinal axis is used (ISO/DIS 7206/3). We applied this load to our model and compared the prosthetic bending stresses to the ones generated during walking. In **Fig. 6**, the envelopes of the prosthetic bending stresses during walking are shown together with the stress curves along the stem due to the fatigue-testing load. Although the load magnitude for the fatigue test is more than 50% higher than the maximal force during walking, the actual distal stresses during walking are higher, particularly on the lateral side. This suggests a more lateralised load to be included in fatigue testing. However, for a realistic comparison, muscle forces should be taken into account as well, as they tend to decrease the distal stress peaks (Rohlmann et al., 1983).

The lateralisation of the hip joint force shows similar phenomena for the interface stresses as described above for the prosthetic bending stresses. Proximally, peak values occur at 0.3 seconds, when the axial component of the hip joint force reaches its maximal value, whereas distally, the peak values occur at 0.5 seconds, when the hip joint force is slightly lower, but more laterally directed.

Although muscle forces still have to be estimated by other experiments or models, telemetric measurements will attribute significantly to more realistic stress analyses of THA. With the data supplied by telemetric experiments, further insight in the load-transfer mechanism of a prosthesis/bone system during daily actvities is obtained, which will surely give more accurate guidelines for prosthetic design. As it has become clear to what kind of loading-cycle the hip joint is exposed, it should be possible to develop more relevant fatique tests and combine these tests with FE models to improve the relevance of FE calculations and finally develop adequate pre-clinical testing methods for prosthetic designs.

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