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The fixation of the cemented femoral component

EFFECTS OF STEM STIFFNESS, CEMENT THICKNESS AND ROUGHNESS OF THE CEMENT-BONE SURFACE

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After cemented total hip arthroplasty (THA) there may be failure at either the cement-stem or the cement-bone interface. This results from the occurrence of abnormally high shear and compressive stresses within the cement and excessive relative micromovement.

We therefore evaluated micromovement and stress at the cement-bone and cement-stem interfaces for a titanium and a chromium-cobalt stem. The behaviour of both implants was similar and no substantial differences were found in the size and distribution of micromovement on either interface with respect to the stiffness of the stem.

Micromovement was minimal with a cement mantle 3 to 4 mm thick but then increased with greater thickness of the cement. Abnormally high micromovement occurred when the cement was thinner than 2 mm and the stem was made of titanium.

The relative decrease in surface roughness augmented slipping but decreased debonding at the cement-bone interface. Shear stress at this site did not vary significantly for the different coefficients of cement-bone friction while compressive and hoop stresses within the cement increased slightly.

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After cemented total hip arthroplasty (THA) several types of movement may occur in the frontal plane in femoral components which have failed.[1] These include metal pistoning within the cement and/or cement pistoning within the femur, medial migration of the proximal part of the stem and lateral migration of the distal part, calcar pivot (with a mediolateral toggle of the distal end of the stem because of lack of cement), and bending cantilever fatigue due to medial migration of the proximal part of the stem and good fixation of the distal end in the cement. In addition, there may be torsional movement because of excessive axial torque. These movements may be associated with failure of the cement-stem interface as a result of debonding of the stem from the cement[2,3] and of the cement-bone interface because of the breakdown of cement-bone interdigitation.[4-6] These may result from the occurrence of abnormally high shear and compressive stresses within the cement and excessive debonding and slipping at both cement-stem and cement-bone interfaces.[7-9] High compressive stress can cause fracture of the cement and subsequent subsidence of the stem[10] while excessive slipping, resulting from pistoning and torsion of the stem, enhances the protrusion of cement debris.[11] Slipping at the cement-bone interface may lead to necrosis of bone which has interdigitised with the cement,[5] while debonding at the stem-cement interface may result in loosening[2] of the stem and periprosthetic osteolysis.[12,13]

Despite the importance of both types of the failure of the interface, most recent studies[3,14-19] have focused either on a fully-bonded interface or on a fully-bonded cement-bone interface with a frictional cement-stem contact. One of the mechanical factors affecting the stress and micromovement at the interface is the coefficient of friction at both the cement-stem and cement-bone interfaces. Improving the bond at the cement-stem interface has been found to increase stress at the cement-bone interface.[7,18,20] Analysis of the anchorage of the stem accounting for discontinuity at both cement-stem and cement-bone interfaces, seems to have been less well studied. The coupling effects of the roughness at the cement-stem and cement-bone interfaces have not been investigated.

Many factors have been implicated in the distribution of stress and micromovement. Experimental tests have shown the influence of the stiffness of the stem on cement stress[21,22] and numerical studies have indicated that stiffer stems induce higher distal stress[23-25] and unload the proximal femoral bone, leading to stress shielding.[22,26] Conversely, in vitro[27] and in vivo,[28] studies have concluded

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Table I. Applied forces of the femoral head and muscles

	F_x[N]	F_y[N]	F_z[N]
Femoral head	-320	448	-1820
Abductors	430	0	1160
Iliopsoas	75	525	560

that stems with lower stiffness are susceptible to wear, which may lead to the generation of particulate metallic debris.[28-30] The influence of material stiffness on long-term behaviour remains unclear. Clinical[31,32] and experimental[33] studies have shown that the thickness of the cement has a strong influence on the stress magnitude.[21] However, conclusions regarding the optimal range of thickness to ensure an excellent long-term result are contradictory.[17,32-34] There have been no studies on the effects of the stiffness of the stem and the thickness of the cement on interfacial micromovement.

Our aim was to investigate the effects of the roughness between the bone and the cement, the stiffness of the stem and the thickness of the cement on the cement stress and on interfacial micromovement.

Materials and Methods

The three-dimensional geometry of the proximal femur was reconstructed using quantitative CT. The distribution of the bone density corresponded to that measured immediately after implantation. Bone anisotropy was reproduced from anatomical observation and was implemented accordingly.[35] The Young and shear moduli of each bone finite element were calculated according to the distribution of the bone density.[36]

A collarless, straight, symmetrical stem with a cement mantle of uniform thickness was digitised and inserted numerically in the reconstructed femur. The stem was implanted after the usual surgical procedure with normal reconstruction of the offset femoral head and positioned in neutral valgus. A previously developed three-dimensional finite-element model of the cemented bone-implant system was used.[37]

We considered two interfaces: the cement-stem and the cement-bone. As in previous models, the cement-stem interface was assumed to behave as a coulombian dry interface.[18] In the present study, the cement-bone interface was also modelled to have discontinuous frictional contact, making it possible to calculate the debonding and slipping as well as the normal and shear stresses. The coefficient of friction at the cement-bone interface was set at 1.0,[6] which decreased to one-third when the cement roughness was removed. Young's modulus of the cement was 2200 MPa.[6]

The loading conditions corresponded to single-limb stance in the gait cycle. The load bearing on the femoral head was simulated with a force of magnitude which was three times the body-weight (patient: 60 years; weight 600N) and divided into axial, in-plane and out-plane directions[38] (Table I). The spatial reference system was defined as follows: the lateral-medial direction is denoted as x, the anterior-posterior direction as y and the vertical direction as z (up positive).[39] The muscle forces of gluteus minimus, medius and maximus and iliopsoas were incorporated in the model.[40]

Our study consisted of three parts. First, a titanium stem (Ti6Al4V) (Young's modulus 110 000 MPa; Poisson's ratio 0.3) was considered and then compared with a cobalt-chromium stem (Young's modulus 200 000 MPa; Poisson's ratio 0.3). The thickness of the cement was set at 4 mm and the friction coefficients at the cement-bone and at the cement-stem interfaces at 1.0 and 0.4, respectively.

Secondly, we investigated the sensitivity of the results with respect to thickness of the cement of 2, 3, 4, 5 mm and 7 mm for both types of implant. The coefficient of friction at the cement-bone interface was 1.0 and at the cement-stem 0.4. For areas where the width of the bone did not allow the prescribed thickness of cement, the local distance between the outer surface of the implant and the cortical endosteal femur was used instead.

Thirdly, we studied the effects of the surface roughness of the cement-bone interface. The friction coefficient at the cement-bone interface was set successively at 0.4, 0.6, 0.8 and 1.0 and the thickness of the cement mantle at 4 mm.

Results

Titanium alloy *v* cobalt-chromium alloy

Titanium implant. We measured the distribution of micromovement (debonding and slipping) at the cement-bone interface during a single limb-stance phase. Slipping was higher than 30 µm at the proximal lateral, intermediate medial and distal lateral regions. The peak value was approximately 67 µm. The debonding was, in general, less than 10 µm over almost the entire region of the interface. Nevertheless, the magnitude could exceed 30 µm in the proximal medial and distal lateral regions (peak 35 µm). At the cement-stem interface, slipping exceeded 30 µm over most of the proximal region and then decreased gradually towards the distal part.

At the cement-bone interface shear stress exceeded 1 MPa at the proximal lateral, intermediate medial and distal medial regions (peak 2.4 MPa). Elsewhere, it was lower than 0.5 MPa. High compressive stress (peak 4.4 MPa) occurred in the same locations as high shear stress (peak 4.4 MPa). At the cement-stem interface, the peak shear stress was 3.0 MPa and the peak compressive stress was approximately 7.0 MPa.

Cobalt-chromium implant. At the cement-bone interface slipping was high at the proximal lateral, intermediate medial and distal lateral regions (peak values 68 µm). Debonding was less than 30 µm over the entire interface (peak value 28 µm). Peak values for micromovement and stress at the cement-stem interface are shown in Table II

Table II. Stress and micromovement at the cement-stem interface for both materials

Thickness (4 mm)	Titanium alloy	Cobalt-chromium alloy
Debonding (μm)	73.6	63.1
Slipping (μm)	116.9	109.0
Pressure (MPa)	7.1	6.58
Friction (MPa)	2.8	2.63

(peaks: debonding 63 μm; slipping 109 μm; compressive stress 6.6 MPa; and shear stress 2.6 MPa).

Peak values of shear and compressive stress were higher than 1 MPa and had nearly the same distribution as for the titanium stem (peak: compressive 4.2 MPa and shear 2.5 MPa).

Effects of cement thickness. For both implants debonding at the cement-bone interface increased significantly with a thickness of cement of 2 mm (cobalt-chromium 115 μm and titanium 93 μm) (Fig. 1). With a thickness greater than 3 mm debonding was less than 30 μm over almost the entire interface. At the cement-stem interface, maximal debonding decreased slightly with the thickness of the cement. Minimal debonding for a titanium implant (33 μm) occurred when the thickness of the cement was approximately 3 mm (Fig. 1) and for a cobalt-chromium stem (28 μm) when the thickness of the cement was 4 mm (Fig. 1). In general the cobalt-chromium implant had a lower level of debonding than the titanium implant.

The evolution of slipping at the cement-bone and cement-stem interfaces related to the thickness of cement is shown in Figure 2. The behaviour of the titanium and cobalt-chromium implants was similar. For a thickness of cement of less than 3 mm, slipping increased drastically and exceeded 100 μm over the entire cement-bone interface (titanium 1650 μm and cobalt-chromium 680 μm). The peak values for slipping were minimised when the cement thickness was 3 mm (52 μm for both implants). For a thickness of cement in the range of 3 to 7 mm slipping remained constant, with highest values occurring in the

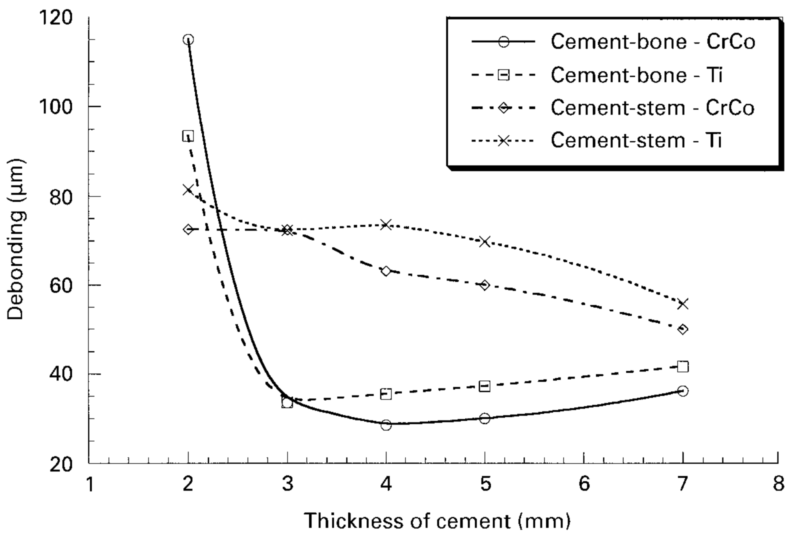

Fig. 1

Maximum values of microdebonding at the cement-bone and cement-stem interfaces related to the thickness of the cement.

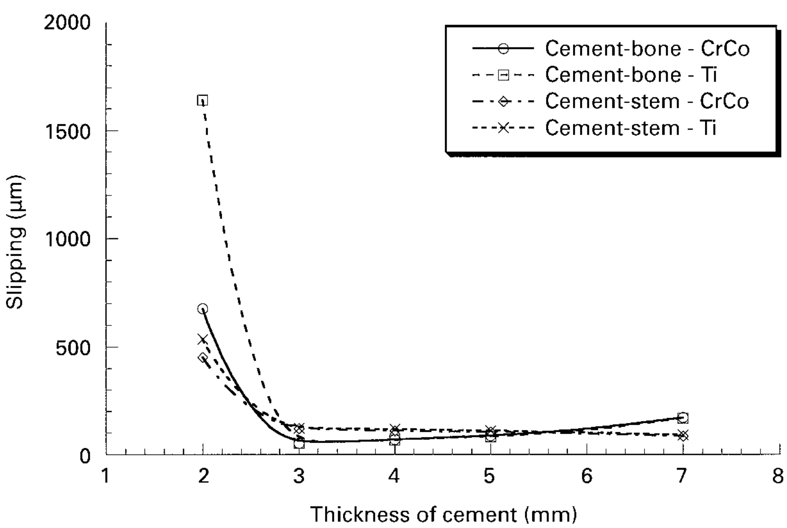

Fig. 2

Maximum values of microslipping at the cement-bone and cement-stem interfaces related to the thickness of the cement.

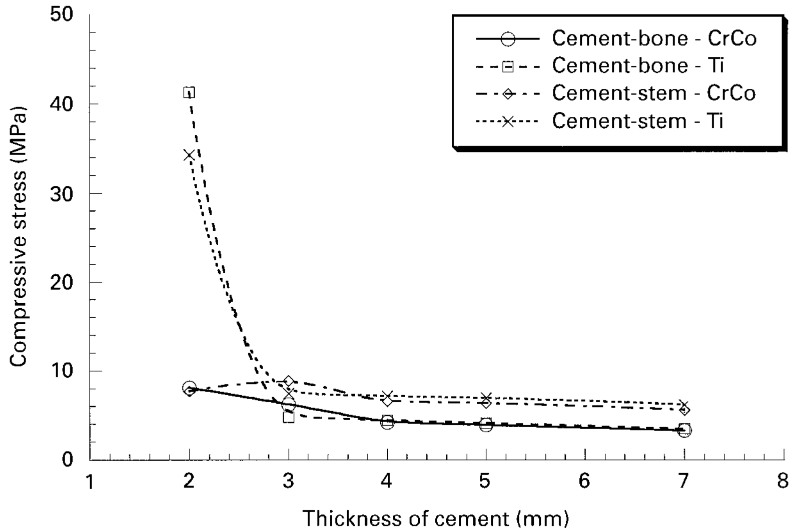

Fig. 3

Maximum values of compressive stress at the cement-bone and cement-stem interfaces related to the thickness of the cement.

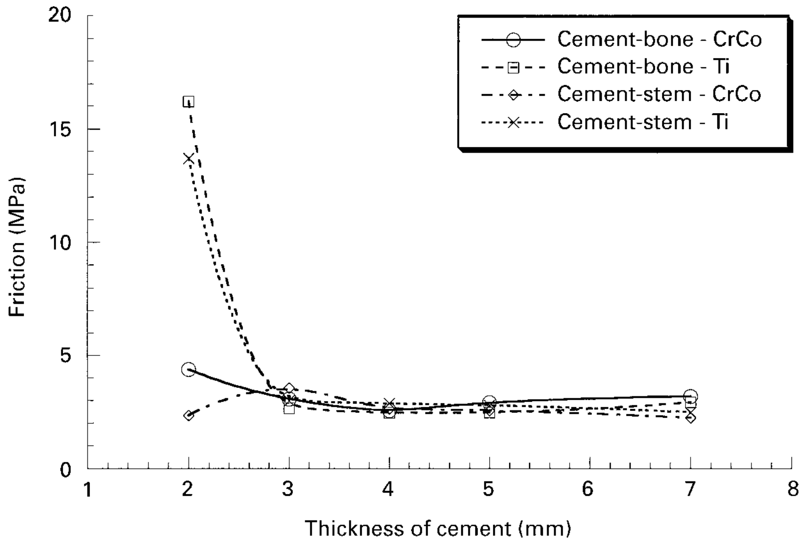

Fig. 4

Maximum values of frictional shear stress at the cement-bone and cement-stem interfaces related to the thickness of the cement.

proximal and distal lateral and medial regions of the cement-bone interface. For thicknesses greater than 7 mm the peak values of shear micromovement increased to 170 μm. In all cases slipping at the cement-bone interface was greater than slipping at the cement-stem interface.

The peak values of shear and compressive stress at both interfaces for thickness of cement greater than 3 mm were not significantly affected by the thickness of the cement for either implant (Figs 3 and 4). In addition, the region of high shear stress shifted gradually to the distal part of the implant with increase in thickness of the cement. For a thickness of 2 mm, abnormally high compressive (Fig. 3) and shear stresses (Fig. 4) were observed at both interfaces for the titanium but there was only a slight increase for the cobalt-chromium stem. Overall, the shear stress was lower than the compressive stress.

Effects of friction coefficient. The magnitude of slipping at the cement-bone interface was higher than 30 μm for the entire interface when a small coefficient of friction ($\mu = 0.4$) was used. The maximal values of slipping were higher for a low coefficient and remained nearly constant for higher levels. Conversely, the debonding was minimal for a lower coefficient and maximal for a higher. When the friction coefficient of the cement-bone interface was increased, slipping at the cement-stem interface increased and debonding decreased (Fig. 5).

The compressive stress was minimal for a low friction coefficient ($\mu = 0.4$) and increased gradually with increase in the coefficient, while the shear stress remained nearly constant (Fig. 6). Maximal compressive and shear stresses in the cement did not depend significantly on the friction coefficient at the cement-bone interface. Cement-bone stress (compressive and shear) was lower than cement-stem stress and the shear stress was lower than the compressive stress.

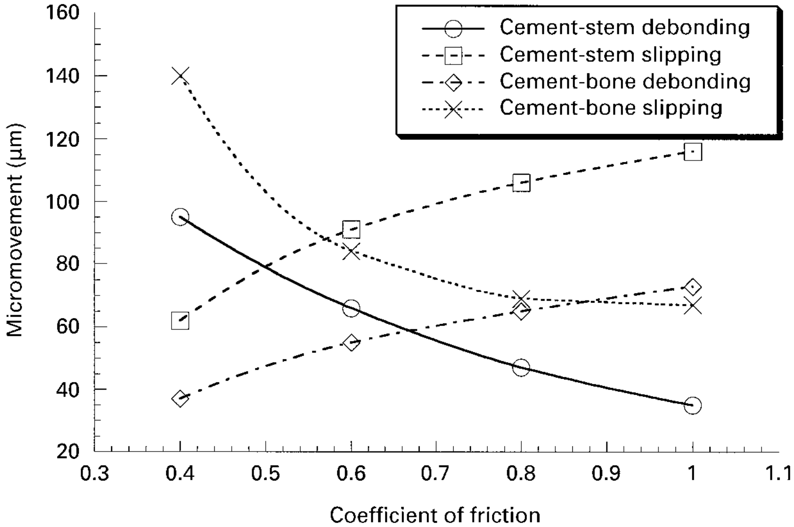

Fig. 5

Relative peak micromovement (microdebonding and microslipping) at the cement-bone and cement-stem interfaces related to the coefficient of friction at the cement-bone interface. The friction coefficient at the cement-stem interface was kept constant.

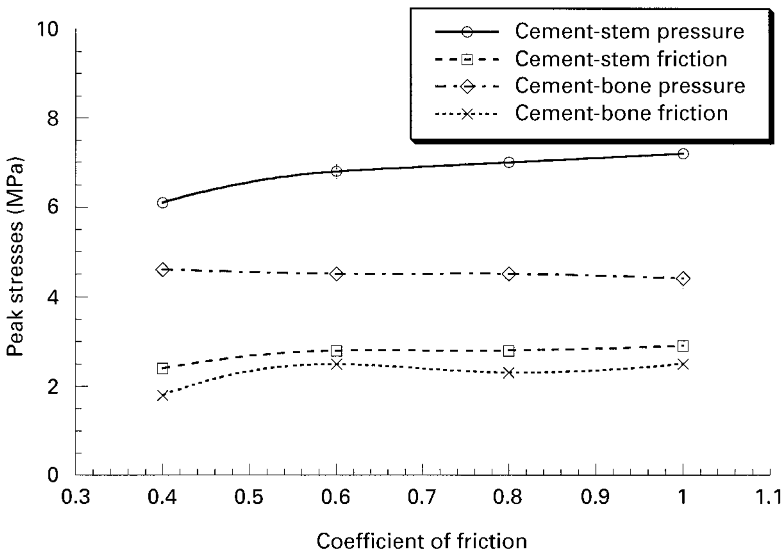

Fig. 6

Peak stress (compressive and shear) at the cement-bone and cement-stem interfaces related to the coefficient of friction at the cement-bone interface. The friction coefficient at the cement-stem interface was kept constant.

Discussion

Effects of stem stiffness. Cobalt-chromium and titanium alloys are widely used for the manufacture of femoral components. For a given shape a titanium stem is less stiff than a cobalt-chromium stem. The clinical issues for these two types of material remain controversial. Previous reports[41,42] suggest that a stiffer stem reduces cement stress and enables it to last longer. Therefore cobalt-chromium seems to be the best material for THA. Previous numerical models[43,44] have shown, however, that a cobalt-chromium stem unloads the proximal femur to a higher degree than a titanium stem and that this may enhance stress-shielding.[22,26] Theoretically, less stiff implants reduce stress within the stem and transfer load over a greater area of the cortical bone.[43,44]

In our study, comparison of similar titanium and chromium-cobalt implants showed that the stiffness of the implant had no significant effect on micromovement at the cement-bone interface. Furthermore, the two stems presented comparable lower peak values for shear stress at this site. High stress occurred over almost the same area for the two implants. The occurrence of high slipping and high shear stress in the proximal part of the two stems may correspond to the clinical finding that osteolysis, due to debris particles generated by slipping, tends to occur laterally on the proximal part of the stem.[45]

Although the stiffness of the stem had no significant effect on micromovement and stress for a large range of cement thickness in the titanium stem, our study showed that a cement thickness of 2 mm drastically increased compressive and shear stress at both interfaces, which could lead to fracture of the cement. In turn, because of the combination of stress and micromovement, the tribophysics of the two materials may be very different and may influence the biological reaction at the cement-stem interface.

Effects of cement thickness. The effect of the thickness of the cement on the initial stability has been investigated in previous studies,[14,17,31-33] but the conclusions as to the optimal range of thickness are contradictory. Ebramzadeh et al[33] showed that a cement mantle thicker than 5 mm was responsible for the radiolucent lines at the cement-bone interface and that one less than 2 mm thick induced fracture of the cement. By using numerical models based on fully-bonded interfaces, Huiskes[34] recommended a non-uniform thickness of cement ranging from 3 to 6 mm for the proximal part of the canal. Kwak et al[17] recommended a uniform layer of 3 to 4 mm. Crowninshield et al[14] found that a decrease in the cross-sectional dimensions of the stem increased the stresses in both the stem and the cement. Brockhurst and Svensson[31] suggested that the thickness of the cement mantle in the proximal medial region should be minimised. Fisher et al[32] demonstrated that a thickness of 2 to 3 mm may provide a more favourable cement strain than a thinner mantle. These findings seemed to be explained by the occurrence of very high shear and compressive stresses[3] which predicted a high risk of fracture of the cement for a cement mantle less than 3 mm thick. This is supported by experimental results[33] which found that a thickness of less than 2 mm induced fracture of the cement.

Our study has shown that the thickness of the cement affected not only stress but also micromovement. A layer of cement of 2 mm thick increased shear stress proximally and at the tip of the stem, and increased micromovement over the entire cement-bone interface. These outcomes could be related to local bone necrosis and the occurrence of osteolysis in these regions due to the presence of the cemented debris resulting from interfacial shear friction.[45] Surprisingly, thickness of cement greater than 7 mm increased slipping at the cement-bone interface, confirming other experimental results.[33] Our study suggested that an optimum thickness of cement was in the range of 3 to 5 mm.

In addition to the usual stress analysis the influence of the size and shape of the stem, its orientation and the non-uniformity of the thickness of the cement should be analysed.

Effects of friction coefficient. Mohler et al[46] found that the failure of the bond at the cement-stem interface may initiate loosening. This remains controversial. The effects of the friction coefficient at the cement-stem interface on stress and micromovement have been studied in numerous models,[15,19] but the effect of the roughness of the two interfaces has not yet been investigated.

Our results have shown that the decrease in the cement-bone coefficient of friction increased slipping and decreased debonding at the cement-bone interface, the cement-stem coefficient of friction being maintained constant. Improvement of the bond at the cement-stem interface significantly increased the relative slipping at the cement-bone interface. Such a phenomenon could be proposed as a biomechanical process promoting early failure at the cement-bone interface.[7] The main change was the magnitude of the compressive stress which was probably due to the hoop effects resulting from the pistoning of the stem in the cement mantle. This abnormal increase in the compressive and hoop stresses induced an overloading of the cement mantle. This model could bring new insight to the biomechanical analysis of cemented stem fixation.

Our study had a number of limitations. We used only loading corresponding to a single limb-stance phase which did not constitute the worst case for the femoral component. The use of other types of loading, such as stair-climbing or standing from a chair, could possibly give more useful information.

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