

Do short-stemmed-prostheses induce periprosthetic fractures earlier than standard hip stems? A biomechanical ex-vivo study of two different stem designs

Eike Jakobowitz · Jörn B. Seeger · Christoph Lee ·
Christian Heisel · Jan P. Kretzer · Marc N. Thomsen

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Abstract

Introduction The causes of periprosthetic fractures of the femur due to the design of the prosthesis and the individual parameters of the patient are unexplored. By different anchorage techniques in cementless total hip arthroplasties, it is assumed that there are various load limits of the implant's bearing femur.

Materials and methods In the present study, we compared a standard hip stem (cementless Spotorno[®]) and a short-stemmed design (Mayo[®]) by an artificial reproduction of periprosthetic fractures in 20 femur specimens.

Results The measured fracture loads showed an extensive range, with higher maximum loads in the standard stem group. The bone mineral density and the subsiding pattern of the standard stems showed a significant correlation to the incidence of the periprosthetic fractures. In the experimental setup, a slightly lower fracture resistance was shown for the short-stemmed prosthesis. Additionally, it was shown that donors with a higher body mass index had a significantly increased fracture risk.

Conclusions Short-stemmed prostheses, especially the Mayo[®] hip, do not constitute a higher fracture risk. In

general, an increased body mass index among patients with a cementless hip stem is associated with an increased fracture risk, particularly at high load values, i.e., resulting from a step during stumbling. Taking into account the ascertained results, the danger of provoking a femoral periprosthetic fracture can be reduced.

Keywords Hip joint · Arthroplasty · Fracture · Bone density · Biomechanics

Introduction

Postoperative periprosthetic fractures after total hip arthroplasty are an uncommon but serious complication that can be difficult to treat [17, 19, 23, 24, 28]. The incidence of these fractures ranges from 0.1 to 4.0% [9, 15, 18, 24, 25]. Predisposing factors seem to be cementless fixation, age and a poor bone quality [2, 5, 21]. The increasing number of implanted total joint arthroplasties will lead to a higher incidence of these fractures [6, 9, 23, 28]. The differences regarding the types of periprosthetic fractures depend on biomechanical characteristics of the implant system [17]. Among others, these characteristics are affected by the stem design and the type of fixation. The aim to preserve metaphyseal bone-stock with a proximal load transmission entails new developments. These so-called short-stemmed - or neck preserving- prostheses are characterized by a metaphyseal stem fixation.

A short stem shows smaller lever arm conditions and other CCD-angles than a conventional standard stem. Therefore, a short stem, especially in combination with decreased bone density, may more likely lead to a periprosthetic fracture.

E. Jakobowitz · J. B. Seeger · C. Lee · C. Heisel · J. P. Kretzer
Laboratory of Biomechanics and Implant Research,
Department of Orthopaedic Surgery, University of Heidelberg,
Schlierbacher Landstrasse 200a, 69151 Heidelberg, Germany

J. B. Seeger · C. Lee · C. Heisel
Department of Orthopaedic Surgery, University of Heidelberg,
Schlierbacher Landstrasse 200a, 69118 Heidelberg, Germany

M. N. Thomsen (✉)
Department of Orthopaedics,
German Red Cross (DRK) Clinic Baden-Baden,
Lilienmattstrasse 5, 76530 Baden-Baden, Germany
e-mail: marc.thomsen@drk-klinikbb.de

Many clinical analyses and treatment protocols regarding periprosthetic fractures of the femur after total hip replacement can be found in literature. But until today no experimental analysis has been made looking at the potential differences of a proximally fixed short stem in comparison to a standard stem.

In an experimental setup the maximum loads, which induced a periprosthetic fracture were measured in paired fresh frozen femora. Influencing factors like bone mass density (BMD), body weight (BW) and stem subsidence were analyzed.

Materials and methods

Prostheses

For the comparison of a neck preserving prosthesis with a conventional prosthesis, two established designs with well documented survival rates and comparable CCD-angles were chosen: The Mayo[®] hip (Zimmer, Warsaw, IN, USA) as the short design and the cementless Spotorno (CLS[®]) (Zimmer, Warsaw, IN, USA) as the standard design (Fig. 1). The Mayo[®] hip is designated for the cementless fixation in the region of the proximal femur (metaphyseal). This design features a 132° CCD-angle, a double tapered design and it is angulated at the lower section to achieve lateral support at the internal femoral cortex. It is made of titanium and has a circumferential grit-blasted region interposed with porous surfaces covering the entire proximal stem. In addition, there are pads of mesh on the anterior, posterior, and medial surfaces proximally. The CLS[®] prosthesis features a CCD-angle of 135°. Initial stability depends on a series of ribs on the proximal anterior and posterior aspects of the double tapered, straight, grit-blasted titanium stem which provides an interference fit in the femur with its rectangular cross-section [1].

Femoral specimens

There were ten paired fresh frozen human femora available to simulate postoperative conditions for the implanted stems. Relevant data of donors such as body mass index (BMI), age and gender were documented. For each femur an osteodensitometric determination (QDR-2000 DXA-Densitometer, Hologic Inc., Waltham, MA, USA) of the bone mineral density (BMD) was performed in five regions of interest: femoral neck, trochanteric, intertrochanteric, Ward's triangle and over the entire femoral head (total). In order to avoid a left–right-deviation the specimens were allocated equally between both stem designs. A preoperative planning with templates was done to choose the correct stem size. Bone preparation and stem insertion were



Fig. 1 The CLS[®] stem (*left*) and the Mayo[®] stem (*right*)

performed by one single experienced surgeon (X.Y.). Visual inspection and repeated X-ray examination in two planes ensured that no fissures or fractures occurred during implantation.

Experimental setup

After approval by the local ethics committee, the specimens were fractured under controlled loading conditions. To carry out a physiological load transfer, a vector had to be found that could be associated with a patient activity immediately after surgery. Concerning the early phase of the mobilization, specific activities are prohibited in line with the therapeutic schedule, including stair climbing or running. They could result in an overstraining of the bone prosthesis fixation. Due to this issue, the loading of “normal walking” was chosen. For a quasi-static load transfer we chose one point of the average curve of the telemetric *in vivo* measurements from Bergmann et al. [3, 4]. At this point the contact force of the hip joint reaches its maximum amount ($F_{\max} = 233\%BW$). Consequently, the strain of oss-

eos fixation is assumedly at its highest level. In order to assure an unaltered transmission of the load vector, the coordinate system described by Bergmann et al. [3, 4] was adopted. Using projected laser beams, the femora were aligned in the coordinate system. In this position, the specimens were adjusted with a clamp-stand. For a distance of 10 cm from the distal end, the femora were fixed with PU-plastic (Rencast FC 53 Polyol, Huntsman, Everberg, Belgium) into an aluminum box, which was part of the fixation device. The compound was aligned into a material testing machine (Frank Universal Testing Machine 81816/B, Karl Frank GmbH, Weinheim, Germany). In order to accomplish the chosen loading, the specimen had to be adjusted with the angles α_Y and α_Z . They were calculated by the partial vector components $-F_Y$ and $-F_Z$ published by Bergmann et al. [3, 4]. The fixation device was then assembled onto a special constructed testing device using a wedge and a displacement slide. This setup allowed the arrangement of the bone in a position of $\alpha_Y = 13.1^\circ$ (adduction) and $\alpha_Z = 31.8^\circ$ (internal rotation) between the plungers. An additional x-y-slide fixed on the upper plunger secured a load transmission free of shearing forces.

Statistical methods

The attained data were analyzed with the two-sample Wilcoxon test. Pearson’s correlation was used to compare donor parameters within one prosthesis-system. The significance level was fixed at $\alpha = 0.05$. Statistical evaluation was performed using the analytical software SPSS® for Windows®, version 12.0 (SPSS Inc., Chicago, IL, USA).

Results

The median BMI of the donors was 24.3 kg/m² (range: 11.9–50.3 kg/m²) (Table 1).

The measured BMD values of the femurs which obtained the CLS® stems were 0.825 g/cm² (neck), 0.775 g/cm² (trochanteric), 1.089 g/cm² (intertrochanteric), 0.664 g/cm² (Ward’s triangle) and 0.951 g/cm² (total). BMD values

of the specimens which obtained the Mayo®-stems were 0.815 g/cm² (neck), 0.765 g/cm² (trochanteric), 1.059 g/cm² (intertrochanteric), 0.664 g/cm² (Ward’s triangle) and 0.951 g/cm² (total). Regarding the CLS® stem, the measured subsidence was $s = 13.9$ mm (range 5.8–21.7 mm) until the periprosthetic fracture occurred. The depth of the subsidence for the Mayo® stem was only $s = 7.9$ mm (range 2.9–14.0 mm). The subsidence of both systems before the fracture occurred was significantly different ($P = 0.013$). The CLS® stem subsided on average $\Delta s = 5.7$ mm deeper into the cavitas of the femur (Fig. 2). The maximum force to induce a periprosthetic fracture was $F_{max} = 5,545$ N (range 3,294–8,102 N) for the CLS® stem. This corresponded to a relative maximum force of $F_{max} = 855\%$ BW (range 311–1,655%BW) in respect to the donor’s body-weight. For the Mayo® stem the fracture force was $F_{max} = 4,825$ N (range 2,651–7368 N). This corresponds to a relative maximum force of $F_{max} = 751\%$ BW (range 256–1,669%BW; median 752%BW). On average, the specimens

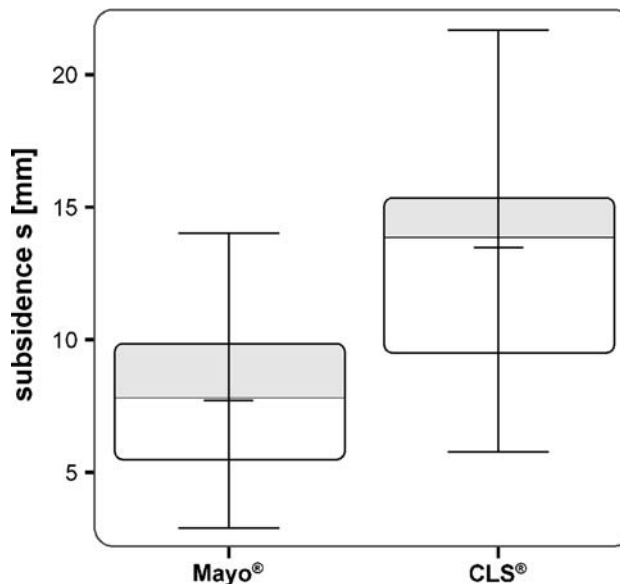


Fig. 2 Comparison of the subsidence until fracture occurred. The CLS®-stem subsided significantly deeper into the cavitas of the femur ($P = 0.014$)

Table 1 Calculated BMI of donors, BMDs, subsidence s until the periprosthetic fracture occurred and absolute and relative fracture loads

	BMI [kg/m ²]	Bone mineral density														s [mm]		F_{max} [N]		F_{max} [%BW]	
		Neck [g/cm ²]		Troch [g/cm ²]		Inter [g/cm ²]		WARD [g/cm ²]		Total [g/cm ²]		Mayo	CLS	Mayo	CLS	Mayo	CLS	Mayo	CLS		
		Mayo	CLS	Mayo	CLS	Mayo	CLS	Mayo	CLS	Mayo	CLS	Mayo	CLS	Mayo	CLS	Mayo	CLS	Mayo	CLS		
Min	11.9	0.541	0.547	0.540	0.553	0.753	0.767	0.428	0.434	0.657	0.664	2.9	5.8	2,651	3,294	256	311				
Max	50.3	0.945	0.950	0.831	0.839	1.216	1.272	0.942	0.952	1.048	1.103	14.0	21.7	7,368	8,102	1,669	1,655				
Mean	25.5	0.792	0.796	0.725	0.737	1.028	1.046	0.654	0.662	0.905	0.922	7.7	13.4	4,744	5,425	774	841				
Median:	24.3	0.815	0.825	0.765	0.775	1.059	1.089	0.659	0.664	0.939	0.951	7.9	13.9	4,825	5,545	751	855				

implanted with Mayo[®] stems fractured at lower forces without statistical significance (Fig. 3). The difference between the systems concerning the maximum fracture loads in relation to the donor's weight F_{\max} (%BW) was also not significant (Fig. 3). The correlation between the maximum fracture load F_{\max} (N) and the BMD in all measured areas of the specimens with implanted CLS[®] stems was highly significant (i.e., trochanter: $P = 0.003$, $r = 0.834$; ward: $P = 0.006$, $r = 0.793$). The maximum performance increased depending on the density. This correlation was not proven for the Mayo[®] stem. Fig. 4 demonstrates the process of the quasi-static loading application in a displacement-load-diagram until the periprosthetic fracture appeared for the specimen (no. 9, left) with an implanted CLS[®] stem. The curve rises in a constant way and the hip stem shows a subsidence of $\Delta s \approx 3$ mm for each $\Delta F = 1,000$ N transferred. Initially, the bony bearing was able to resist the forces, but at a maximum force of $F_{\max} = 6,254$ N and a subsidence of $s = 18.0$ mm the abrupt release of the fracture occurred. All fractures proceeded along the femoral axis; they began at the rim of the resection's cut and spread with varying lengths distally. Two Mayo[®] stems and one CLS[®] stem disrupted a medial fragment and two Mayo[®] stems burst out of the lateral cortex with the tip of the stem. The fractures were located as follows: Mayo[®], eight fractured medially in the area of the calcar and two laterally in the area of the distal stem. CLS[®], seven fractured medially in the area of the calcar, two laterally in the proximal area and one ventrally in the proximal area. A significant correlation between the localization of fracture and other parameters such as maximum fracture loads F_{\max} (N) and F_{\max} (%BW), subsidence s or BMD could not be found. The length of the fracture lines were $l = 60$ mm (range 6–120 mm) for the CLS[®] group and $l = 65$ mm (range 9–120 mm) for the Mayo[®] group. It was ascertained that the BMI of the donors showed a significant correlation to the relative fracture loads F_{\max} (%BW) for both types of stems (CLS[®]: $P = 0.023$, $r = -0.705$; Mayo[®]: $P = 0.016$, $r = -0.731$). Periprosthetic fractures are more

Fig. 3 Absolute and relative maximum fracture loads. There was no significant difference between both systems [F_{\max} (N), $P = 0.436$; F_{\max} (%BW), $P = 0.684$]

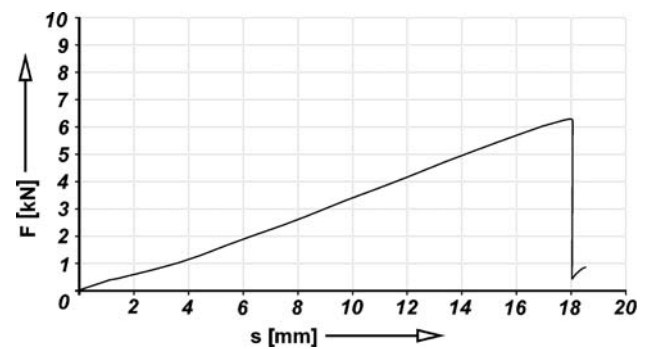
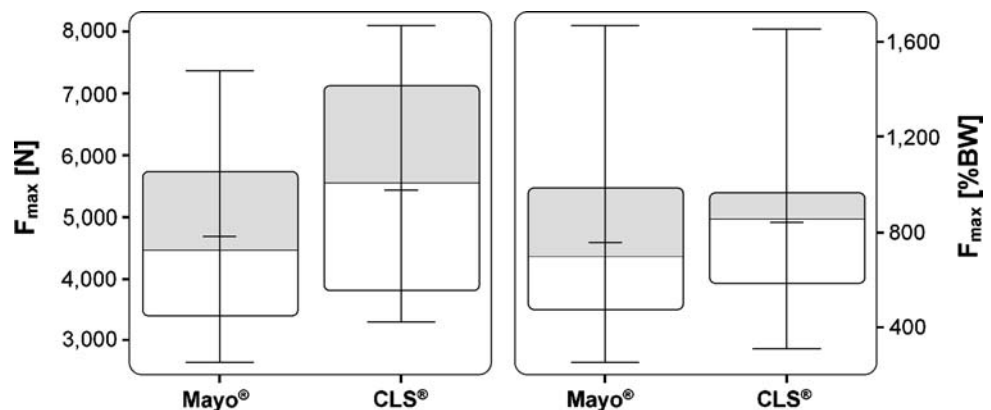


Fig. 4 Load displacement diagram of the quasi-static loaded specimen no. 9 left implanted with CLS[®] prosthesis (size: 9)

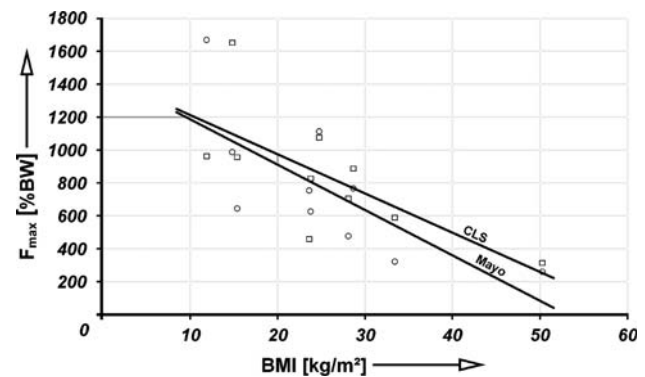


Fig. 5 Correlation of the relative fracture load F_{\max} and the calculated donors BMI (kg/m^2) (quad CLS[®], ring Mayo[®]). With a BMI = 30 kg/m^2 , both prosthetic systems already vary about $\Delta F_{\max} = 100\%$ BW

likely to occur with a higher BMI; therefore, the slope of the linear ratio for the CLS[®] stem turned out to be slightly steeper (Fig. 5).

Discussion

Regarding the BMD-measurements the results are comparable. There is an enormous difference in the subsidence

values, but only an insignificant one between the maximum forces provoking the fracture. Against our expectations, there were three cases of higher maximum forces in the Mayo[®] group. The ascertained maximum fracture loads of specimens, which were implanted with one type of prosthesis, showed an extensive range. This applies to both the absolute and relative values. This range could be closely compared with the results of Carls et al. [7]. For their stress analysis of cerclages, artificial fractures were induced with Wagner[®] stems in ten pairs of femur specimens with applied axial forces. The maximum fracture load was $F_{\max} = 6,531$ N (range 1,915–9,288 N). The measured fracture loads showed a salient contrast between both systems. The Mayo[®] stem induced a periprosthetic fracture much earlier as the CLS[®] stem, differing about $\Delta F_{\max} = 800$ N, respectively, $\Delta F_{\max} = 100\%$ BW. But this distinction had no statistical significance. Thus, it was clear that the maximum fracture loads of the CLS[®] stem were higher in just seven of ten cases. We found that the used number of cases should be increased for further investigations. With the doubling of cases, a significant difference may be expected. The average of the maximum force acting on the hip joint during normal walking is $F_{\max} = 233\%$ BW [3, 4]. In all specimens tested, the necessary force to cause a periprosthetic fracture was more than $F_{\max} = 233\%$ BW. This would imply that femoral fractures that occur around the tested designs may be no risk for a fatal failure with normal weight bearing. The ascertained results, however, showed that the contact forces of the hip, which rise above the bodyweight about $F_{\max} = 250\%$ BW, may cause a danger. This represents, for example, the step to keep from falling when stumbling ($F_{\max} = 720\text{--}870\%$ BW [4]). In twelve cases, the magnitude of the vector that acts on the hip while stumbling was not topped at all. Consequently, the step to keep from falling is a potential risk for more than half of the patients treated with a cementless hip stem of tested designs. This critical conjuncture even supports the thesis of Peicha et al. [21]. They assumed that periprosthetic fractures are facilitated by excessive stress concentration, caused by overloading. From majority of authors, the weak bone density as well as osteoporosis, generally is quoted as further predisposing factors of periprosthetic fractures [2, 5, 21, 29]. This assumed connection could be found only for the CLS[®] stem in this examination, not for the Mayo[®] stem. This shows that the prevailing opinion is not applicable for all hip prostheses. For the Mayo[®] stem there seems to be other factors playing a more important role.

Both the CLS[®] stem and the Mayo[®] stem showed an extensive subsidence into the cavity of the femur until the periprosthetic fracture was induced. The difference of the depth of subsidences of both implants proved to be statistically significant. The CLS[®] prosthesis subsided on average $\Delta s = 5.7$ mm deeper into the cavity. This relation coincides

with the analysis of Huiskes et al. [12, 13] and Hube et al. [10, 11], which dealt with the process of load transmission of the Mayo[®] prosthesis. However, in 1986 Huiskes et al. proved in a 3D-FE-analysis [12, 13] that the effect of the bending moment from the force of hip contact on the Mayo[®] implant was the most significant out of all the loads. Therefore, the biggest values of the force were located in the ventral-medial femur. The comparatively marginal subsidence of the Mayo[®] prosthesis of $s = 0.15$ mm at $F = 3$ kN (subsidence ratio to the load $s/F = 0.05$ mm/1 kN) could not be proved. In the study at hand a minimum $s/F = 1.4$ mm/1 kN was detected. Hube et al. [10, 11] also described a load transmission of the Mayo[®] prosthesis particularly in the region of the calcar.

Periprosthetic fractures of the femur are classified differently [9]. Basic criteria are the localization and the morphology of the fracture [9, 21]. Frequently cited classifications like Whittaker et al. [26], Johansson et al. [14], Mont et al. [19] and Duncan and Masri (Vancouver-classification) [8, 20] usually do not refer to fractures, which are induced by a subsiding, cementless implant (radial adjusted force vectors). On the other hand, the generation of fractures due to adequate traumas plays an important role. These fractures are located at different heights crosswise to the diaphysis longitudinal axis. The fractures of this study disperse longitudinally to the diaphysis from the proximal to distal side. They split the femur lengthwise. Thus the numerous classifications of fractures for this study were not reasonable. Both stems proved to have a tapered design lengthwise. This is why the authors assumed that, due to the massive subsidence of the implants into the cavity (between $s = 2.9$ and 21.7 mm), a lot of tension evolved in the lumen of the femur. This can lead to radial tension especially in the entry level, which causes a proximal-to-distal lengthwise splitting of the femur. In 1998 Carls et al. [7] had already showed that in particular, the medial cortex of the femur should be regarded. In their analysis they found that all longitudinal fissures have always been located at the point where the cortex has the thinnest wall. In agreement with the present study, they almost always detected medial and ventral fissures lengthwise, thus a splitting, which were located at the artificial impression of the conical stem of the prosthesis. Another conclusion shows that the medial crack of the fracture, which has also been defined as an incomplete fracture by Schwartz Jr et al. [22], often cannot be detected in X-rays. This is due to the fact that longitudinal fractures often coincide with the projection of the stem. Hence, it is presumed that fractures which are located medially in particular often remain clinically undetected. In the case of a fracture, the implant would not be sufficiently fixed. However, due to the fact that a high number of periprosthetic fractures emerge from loosening of the implant [6, 21, 27]

and longitudinal fractures are found rarely in practice, it is assumed that only a consequent fragmentation of the bony bearing can be noticed in X-ray as a periprosthetic fracture.

The relationship between the BMI of the donor and the relative fracture load F_{\max} (%BW) was especially interesting. Due to the fact that the bodyweight was included in both parameters, a correlation that evoked a statistic significance could be found. Initially we assumed that the bone should be accommodated to its massive loading in relation to the bodyweight and therefore a significant correlation between BMI and BMD should be found, but this was not the case. According to this result, the ability to resist high strains from a prosthetic stem with a high BMI decreases. On the other hand it appears reasonable that different bodyweights will produce different fracture loads relatively [F_{\max} (%BW)]. The factor of bodyweight with which the prosthesis can be loaded can now be calculated for clinical use by the patient's weight and length using the ascertained linear coherence, so the danger of provoking a periprosthetic fracture can be reduced. With a BMI = 30 kg/m², both systems of prostheses already vary about $\Delta F_{\max} = 100\%$ BW of load ability. This fact can be used for the required prevention of periprosthetic fractures [18] in clinical practice. This presented difference can contribute to the surgeon's decision between these systems subject to the BMI.

It seems to be significant that femora with cementless stems tend to fracture within the first half year after implantation [2, 21]. This is most likely due to cortical stress risers created by reaming and broaching of the femoral canal [2, 27]. The applied setup only reflects the situation of periprosthetic fractures as a result of an overload of the leg directly after implantation. As with all in vitro studies, the experimental design does not necessarily represent in vivo situations. Since the inserting muscle forces have been summarized into one single vector, there is no stiffening effect of antagonistic muscles, e.g., the tensor fascia latae. This has to be kept in mind when applying the results to clinical circumstances.

Furthermore, there are many restricting factors influencing experimental examinations with biological systems. Due to these factors, the specimens signalized a distinct heterogeneity in the present study, which was shown by the donors data and the structural dimensions of the femora, as well as the differences at measured values of the BMD. However, the linear dependency between BMD in Ward's triangle and the donor's age corresponded exactly to the Bone Mineral Density Reference Database published by Kelly [16] in 1990.

The different results, and especially the differing dependencies of both prostheses designs, show that it is not possible to transfer the results found for the tested implants to other systems of prostheses. These results, like those

described by Beals and Tower [2], show that the design and the type of prosthesis do in fact play an important role in fracture mechanisms.

Regardless of whether the prosthesis is short- or standard-stemmed, we conclude that an increased body mass index (greater obesity) among patients with a cementless hip stem is associated with an increased fracture risk. This risk is particularly valid for high load values, i.e., resulting from a step during stumbling. For clinical application, it should be recognized that short-stemmed prostheses, especially the Mayo[®] hip, do not constitute a higher fracture risk. With regard to the missing comparability of our fracture appearance to the well known fracture classifications, the ascertained longitudinal splitting should be verified clinically and in further in vitro investigations.

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