Strut grafts in revision hip arthroplasty faced with femoral bone defects: an experimental analysis

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Abstract

Purpose In total hip arthroplasty fixation of revision stems can be demanding due to femoral bone loss. Strut grafts are often used for bone augmentation and stabilization of the newly inserted prosthesis. The aim of this study was to assess the effect of strut grafts on primary stability under various stem fixation conditions.

Methods Two different revision stems (cylindrical and conical shape) were implanted into synthetic femora. Following a semicircular transfemoral osteotomy, three deficient femoral bearings were simulated (bony lid reattached with cable wires; weakened lid reattached with cable wires; strut grafts placed to the weakened lid with cable wires). Relative micro-movements were measured between prostheses and bones due to an axial moment applied to the stems.

Results Relative movements correlated to the stem shape. The cylindrical stem showed higher movements increasing significantly with a weakened bony lid and portrayed a slight decrease of movements with strut graft application. No

unequivocal influence of the weakened lid could be detected for the conical implant. Strut graft application did not show an additional stabilizing effect.

Conclusions The primary stability of the cylindrical fixation concept decreases with impaired fixation conditions of the femur. A clear restabilizing effect with strut grafts could not be proven. A decrease of primary stability due to the impaired bone could not be observed for the conical stem shape. Additionally, strut grafts do not enhance fixation for this stem shape. We conclude that surgeons should not rely on a stabilizing effect of strut grafts in revision hip surgery.

Keywords Total hip arthroplasty · Revision · Femoral defect · Strut allograft · Primary fixation

Introduction

Following aseptic loosening in total hip arthroplasty (THA), it is often difficult to achieve primary stability of the revision stem due to femoral defects [1]. In order to manage these defects, strut allografts are frequently used [2–10]. They are individually sized beams of non-vital bone attached to the deficient femur via tension or cable wires. Incorporation and remodeling of these allografts to the host bone has been shown to be successful in 80–100 % of cases in numerous studies [2–11]. Some authors solely use strut grafts to rebuild the proximal femur [7, 8, 9], whereas other orthopaedic surgeons also rely on these grafts for a support of the stem to achieve better stability [3, 6, 9]. Although primary stability is the basic requirement for osteointegration, and subsequent good long-term stability of the implant [12–14], there is no evidence that strut grafts have an effect on primary fixation of revision stems. Our hypothesis was an enhanced stability of stems implanted under impaired bony conditions due to the application of strut grafts. The aim of this study was to analyse the

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influence of strut grafts on the primary stability of revision stems in femora with proximal defects. Two commonly used revision stems [5, 15–18] with different fixation philosophies [19] were included to assess this influence for different load transfers from stem to femur.

Materials and methods

Experimental study protocol

To allow for standardized implantations in the light of an enhanced comparability to previous investigations [19, 20], we decided to use second generation composite femora (#3106, Pacific Research Laboratories, Inc., Vashon, USA). Both strut grafts applied were taken out of a femur which was not to be further used and therefore had standardized dimensions of 16×155 mm and 16×175 mm for all measurements.

Regarding different fixation philosophies we chose a cylindrical and a conical shaped stem design. The cylindrical shaped S-ROM® stem (DePuy Orthopaedics Inc., Warsaw, USA) with a porous coated proximal sleeve (Fig. 1) represented a more metaphyseal load transfer [19]. It was intentionally included to compromise its primary stability due to its mainly proximal fixation being directly surrounded by the femoral defect defined later.

To simulate a more common clinical situation—bridging the proximal defect zone [3, 9, 10, 17, 18] by the use of a more diaphyseal fixating revision stem—the MRP® stem (Peter Brehm GmbH, Weisendorf, Germany) was included (Fig. 1). It features a conical shaped and fluted stem with a grit-blasted surface.

A surgeon with significant experience in revision THA (C.H.) templated stem sizes, performed a standardized neck resection [19] and identically prepared the femoral cavity with implant specific instruments. We followed an established protocol [19, 20] to achieve a standardized alignment and press-fit fixation for stems comparable to hip surgery [21] and to loads occurring during walking [22]. Therefore, the stems were loaded with 25 cycles of 2 kN and 4 kN using a materials testing machine (Type 81816/B, Karl Frank GmbH, Weinheim, Germany).

Three different conditions of a deficient femoral bearing were simulated sequentially on each femur. First, a standardized semitransverse transfemoral osteotomy was performed, which resulted in a ventral cortical lid. With respect to the lesser trochanter, the longitudinal saw cuts extended distally 110 mm (Fig. 2a). This constitutes a level III segmental defect according to the AAOS classification [23]. For the first bearing condition (I) the created lid was reattached to the femur with cable wires (Fig. 2b). For the second bearing condition (II) the lid was reduced to 3 mm cortical wall thickness in order to emulate frequently encountered cavitory defects with

Fig. 1 Hip revision stems S-ROM® (*left*) and MRP® (*right*) which were investigated. Digital templating using X-rays of unaffected femora prior to investigation revealed stem sizes of 18/215 for the S-ROM® and 17/200 for the MRP®. The sleeve size “large” fitted as best concerning the S-ROM®

cortical thinning as typically caused by osteolyses and osteopenia [23] (Fig. 2c). Standardized application of strut grafts to the weakened lid with cable wires (Fig. 2d) resulted in the third bearing condition (III).

To analyze the fixation of prostheses, relative micro-movements between stem and bone were measured according to a previously utilized method [19, 20]. Briefly, we applied an axial moment T_z onto the Morse taper lock of stems via a lever arm using long ropes. T_z was limited to one-third (± 7 Nm) of moments measured during gait by Bergmann et al. [22] to ensure a non-destructive loading. Therefore, a force system consisting of two linear motors hanging balanced and opposed on these ropes moved two weights in a reciprocal and cyclic way. This procedure provided negligible counteractions between movements of stems and force system’s deflections.

To depict the effect of femoral conditions regarding the complete fixation from the proximal to the distal stem a total

Fig. 2 Survey of femoral preparation steps: medial view of the osteotomized femur with ventrally displaced bony lid (a); bearing condition I: reattached lid (b); bearing condition II: weakened and reattached lid (c); and bearing condition III: application of strut grafts to the weakened lid (d). The ordinate represents levels within movement graphs (Figs. 3 and 4) for comparison purposes

of 11 measuring points distributed over the prosthesis (P0 to P4) and the femur (F0 to F4 and R0) were required (see Figs. 3 and 4). Angles of axial rotation α_Z [mdeg] at these points were calculated from spatial movements of bones and stems measured with a high-resolution measuring device (linear and angular resolution: $<0.1 \mu\text{m}$ and $<0.5 \text{ mdeg}$). This device consisted of a metal cube—coupled either to the prosthesis or the bone—probed by six room fixed LVDTs (Type P2010, Mahr GmbH, Göttingen, Germany) in a three-two-one configuration. These angles α_Z were then normalized with respect to T_Z resulting in α_Z/T_Z [mdeg/Nm], averaged from repeated measurements (see sample size calculation) for each stem and bearing condition and transcribed into motion graphs (Figs. 3 and 4). Differences of α_Z/T_Z between prosthesis and bone constituted the relative rotational movement ($rm = \Delta\alpha_Z/T_Z$ [mdeg/Nm]) between them. Therefore, rm is inversely proportional to primary stability. In other words, a higher torque transfer from the stem onto the femur coincides with a better fitting area and could be identified by a smaller value of rm . Finally, the absolute interface slipping in μm was calculated correlating rm with stem diameters multiplied by a physiological axial hip moment during walking ($\sim 25 \text{ Nm}$) [22].

Statistical computations

According to previous data [19], a power analysis with alpha and beta errors of 5 and 20 % (power 80 %) and assumed relative rotatory motions of 4.4 and 10.7 (SD=3.0) mdeg/Nm of groups indicated a group size of $n=4$. Friedman's test was used to assess reliability of the resulting fixation behaviour of

Fig. 3 Motion graph for the S-ROM®-stem. Connecting coloured lines (solid with squares = prosthetic measuring points P0 to P4; dashed with circles = femoral measuring points F0 to F4 and R0) result from plotting normalized angles of rotation α_Z/T_Z (ordinate) against corresponding measuring levels (abscissa). It is the difference between colour-coherent solid and dashed lines that constitutes relative movements rm_1 to rm_4 . They arise from opposing prosthetic and femoral measuring points (P1-F1, P2-F2, P3-F3 and P4-F4). Colouring indicates motions depending on femoral conditions II and III (red = weakened lid reattached with cable wires; blue = strut grafts placed to the weakened lid with cable wires). Data are represented as mean and SD in one direction (indicators)

prostheses (distribution of value order from proximal to distal) in different bearing conditions. Differences in interface slipping at measuring points were revealed with a hierarchically designed analysis of variance.

A further analysis of variance was carried out to compare primary stabilities over all bearing conditions. Therefore, we needed standard values for each stem and bearing condition and built a mean overall relative movement out of rm_1 to rm_4 (Table 1). For both analyses of variance a “least significant difference”-test (LSD) was calculated to disclose particular effects within the statistical models.

Results

A highly significant level of reproducibility for value order of relative movements and therefore the fixation behaviour of stems over all bearing conditions was observed (p -value always ≤ 0.01 , except for S-ROM® for bearing condition II: $p=0.03$). Histograms confirmed normally distributed data.

Fig. 4 Motion graph for the MRP®-stem. Normalized angles of rotation α_z/T_z [mdeg/Nm] are plotted on the ordinate and measuring level [cm] on the abscissa. Distances of connected prosthetic measuring points P0 to P4 (solid lines with squares) and femoral measuring points F0 to F4 and R0 (dashed lines with circles) at the same level (e.g. P1 and F1) allows for a characterization of relative movements rm_1 to rm_4 . Conjugation of line pairs can be identified by colored condition (red = weakened lid reattached with cable wires (II); blue = strut grafts placed to the weakened lid with cable wires (III)). Data are represented as mean and SD in one direction (indicators)

Mean relative movements

Figure 5 indicates smaller mean overall relative movements for the MRP® than for the S-ROM® in every bearing condition ($p < 0.01$). Compared to the initial situation (bearing condition I), a significant decrease in overall stability in bearing

conditions II and III for the S-ROM® could be shown ($p < 0.01$). The application of strut grafts (bearing condition III) resulted not in a significant increase of fixation for the S-ROM®.

Between bearing conditions I, II and III no differences could be found for the MRP® (Fig. 5).

Relative movements at different levels

Regarding fixation behaviours, relative movements increased almost continuously from proximal to distal for the S-ROM® whereas the MRP® showed a more homogenous fixation with accentuation in the proximal isthmus.

Variation of bearing conditions resulted in diverse influences on fixation behaviors. In bearing condition I both stems had comparable fixations within the proximal femur, whereas the MRP® moved less around the isthmus (Table 1).

After weakening the lid (bearing condition II) the distal part of the S-ROM® fixation decreased. Primary stability of the MRP® did not appear to be affected by the weakening. For the S-ROM® the additionally placed strut grafts (bearing condition III) resulted in a small recuperation of the lost distal implant fixation during bearing condition II (Fig. 3). However, application of grafts led to a slightly impaired stability for the MRP® as proximal movements increased with bearing condition III (Fig. 4).

The calculated interface slipping did not exceed 150 μm for both stems in all bearing conditions.

Discussion

Clinical studies analysing revision THAs showed no distinct differences of mechanical implant failure with strut graft application for hip stems as primary support [3, 6, 9] or exclusively as bone augmentation of the femur [7, 8, 10]. Based on these studies it is exceedingly difficult to compare the reported

Table 1 Femoral condition (fc) divided relative movements rm_1 to rm_4 [mdeg/Nm] between implant and bone resulting from corresponding measuring points P1 to P4 and F1 to F4 (see Figs. 3 and 4)

Stem	fc	Averaged relative rotational movement [mdeg/Nm] (μm)				
		rm_1 Distal metaphysis	rm_2 Proximal isthmus	rm_3 Distal isthmus	rm_4 Tip of stem	Mean overall Movement rm_1-4
S-ROM®	I	7.0 (10.8)	19.3 (30.0)	47.7 (73.9)	55.3 (85.1)	32.3 (50.0)
	II	10.9 (16.0)	32.5 (50.3)	75.5 (117.0)	77.8 (119.6)	49.0 (75.7)
	III	12.7 (20.0)	32.3 (50.1)	64.0 (99.1)	69.2 (106.4)	44.5 (68.9)
MRP®	I	7.2 (14.5)	2.8 (5.2)	17.9 (29.7)	25.7 (35.5)	13.4 (21.2)
	II	6.9 (13.7)	6.7 (12.5)	15.5 (25.6)	25.7 (35.3)	13.7 (21.8)
	III	15.3 (31.3)	7.4 (13.8)	15.7 (25.8)	25.3 (34.8)	15.9 (26.4)

The resulting interface slipping rm_1 to rm_4 (μm) refers to a physiological axial hip moment during walking [22]

Fig. 5 Interaction graph of mean overall movements rm_{1-4} (cp. Table 1, last column) over all measured femoral conditions: bony lid reattached with cable wires (I); weakened lid reattached with cable wires (II); and strut grafts placed to the weakened lid with cable wires (III). The stems can be separated by connecting lines: S-ROM®-stem (red line with squares) and MRP®-stem (green line with circles). Indicators represent least significant difference = 22.3 mdeg/Nm (alpha error 0.05)

results due to inconsistent defect classifications and varying degrees of bone loss. However, the highest mechanical failure rate of revision stems seems to be related to the largest femoral defect [3] and no conclusions can be drawn for a mechanical influence of strut grafts.

The present study confirmed differences in the mentioned fixation philosophies of the examined prostheses. The load transfer of the S-ROM® is mainly intended to proceed via the porous coated sleeve [5, 15, 16, 24]. Rotational stability can be achieved distally with the fluted, polished and split stem [19]. In accordance with this concept, we detected smaller relative movements of the proximal stem in bearing conditions I and III compared to the MRP®, which mainly transfers loads to the diaphysis due to its conical and fluted stem [17, 18]. Therefore less micro-movements of the distal stem were detected for the MRP®.

These fixation philosophies also explain different fixation behaviors due to bearing conditions. As the S-ROM® fixation was highly compromised due to weakening of the proximal bony lid, the stability of stem correspondingly decreased (Fig. 5). By applying strut grafts to the deficient bone, the stem partly tends to achieve its fixation again, even though this was not significant.

To transfer our results to in vivo situations, we calculated the interface slipping for a physiological loading of stems. Even though we could not find any significant effect due to a strut graft augmentation, all relative movements for both

stems were less than 150 μm , which is considered to be a critical limit for osteointegration [12–14]. Concurring with clinical observations [6, 8, 10], our results indicate that the stems themselves should provide a sufficient primary stability, as a stabilizing effect of grafts seems to be limited.

In contrast to the influence of the femoral bearing condition on primary stability of the S-ROM®, different results were obtained for the MRP®. A distinct change of movements was not obvious between bearing conditions I, II and III. We assume an impact of the diaphyseal fixation of the conical stem, which is not affected by the bony weakening. Interestingly, the goal to support primary fixation with strut grafts could not be proven so as to result even in a non-significant increase of relative movements at the proximal part of the MRP®. This may be due to the contact of strut grafts with the proximal stem, which is not primarily designed to provide fixation. At the distal part of the stem, which is fixed within the isthmus, no contact with grafts is provided. They are in contact with the bone more distally. As a consequence, the area of load transfer is bypassed and relative movements distal to this area are directed to the proximal part of stem. This may explain a slight decrease of primary fixation in this area.

Due to a different fixation philosophy of the S-ROM®, strut grafts had direct contact with the proximal part of prosthetic load transfer and therefore could augment the fixation to some extent.

Beyond implant stability, the influence of mechanical loading of the strut graft itself merits attention. In current literature [7, 8, 10], no distinct differences of graft union to host bone have been observed in relation to primary mechanical support of the implant by the allograft. Nevertheless, physiological stress [4, 8, 10] appears to enhance allograft remodeling and incorporation, and proximal resorption of the graft has been interpreted as stress shielding [4]. Thus transferring load to strut grafts, which are provided for primary support, might ensure the remodeling and incorporation of these grafts. This mechanical stress on the graft may be expected if contact with the prosthetic section of load transfer is provided as with the S-ROM®.

There are some limitations regarding a clinical transfer of our results. First, we were not able to evaluate the mechanical effects of biological processes such as temporary or partial bone changes [6, 11] during incorporation and remodeling of strut grafts. Osteointegration of the created bony lid and the biomechanical consequences that may ensue can not be addressed. The main technical limitation concerns the mechanical behaviour of artificial femora as they match rather “healthy” bones. Their elastic behaviour and interfragmentary friction are different to cortically thinned bones typically undergoing the studied type of revision surgery. Therefore, the reliability of these results might have been facilitated by

these femora, for which comparable biomechanical characteristics to human femora have been demonstrated with a low variability [25]. Nevertheless, in contrast to cadaveric femora, the consistent anatomical properties of synthetic femora allow the creation of femoral conditions in a highly standardized manner. A transfer of results to clinical situations should be possible, as in our perception the simulated femoral bearing conditions correspond to revision situations orthopaedic surgeons are commonly confronted with [1, 23]. As stabilization of transfemoral osteotomies [4, 7] and uncontained femoral defects [1, 2, 10] with strut grafts are recommended in such cases, the management of these revision situations is also consistent with current orthopaedic practice.

In addition, the extrapolation of relative micro-motions to human gait from normalized relative movements may be a further limitation. We cannot safely exclude a possible transition from static to dynamic friction in a range exceeding our experimental moment T_Z , perhaps resulting in a nonlinear behaviour.

However, with the experimental set-up used in this and previous studies [19, 20], reliable results have been achieved in different femoral bearing conditions, and clinical observations could be biomechanically substantiated.

Against our hypothesis we conclude that the primary stability of a revision stem can only be slightly augmented by strut grafts, if these support femoral defects in the area of prosthetic load transfer. But even this small effect was only observed if the allografts had direct contact with the prosthetic part of the load transfer.

Therefore, when stabilization of revision stems by strut grafts is indicated, orthopaedic surgeons should not only consider the localization of bone defect but also the prosthetic design and expected load transfer.

Transferring these results to clinical situations, it is uncertain whether revision stems faced with femoral defects secondary to a transfemoral osteotomy could be stabilized by strut grafts. Nevertheless, no adverse consequences for implant stability are to be expected. Therefore other indications which are beyond the scope of this study, such as stabilization of the osteotomy or restoration of bone stock, can justify the application of strut grafts in revision surgery.

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