The number of revision arthroplasties is increasing. This is due to the aging of the population and the increasing indications for primary implantation.\(^{1-3}\) The reasons for changing a prosthesis are many and varied. In addition to aseptic and septic loosening, periprosthetic fractures are becoming more common.\(^{4-6}\) In Germany, 11.6% of all endoprosthetic operations are revisions.\(^{7}\) If a prosthesis is loose, exchanging it with a prosthesis that has a stable stem for diaphyseal anchoring is indicated.\(^{8,9}\)

Cementless revision arthroplasty of the hip has become the gold standard, having excellent results. After such an operation (eg, an implantation for a hip replacement), the rate of re-revision is less than that after an operation involving a cemented prosthesis because the osseous prosthesis bed is not only somewhat loosened but also usually thinner, sclerosed, and enlarged.\(^{10}\) Moreover, the key technique to ensuring a substantial interface between bone and implant currently involves the press-fit procedure. Here, the intent to maintain the femoral revision stem’s stability together with less than 50 µm of micromotion per gait cycle has been fulfilled.\(^{11,12}\) The primary form...
of fixation for a cementless hip revision stem is a press-fit implant that allows for osseointegration along the entire length of the stem for most implants (excluding modular metaphyseal fixation stems). High-strength titanium compositions are usually used for cementless stems, which are more flexible than cemented stems. They have a more femur-like modulus of elasticity with physiological force transmission. To address any negative forces due to axes and rotation, the pre-tension within the metal–bone interface must be large enough to convert such forces into compression forces against the bone. The stem is designed to achieve a long-term secondary form of fixation via osseointegration.

Controversy surrounds revision arthroplasty of the knee joint and distal femur. The primary form of fixation for knee revision implants using a cementless stem involves a cemented condylar portion with a nonporous surface. This hybrid fixation technique involves cement in the metaphysis with the cementless stem serving to distribute forces away from the cemented portion of the implant. Equivalent results have been reported for cemented and cementless implantation of knee revision stems. However, the outcomes for knee revision implants have mainly been described as poor compared with those for hip revision stems. This difference in outcomes might be related to the different anchorage of cementless stems at the distal compared with the proximal femur. The specific geometry of the femur (tapering in the isthmus, position of the antecurvature, and so forth) may be responsible for this, with stem designs that are successful at the hip (Wagner profile) possibly being less successful at the distal femur. On an axial section of the proximal femoral diaphysis, the intramedullary canal has a round circumference. At the distal femur, this looks more oval.

The objective of this study was to evaluate the anchoring principles of different femoral revision stem designs in extended bone defect situations, taking into account the anatomical conditions of the proximal and the distal femur with their dissimilar bone profiles, and the resulting primary stability.
**Materials and Methods**

Four different cementless press-fit models, used clinically for revision in total hip and total knee arthroplasty, were studied. These were implanted into synthetic femurs. A kinked conical design (MP reconstruction prosthesis; Waldemar Link GmbH & Co KG, Hamburg, Germany) and a straight conical design (Revitan Hip System; Zimmer Biomet, Warsaw, Indiana) were used in the proximal femur, and a straight conical design (Endo-Modell SL; Waldemar Link GmbH & Co. KG) and a straight cylindrical design (NexGen; Zimmer Biomet) were used in the distal femur (Figure 1).

Each model had 5 samples. The axial/torsional stiffness and the migration resistance of each stem were analyzed. Standard operation procedures used corresponded to those used previously for press-fit stems and synthetic femoral bone tests. This involved use of an 8874 Axial-Torsion Servohydraulic Fatigue Testing System (Instron Germany GmbH, Darmstadt, Germany). With this, the axial force capacity for load cells is ±10 kN, torque capacity is ±100 Nm, and accuracy is ±0.5%. The results of the test were then summarized using the FastTrack console software provided by the multi-axis fatigue system (FTStartUp version 7.22, MAX version 9.2; Instron). The authors made sure that all hardware components were always intact and that all were used in exactly the same way from one implant to another. Twenty samples of one synthetic femur bone type (Model #3403 medium left, 4th generation, composite; Sawbone, Vashon, Washington) were used. Each of these had a centrum-collum-diaphyseal angle of 135°, a mid-shaft outer diameter of 27 mm, and an inner canal diameter of 13 mm standard.

Initially, the proximal (subtrochanteric) or distal (subcondylar) parts of the 4 groups of femurs were sawed prior to implantation of the stems into the synthetic bones. For this, the point of resection was based on the stems’ nominal length and their diameter in such a manner that intramedullary interfaces between the bone and the fluting part of the stems were retained as accurately as possible. As previously explained in the operating instructions, the femur’s inner canals were reamed so as to match the stem diameter of each implant’s design. Femurs were prepared with the reamer until endosteal contact was achieved, which is usually 1 mm below the desired diameter of the prosthesis. Subsequently, in accordance with the press-fit technique, the various stems were inserted into the femora. The authors restored the original length of the femur and implanted the stems at a concentric depth. Finally, to assess the interface between bone and metal, each sample was analyzed by scanning with a computed tomography device (Bright-Speed Performix 16 SI; General Electric Healthcare, Munich, Germany) (Figure 2). Slice thicknesses of 0.625 mm were made for this purpose. For each image, the proportion of contact between implant and inner cortical bone was determined as a percentage of the entire circumference of the implant (0%, 25%, 50%, 75%, or 100%). A circumference contact between implant and cortical bone was judged to be 100%; if the implant had been placed in the medullary cavity without any contact to cortical bone, this was 0%. To determine the anchored implant length in millimeters for each case, the image proportions (0%, 25%, 50%, 75%, or 100%) were each multiplied by the slice thickness. The authors then evaluated the data using the IMPAX EE CD Viewer (Agfa Health Care GmbH, Bonn, Germany).

Next, the femoral longitudinal axes including the stems were placed into the multi-axial clamping system of the servohydraulic testing machine. To analyze the axial stiffness of the press-fit femoral stems as precisely as possible, a simulation of biomechanical conditions was attempted. To take into account force transmission in the distal femur (30% medial and 70% lateral), the axes of the femurs were arranged in 18° adduction for the proximal stems of the revision total hip arthroplasties. The same procedure was performed using a 10° adduction for the proximal stems of the revision total knee arthroplasties. As already performed in previous investigations, this resembled the single-legged phase of walking. The trochanteric or the condyle region was fixed in a negative imprint as a vice, depending on the sample used. The end of the stem was inserted into an applicable femoral head. After a compatible inlay
had been fixed onto a metal plate of the testing machine, the force was transmitted to the sample (Figure 3A). Next, loads were applied vertically to the femoral heads. A linear ramp-up/ramp-down wave with 10,000 cycles and a load of 2.4 kN was used for this. To simulate the load of a normal gait cycle, the force was applied to the specimen by a 2-phase waveform (Figure 4).

Torsional stiffness was evaluated by positioning the long axes of the femurs vertically in the frontal and sagittal planes. To ensure that the angle was correct, the authors used a leveling gauge, which was held in a position relative to the loading machine. The stems’ ends were first fixed onto the metal plate with a screw. Next, torsional loads with a linear ramp-rotation were applied using 10,000 cycles with 10 Nm and an axial pre-load of 0.8 kN (Figure 3B).

Mann–Whitney tests were conducted for comparison of the axial and torsional stiffness as well as the migration among the various implants. Significance was set at $P < .05$.

**Results**

Both in the proximal femur and in the distal femur, the conical stems showed a combination of circumferential conical and 3-point anchorage with comparable anchorage lengths (Table 1).

The only difference between the stem designs was the fixation of the straight stem closer to the joint compared with the kinked stem (Figure 5). In contrast to this, in the distal femur, the cylindrical stem was anchored significantly shorter circumferentially (29.13±3.24 mm vs 44.13±3.08 mm, $P < .05$), beginning immediately at the resection plane and practically without 3-point impaction.

Despite different anchorage, the implants migrated comparably and without asymptotes (Figure 6). The data could be best approximated with a logarithmic function. The maximum migration after 10,000 gait cycles was 0.51±0.16 mm for the proximally anchored kinked stem, 0.58±0.28 mm for the proximally anchored straight conical stem, 0.50±0.39 mm for the distally anchored straight conical stem, and 0.35±0.05 mm for the distally anchored cylindrical stem.

No differences were found between the various designs and anchorage principles in either axial or torsional stiffness (Table 2).

No correlation was found between fixation length and migration or axial and torsional stiffness of the investigated samples. This suggests that the minimum fixation length in cortical bone necessary for primary stability was achieved in all cases.

**Discussion**

The main finding of this study was that different stem designs anchored in differ-

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**Table 1**

<table>
<thead>
<tr>
<th>Contact Area</th>
<th>Proximal Kinked Stem</th>
<th>Proximal Straight Stem</th>
<th>Distal Tapered Stem</th>
<th>Distal Cylindrical Stem</th>
</tr>
</thead>
<tbody>
<tr>
<td>25% contact, mm</td>
<td>14.75±5.26</td>
<td>33.25±18.95</td>
<td>23.13±11.97</td>
<td>1.63±0.95</td>
</tr>
<tr>
<td>50% contact, mm</td>
<td>11.25±5.06</td>
<td>10.00±5.06</td>
<td>9.63±2.15</td>
<td>1.75±1.20</td>
</tr>
<tr>
<td>75% contact, mm</td>
<td>12.88±11.12</td>
<td>13.00±5.22</td>
<td>3.88±1.49</td>
<td>0.63±0.63</td>
</tr>
<tr>
<td>100% contact, mm</td>
<td>47.38±10.23</td>
<td>55.38±4.30</td>
<td>44.13±3.08</td>
<td>29.13±3.24</td>
</tr>
<tr>
<td>Total interface, mm</td>
<td>86.25±1.25</td>
<td>111.63±29.38</td>
<td>80.75±10.86</td>
<td>33.13±3.64</td>
</tr>
<tr>
<td>100% contact in relation to total interface</td>
<td>54.97±12.02%</td>
<td>51.73±10.73%</td>
<td>55.78±10.93%</td>
<td>88.07±5.34%</td>
</tr>
</tbody>
</table>
ent femoral canal geometries achieved comparable primary stability.

Cementless revision arthroplasty using distal fixation has been a proven technique for several years. This method relies on the principles of press-fit and multiple-point impact.27-32 Anchoring using the press-fit is caused by a conical impaction of the implanted stem into a prosthesis bed of the femur that had been prepared in advance.33,34 Multiple-point impaction results from the incongruence of stem design and femur form, which leads to contact between the inner cortical bone and the stem of at least 3 points.

Whereas cemented stems produce a reproducible and thereby predictable fixation, cementless fixation differs significantly between different materials and designs. The predominant fixation principle is based on a cortical force distribution, in contrast to a solely cancellous bone fixation (eg, Natural Knee II; Zimmer Biomet). Cortical anchoring stems are made from titanium or cobalt-chrome and are available in solid or fluted designs.35,36

The amount of osseointegration depends on the cross-section design, the surface roughness, and the individually achieved press-fit.

Compared with those reported for the proximal femur, results reported for cementless knee revision stems have been predominantly poorer.13-18 This may be due to the specific geometry of the femur, with different anchorage of cementless stems at the distal part compared with at the proximal femur. Thus, stem designs that are successful at the hip produce less favorable outcomes at the distal femur. However, the clinical differences observed cannot be attributed to the primary stability of the investigated stems. Despite the considerably different anchorage lengths, no difference in migration or stiffness was found.

This is contrary to the results of Zdero et al.,37 who showed correlations between the femur–stem contact area (interface area) and the lateral stiffness, axial stiffness, and torsional stiffness in 4 different stem systems.37 This may be because, in contrast to the current study, they investigated slotted stems with a star profile and stems with a rounded profile. All

![Figure 5](image1.png)

**Figure 5:** Intramedullary length and particular contact area of the 4 groups (0= resection line).

![Figure 6](image2.png)

**Figure 6:** Example for migration of the proximal kinked stem design.

<table>
<thead>
<tr>
<th>Stem Design</th>
<th>Axial Stiffness, kN/mm</th>
<th>Torsional Stiffness, Nm/°</th>
</tr>
</thead>
<tbody>
<tr>
<td>Proximal kinked</td>
<td>0.197±0.058</td>
<td>6.484±0.173</td>
</tr>
<tr>
<td>Proximal straight</td>
<td>0.167±0.061</td>
<td>6.418±0.663</td>
</tr>
<tr>
<td>Distal tapered</td>
<td>0.176±0.091</td>
<td>6.258±0.118</td>
</tr>
<tr>
<td>Distal cylindrical</td>
<td>0.257±0.065</td>
<td>6.192±0.068</td>
</tr>
</tbody>
</table>

*p not significant in all cases.

<table>
<thead>
<tr>
<th>Mean±SD*</th>
</tr>
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<tbody>
<tr>
<td>Stiffness</td>
</tr>
<tr>
<td>-----------</td>
</tr>
<tr>
<td>Stem Design</td>
</tr>
<tr>
<td>Proximal kinked</td>
</tr>
<tr>
<td>Proximal straight</td>
</tr>
<tr>
<td>Distal tapered</td>
</tr>
<tr>
<td>Distal cylindrical</td>
</tr>
</tbody>
</table>

*p not significant in all cases.
of the stems investigated in the current study had an angular star profile and were not slotted. The good clinical outcomes achieved with these stems indicated that they all function in their individual design and that the respective fixation principles, fixation lengths, and primary stabilities determined in this study were sufficient.

In the investigated osseous defect model, the stem design (conical vs cylindrical), not the geometry of the femoral canal (proximal vs distal), was decisive regarding circumferential anchorage length. Both in the distal femur and in the proximal femur, the conical stems showed a combination of conical and 3-point anchorage. A stem impaction close to the joint was achieved with a straight conical design at the proximal femur and with a straight cylindrical design at the distal femur.

To the authors’ knowledge, the use of a synthetic bone model for simulating in vivo conditions was the most limiting facet of the current study. Synthetic bone has no soft tissue and no properties similar to the decreased bone density and cortical thinning of osteoporotic bone. A synthetic femur’s attributes are of course different from those of a real, human femur in revision arthroplasty settings. This will entail the amount of bone stock, interfacial friction, long-term influence of bone ingrowth, axial deviations, periprosthetic fractures, and other characteristics. This holds true when the objective is to compare various stem designs.

CONCLUSION

For the conical stems, it is postulated that there are reserves available for achieving a conical–circular fixation as a result of the large contact length. For the cylindrical stems, a much shorter anchorage length was sufficient in the distal femur to achieve a primary stability comparable to that with the conical stems. However, because of the short contact length at the distal femur, only a small reserve for a stable anchorage can be assumed. This should be taken into account in clinical application, and a cortical contact length of at least 30 mm should be achieved in the distal femur for cylindrical knee stems. In the proximal femur, the length of the anchorage for conical hip stems should be at least 55 mm.

REFERENCES

26. Ferguson PC, Zdero R, Schemitsch EH, Deheshi BM, Bell RS, Wunder JS. A biomechanical evaluation of press-fit stem constructs for tumor endoprosthetic reconstruc-


