Cemented short-stem total hip arthroplasty

Azari, Fahimeh; Sas, Amelie; Kutzner, Karl Philipp; Klockow, Andreas; Scheerlinck, Thierry; van Lenthe, G Harry

Published in:
Journal of Orthopaedic Research

DOI:
10.1002/jor.24887

Publication date:
2021

Document Version:
Accepted author manuscript

Citation for published version (APA):
Azari, F., Sas, A., Kutzner, K. P., Klockow, A., Scheerlinck, T., & van Lenthe, G. H. (2021). Cemented short-stem total hip arthroplasty: characteristics of line-to-line versus undersized cementing techniques using a validated CT-based Finite-Element-Analysis. *Journal of Orthopaedic Research*, *39*(8), 1681-1690. https://doi.org/10.1002/jor.24887

Cemented short-stem total hip arthroplasty: characteristics of line-to-line versus undersized cementing techniques using a validated CT-based Finite-Element-Analysis

Fahimeh Azari,[1] Amelie Sas,[1] Karl Philipp Kutzner,[2] Andreas Klockow,[3] Thierry Scheerlinck,[4] G. Harry van Lenthe[1]
[1]Biomechanics Section, KU Leuven, Celestijnenlaan 300C, 3001 Leuven, Belgium, [2] Department of Orthopaedic Surgery and Traumatology, St. Josefs Hospital Wiesbaden, Beethovenstr. 20, 65189 Wiesbaden, Germany, [3] Mathys Medical, Robert Mathysstr. 5, 2544 Bettlach, Switzerland, [4]Department of Orthopedic and Trauma Surgery, UZ Brussel, Laarbeeklaan 101, 1090 Brussels, Belgium

Correspondence to: Harry van Lenthe (T: +32 16 32 25 95; F: +32 16 32 89 99; E-mail: harry.vanlenthe@kuleuven.be)

Author’s contribution

FA contributed to research design, acquisition, analysis and interpretation of data, and drafting the paper. GHVL contributed to research design, acquisition and interpretation of data and revising of the paper. TS and KPK contributed to research design, acquisition and interpretation of data and revising of the paper. AS contributed to acquisition and analysis of data and revising of the paper. AK contributed to acquisition and interpretation of data. All authors have read and approved the final submitted manuscript.

Abstract

Short-stems are becoming increasingly popular in total hip arthroplasty since they preserve the bone stock and simplify the implantation process. Short-stems are advised mainly for patients with good bone stock. The clinical use of short-stems could be enlarged to patients with poor bone stock if a cemented alternative would be available. Therefore, this study aimed to quantify the mechanical performance of a cemented short-stem and to compare the 'undersized' cementing strategy (stem one size smaller than the rasp) to the 'line-to-line' technique (stem and rasp with identical size). A prototype cemented short-stem was implanted in eight pairs of human cadaveric femora using the two cementing strategies. Four pairs were experimentally tested in a single-legged stance condition; stiffness, strength ,and bone surface displacements were measured. Subject-specific nonlinear finite element models of all the implanted femora were developed, validated against the experimental data, and used to evaluate the behavior of cemented short-stems under physiological loading conditions resembling level walking. The two cementing techniques resulted in non-significant differences in stiffness and strength. Strength and stiffness as calculated from finite element were 8.7% ± 16% and 9.9% ± 15.0% higher than experimentally measured. Displacements as calculated from finite element analyses corresponded strongly ($R^2 \geq 0.97$) with those measured by digital image correlation. Stresses during level walking were far below the fatigue limit for bone and bone cement. The present study suggests that cemented short-stems are a promising solution in osteoporotic bone, and that the line-to-line and undersized cementing techniques provide similar outcomes.

Keywords: short-stem, cementing technique, total hip arthroplasty, mechanical testing, finite element analysis

Introduction

Short-stems have been introduced as an alternative to conventional stems in uncemented total hip arthroplasty (THA), especially for young and active patients. Short-stems aim to preserve the proximal bone stock and simplify the implantation process.[1,2] Currently, short-stems of the newest generation have shown good clinical outcomes in the medium-term.[3,4] However, using uncemented short-stems in elderly patients with reduced bone quality increases the risk of postoperative periprosthetic fractures.[5] Thus, a cemented version would potentially offer a solution for patients with poor bone quality or uncommon anatomy.[6,7] Yet, clinical data on cemented short-stems are still limited and concerns have risen about the risk of periprosthetic fractures and long-term survival.[8] Currently, mainly two competing strategies for cementing a femoral hip implant are being used.[9] First, using a stem that is equal in size as the largest broach that fits the femoral canal. This "line-to-line" cementing technique results in a thin but significant cement mantle that corresponds mainly to the cement pressurized into the cancellous bone. Second, using a stem that is smaller ("undersized") than the largest broach. This results in a thicker cement mantle composed of a pure cement layer and a layer of cement pressurized into the cancellous bone.[10] In the line-to-line cementing technique, cement-bone interdigitation and areas of thin cement being supported by cortical bone provides excellent support to the cement mantle[9,11,12] and results in a promising long-term outcome.[13,14] In the undersized cementing technique, however, the thicker cement mantle reduces cement stresses[15] and reduces micro-motion at the cement-stem interface resulting in less cement cracks.[1,16,17] Hence, both the line-to-line and the undersized technique have demonstrated good longevity when using traditional stems, depending on the stem design.[10] Yet, it is not known what the mechanical consequences of these cementing techniques are when using short-stem designs. Therefore, the present study aimed to determine which cementing strategy, i.e., the line-to-line or undersized technique, is preferable in cemented short-stem THA. For that purpose, we determined which cementing technique would give the highest load to failure and the lowest bone and cement stresses. We also measured stiffness to evaluate potential differences in deformation behavior. We hypothesized that, in analogy to traditional stems, both techniques would result in similar fracture loads and that cement stresses will be lower when using the undersized technique.

Methods

Specimens

After approval from the UZ Brussels Ethics Board (approval number of the project: B.U.N 143201733043), eight pairs of fresh frozen human cadaveric femora (i.e., 16 femora in total) were obtained from the Anatomy lab of the Brussels University Hospital (UZ Brussel). Donors' age at death was 73.5 ± 5.9 years, height was 163.7 ± 7.4 cm, weight was 52.8 ± 4.3 kg and body mass index (BMI) was 19.8 ± 2.4 kg/m^2. All specimens had been kept intact, were fresh frozen at −20 °C, and thawed for implantation and mechanical testing. The left and right femur of each pair were prepared for implantation with the optimys short-stem (Mathys Ltd., Bettlach, Switzerland) using broaches of identical size. For each pair, the line-to-line cementing technique was used in one femur, and the undersized technique in the contralateral one. Specimen allocation was random and, preparation and implantation were performed by two experienced orthopedic hip surgeons (TS and KPK). Information regarding the specimens and cementing method is outlined in Table 1.

Medical imaging

Computed tomography (CT)-scanning was performed two times for all 16 specimens: first, of the intact bone; and second, after implantation. CT scanning parameters were: 0.60 mm slice thickness and 0.21 mm pixel size. X-ray tube current, energy level, and exposure time were 160 mA, 140 kV, and 1000 ms, respectively. A dipotassium hydrogen orthophosphate (also referred to as dipotassium phosphate; K_2HPO_4) calibration phantom (SN: 3931, part No: 13002) was scanned together with the femora as a reference for quantifying bone density from the CT images. Image data sets were reconstructed using an ultra-sharp (B80) reconstruction kernel. Bone quality was assessed by dual-energy X-ray absorptiometry (DXA), using a Hologic scanner (Hologic, MA) of the femora before broaching.

Mechanical testing

Mechanical testing was performed in four pairs of specimens (specimens 5 to 8) according to the workflow of Sas *et al*.[18] To prepare specimens for mechanical testing, soft tissues were removed from the femora. The femora were shortened such that all specimens had an equal size of 25 cm as measured from the tip of the greater trochanter. The distal part of each specimen was embedded in a stainless steel holder using polymethylmethacryate (PMMA, Technovit 3040, Heraeus Kulzer, Germany). The height of the holder was 5 cm. The anterior aspect of each femur was painted with a white background spray layer, and a random black speckle pattern was applied with an airbrush to obtain a unique pattern on each sample that could be used for tracking displacements during mechanical testing with digital image correlation (DIC). A 6 by 6 pixel speckle size was aimed for, which resulted in a physical speckle size of approximately 0.4 mm for the adopted camera setup.[19] The distal fixation was rigidly mounted onto the INSTRON 3367 quasi-static testing machine so the femoral axis had an angle of 12° with respect to the loading axis (Fig. 1). Load was applied on a prosthetic head attached to the stem and the contact point between the loading plate and the head was greased to minimize the friction and to avoid undesired shear loading. Mechanical loading was applied until macroscopic failure (fracture of the bone) occurred. The mechanical tests were force-driven at a speed of 10 N/s. Prior to the test, a preload of 50 N, followed by 20 sinusoidal preconditioning cycles (50-500 N, 1 Hz) were applied to the prosthetic head. Actuator displacement and load were recorded at 5 Hz. The strength of the femora was defined as the maximum force magnitude as taken from the force-displacement curve. The stiffness of the bone-implant construct was determined as the steepest slope for a 20% portion of the force-displacement curve.[20] The entire experiment was recorded with two cameras (Grasshopper3, Flir Systems Inc., 5 Mpx) that captured images at 5 frames per second (fps). This frame rate suffices to capture the deformation behavior of the specimens, but does not capture (sudden) fracture in brittle materials like bone; yet, this study focused on the use of cemented implants before fracture, hence, a detailed quantification of the exact fracture pattern was not needed. Light intensity and the shutter time were adjusted to have good contrast. After mechanical testing, DIC was performed using the software tool *Vic-3D* 8.0.0 (Correlated solutions Inc., Irmo, SC) for each specimen to obtain a discrete displacement field for each recorded frame. DIC was calculated with a subset size (that is, the size of the area used to evaluate the gray level pattern) of 25 px and a step size (defined as the number of pixels by which the subset is shifted to calculate the displacement field) of 5 px. All the frames were compared to the same reference image, taken in the unloaded and undeformed configuration.

Image processing

The CT scans were processed to develop specimen-specific CT-based FE models of all sixteen bone-implant specimens, mimicking the same loading configuration as in the experimental tests. Due to implant-related artifacts, the scans of the implanted femora could not be used directly for FE analysis, as these artefacts prevented proper quantification of bone density. Hence, data from two separate CT scans were combined. Specifically, the intact bone geometry was obtained from the CT scans of the intact bone. Semi-automatic, threshold-based segmentation was performed in Mimics 22.0 (Materialise NV, Leuven, Belgium) to construct solid 3D models of the intact femurs. Geometric data on the stem and the cement were retrieved from the CT scans of the implanted femora. Specifically, the stems were segmented by simple thresholding. The cement was manually contoured and the segmentation was verified by an experienced orthopedic surgeon (TS).

Prior to combining the stem and cement data with the data of the bone, a registration of the implanted 3D bone model on the intact bone model was performed using 3-matic 14.0 (Materialise NV). The most proximal part of both 3D models was removed since this differed and would impede the registration. Subsequently, a rigid registration was performed to align the implanted femur (stem and cement were moved along) with the intact femur. The registered 3D models of the cement and the stem were overlaid on the intact femur scan and converted into a mask. A mask of the cortical bone was subtracted from the cement mask to assure that there was no overlap between cement and cortical bone. Region growing and morphologic closing operations were performed on the mask to remove floating parts and to remove sharp features. Afterwards a 3D model of the cement was generated based on this mask. The head of the intact bone was resected using a cutting plane that was fitted to the resection plane of the implanted femur. Finally, wrapping and smoothing operations were applied to refine our 3D model of the bone and cement.

The volume of the bone cement, including all cement proximal to the stem tip, was quantified by simply counting the voxels of the segmented data set multiplied with the volume of the voxel. We used CTAn 1.19.4.0 (Bruker, Kontich, Belgium) to calculate the cement thickness using the sphere-fitting algorithm.[21]

Mesh creation

A "non-manifold" assembly was performed between the bone, the cement and the stem to assure that the nodes at the interface of the parts matched exactly. Next, volume meshes were created using linear tetrahedral elements (C3D4) with a maximal edge length of 3 mm for the femur and the stem and an edge length of 2 mm at the cement and interfaces. A convergence analysis showed that for FE analysis of the proximal femur, this mesh size gives accurate results. Finally, material properties were assigned in Mimics, based on the Hounsfield units (HU) from the CT scan. Since calibration phantoms for the CT images of specimen pairs 1 to 4 were not available, the linear relation was estimated by calibrating the HU values from CT against bone mineral content (BMC) measurements from dual energy X-ray absorptiometry (DXA) scans following the work of Takada *et al.*[22] and the instructions from the Hologic manual.[23] For the specimen pairs 5 to 8, the linear relation was estimated by calibrating the HU values from CT against bone mineral density (BMD) measurements from the calibration phantom. The HUs were divided over 40 material categories and converted to ash density for implementation of the non-linear material behavior according to Keyak *et al*.[24] The steel implant was assumed to be a uniform material with a Young's modulus of 180 GPa. The cement mantle was assigned a Young's modulus of 3 GPa.[11] The Poisson coefficient of all materials was set to 0.3. The FE analysis was performed in Abaqus Standard 2017 (Dassault Systèmes, Vélizy-Villacoublay, France) using the non-linear geometry solver. Assuming a proper fixation of the cement to the bone and the implant, the interfaces were tied together in accordance with other studies.[25]

Model validation

The data from the mechanical tests were used to evaluate the accuracy of the FE models for specimens 5 to 8. To mimic the experimental set up for the validation purpose, the distal part of the femur diaphysis was positioned under an angle of 12° with the vertical axis. The load was applied at the implant head center and distributed over the top surface of the stem. The load was applied as displacement (6 mm) in 12 steps of uniform distribution.[26] The nodes of the distal elements, positioned more than 20 cm distal to the tip of the greater trochanter, were restrained to simulate the fixation of the distal part of the femur. The agreement between the displacement as calculated by FE analyses and the displacements measured experimentally (using DIC) was evaluated at a force of 5 kN. Validating at this force provides us with FE data which are still in the linear elastic range, and representative of the stiffness of the bone-implant systems, yet, also represents a considerable force resulting in measurable deformation in the experimentally tested femora. Ordinary least squares regression analysis between the experimental and the FE data was performed and coefficients of determination (R^2), root mean square error (RMSE), and slope were calculated. Statistical analysis was performed using Bland-Altman plots and paired t-tests to analyze the agreement between the strength and stiffness data as calculated by FE analyses and measured by mechanical tests. A p-value smaller than 0.05 was considered significant.

Simulating in vivo loading

In vivo loading conditions were simulated by subjecting the models to hip contact and muscle forces representing walking loads according to Heller *et al*.[27] These analyses were performed for the models of the specimen pairs 5 to 8, because only for these specimens information about body weight, required to quantify the joint and muscle loads, was available. The in vivo forces were defined with respect to the patient-specific coordinate system. The hip contact force was applied at the implant head center and distributed over the top surface of the stem. The muscle forces were distributed over node sets including the ten closest nodes to the muscle force application points.[7] The nodes of the most distal elements were restrained to prevent rigid body motions. Additionally, a constraint was imposed to the head center such that it could only translate along the axis joining the hip and knee center. This constraint leads to a more physiological deflection of the femoral head as demonstrated by Speirs *et al*.[28]

Assessment of cementing technique

The mechanical consequences of the cementing technique were evaluated using specimen-specific FE analyses of the 16 implanted stems and by mechanical testing of 8 implanted stems. For each donor one stem was implanted using the line-to-line method and a one-size undersized stem was implanted in the undersized technique. As such, variability in bone geometry and density were minimized, allowing us to evaluate the effect of the cementing techniques.

Results

Due to an unfortunate human error specimen 7R broke prior to testing; this sample was excluded from the validation process.

Cement volume and thickness

The amount of cement in the undersized cases was non-significantly ($p > 0.05$) higher than in the line-to-line cases; on average 1.6 cm^3 (Table 2). The average cement thickness was slightly, higher for the undersized cases (Table 2). The cement around the stem was not limited to a (small) volume just around the stem, but it penetrated into the cavities present in the trabecular bone (Fig. 2). Cement thickness was varying along the length of the stem (Fig. 2).

Mechanical testing

For all the specimens we found that the force-displacement curve consisted of an initial linear part, followed by a second linear part with a higher slope than the first one (Fig. 3). Hence, stiffness was always based on the second linear portion of the curve. The transition of the two linear sections occurred at a displacement between 1 and 2 mm. The second linear part ended with a sudden drop in the measured force, indicating failure of the construct. Strength and stiffness did not differ significantly between line-to-line and undersized cases (Table 3).

Validation of the finite element models

Strength and stiffness as determined from the FE models agreed well with the experimentally measured data (Fig. 4). Paired t-tests, showed non-significant differences for strength ($p > 0.05$) and for stiffness ($p > 0.05$).

An excellent agreement between displacement data from FE analysis and mechanical testing was found (Fig. 5) with $R^2 > 0.97$ and RMSE < 20 µm for all specimens.

Mechanical behavior under physiological loading

The validated FE models showed that stresses in the bone and cement during level walking were always less than 24.9% of the yield stress and 29.2% of the fatigue strength.[29] To assess whether cement failure due to compressive and tensile stresses would occur, the minimum and maximum principal stresses were calculated for all elements and indicated that no cement failure due to compressive and tensile stresses is expected (with a safety factor equal to 3.5).

Comparison between line-to-line and undersized cases

Strength and stiffness data showed very similar behavior for the undersized and the line-to-line technique (Fig. 6a and Fig. 6c, respectively). Bland-Altman plot (Fig. 6b and Fig. 6d, respectively) and paired t-test demonstrated non-significant differences in strength ($p > 0.05$) and stiffness ($p > 0.05$).

Discussion

While at present short-stems are advised mainly for patients with sufficient bone stock, the development of a cemented calcar-guided short-stem for patients with poor bone quality may be a useful complement in THA. In this study we evaluated the biomechanical characteristics of cemented short-stem prototypes and the effect of different cementing techniques ('undersized' versus 'line-to-line'). We measured experimentally that the undersized and line-to-line cementing technique gave similar fracture loads; a finding that we confirmed with finite element models. The finite element models also showed that the maximum cement stresses were 11.7% lower when using the undersized technique.

This study is unique in that we used a combined experimental-computational approach, whereas similar studies have been limited to either *in silico* modeling or *in vitro* experiments. In our study we used stems that were implanted in left and right femora from the same donor, hence, the stems were placed in bones with similar geometry, density, and mechanical properties. The cadaveric bones were mainly osteopenic (T-score < -1.0; 8/16, 50%) and osteoporotic (T-score < -2.5; 5/16, 31%) hence, 81% of the bones we tested reflect the target population. We used a stem design that was identical to the clinically successful uncemented optimys stem,[30-33] the only difference being a polished surface and steel material instead of titanium. A further strength of our study was that we used identical stem designs in both femora; the only difference was that the undersized stem was one size smaller than the stem that was cemented line-to-line.

From our CT data on eight pairs of cemented femora (N = 16 in total), we found that the cement volume of the undersized cementing technique was, on average, 5.8% larger than that of the line-to-line technique. However, the difference was smaller than expected from the difference in stem size, which was on average 2.6 cm^3. This suggests that in the line-to-line case about 1.0 cm^3 extra bone cement is pressurized in the cancellous bone. We hypothesize that this is the result of slightly higher pressures in the cement when inserting the thicker stems in the line-to-line scenario. Cement distribution was also similar in both cementing techniques. The average cement thickness of the undersized technique was 4.0% larger than that of the line-to-line technique.

The biomechanical effects of cementing technique were evaluated using specimen-specific FE models of the 16 implanted stems. Half of the models were validated against experimentally measured data. In our study, we found a very good agreement between mechanical tests and FE analysis for both stiffness and strength. The strength, quantified by the FE models (7.9 kN ± 0.9), was 9.9% ± 15.0% higher than that of the experimental tests (7.2 kN ± 1.3); the stiffness, quantified by the FE models (2.6 kN/mm ± 0.6), overestimated the measured stiffness (2.3 kN/mm ± 0.5) by 8.7% ± 16%. Displacement data calculated at the surface of the FE models also strongly matched the DIC measurements during mechanical testing (for all specimens $R^2 > 0.97$, RMSE < 20 µm). Yet, a substantial offset was noted between the measured displacements and those obtained with FE. Using DIC we could show that this discrepancy was caused by movement of the stem inside the bone, most likely related to creep of the bone cement. In the DIC data, the vertical displacement pattern was varying along the stem length compared to the bone length, showing stem movement inside the bone. We quantified the movement of the stem relative to the bone and evaluated this as a function of the applied load. We saw that this relative motion showed a bi-linear behavior, which perfectly matched the bi-linear behavior of the bone-cement-implant system (Fig. 3)". We hypothesize that the bi-linear behavior of the experimentally measured force-displacement curves is related to non-linear (creep) behavior of the bone cement. Note that due to preconditioning any potential settling of the bone-implant construct inside the clamps had been removed.

The experimental protocol in our study closely relates to a recent biomechanical study that also determined the primary stability of the same cemented short-stem design.[6] In line with our study, the authors also found very small and non-significant differences between the strength of the implanted femora after a line-to-line or undersized implantation. Yet, the strength as measured in our study was substantially higher than in the earlier study by Kutzner *et al*.[6] This may be related to slight differences in the experimental set-up. Whereas we tested the femora under 12 degree of inclination, the inclination angle was 8 degrees in the study by Kutzner *et al*. Furthermore, also the length of the femora differed (25 cm in our study compared to 37 cm). In the present study, we demonstrated that the mechanical behavior of the undersized and line-to-line stems was very similar; only small and non-significant differences in stiffness (average difference of 3.5% ± 3.0%) and strength (average difference of 2.3% ± 1.9%) were found between the femora from each pair.

Under physiological loading conditions acute and fatigue failure of the bone and of the cement is very unlikely and both techniques performed similarly from a mechanical point of view. From a clinical perspective, we would prefer the line-to-line technique. First, because the stem is guided into the broached cavity by cortical contact making centralizing devices unnecessary. Second, because the stem is stabilized by cortical contact avoiding micro-movements during cement curing.[6] And finally, because a higher rotational stability could be expected.[11]

There are also some limitations to this study. First, we only performed mechanical testing of four pairs of femora (specimens 5 to 8). One of the specimens (7R) fractured in the preparatory phase, leaving 7 specimens for validation. However, and despite the small number of specimens, we found a good agreement between experimental measurements and computational models. A second limitation is the restricted physiological loading model we used. We included only a limited number of muscles in the model.[27] Also, interface between bone, implant, and cement were modeled by tied constraints, which might differ from reality. However, according to similar studies[34,35] the influence of the implant-bone interface on the reported results is negligible.

In summary, we experimentally validated a CT-based FE method for the assessment of bone strength and stiffness of a cemented short-stem total hip arthroplasty model. We conclude that the line-to-line technique withstands similar loads as the undersized technique, and that both are unlikely to fail under normal physiological loading. Regardless of the specific cementation technique, cemented short-stems appear promising in patients with low bone quality. As both cementing technique behave similarly from a mechanical point of view, we prefer the line-to-line technique from a clinical point of view.

Acknowledgments

This project was funded by Mathys (Bettlach, Switzerland). We thank the Department of Anatomical Research Training and Education (ARTE) of the VUB (Brussels, Belgium) for providing human cadaveric femora and Department of Radiology of the Universitair Ziekenhuis Brussel for performing CT scans of the specimens.

References

1. Morellato K, Grupp TM, Bader U, et al. 2018. Standard and line-to-line cementation of a polished short hip stem: Long-term in vitro implant stability. J. Orthop. Res. 36(10):2736–2744.

2. Burchard R, Braas S, Soost C, et al. 2017. Bone preserving level of osteotomy in short-stem total hip arthroplasty does not influence stress shielding dimensions - A comparing finite elements analysis. BMC Musculoskelet. Disord. 18(1):1–7.

3. Schnurr C, Schellen B, Dargel J, et al. 2017. Low Short-Stem Revision Rates: 1-11 Year Results From 1888 Total Hip Arthroplasties. J. Arthroplasty 32(2):487–493.

4. Schilcher J, Ivarsson I, Perlbach R, Palm L. 2017. No Difference in Periprosthetic Bone Loss and Fixation Between a Standard-Length Stem and a Shorter Version in Cementless Total Hip Arthroplasty. A Randomized Controlled Trial. J. Arthroplasty 32(4):1220–1226.

5. Gkagkalis G, Goetti P, Mai S, et al. 2019. Cementless short-stem total hip arthroplasty in the elderly patient - Is it a safe option?: A prospective multicentre observational study. BMC Geriatr. 19(1):1–11.

6. Kutzner KP, Freitag T, Bieger R, et al. 2018. Biomechanics of a cemented short stem: Standard vs. line-to-line cementation techniques. A biomechanical in-vitro study involving six osteoporotic pairs of human cadaver femurs. Clin. Biomech. 52(January):86–94.

7. Sas A, Pellikaan P, Kolk S, et al. 2019. Effect of anatomical variability on stress-

shielding induced by short calcar-guided stems: Automated finite element analysis of 90 femora. J. Orthop. Res. 37(3):681–688.

8. Hamadouche M, Jahnke A, Scemama C, et al. 2015. Length of clinically proven cemented hip stems: State of the art or subject to improvement? Int. Orthop. 39(3):411–416.

9. Scheerlinck T, de Mey J, Deklerck R, Noble PC. 2006. CT analysis of defects of the cement mantle and alignment of the stem. J. Bone Jt. Surg. - Ser. B 88(1):19–25.

10. Scheerlinck T, Casteleyn P-P. 2006. The design features of cemented femoral hip implants. J. Bone Joint Surg. Br. 88(11):1409–1418.

11. Janssen D, Van Aken J, Scheerlinck T, Verdonschot N. 2009. Finite element analysis of the effect of cementing concepts on implant stability and cement fatigue failure. Acta Orthop. 80(3):319–324.

12. Alfaro-Adrián J, Gill HS, Murray DW. 2001. Should total hip arthroplasty femoral components be designed to subside? A radiostereometric analysis study of the Charnley Elite and Exeter stems. J. Arthroplasty 16(5):598–606.

13. Dairaku K, Ishii M, Kobayashi S, et al. 2010. Initial Evaluation on Subsidence of Cemented Collarless Polished Tapered Stem Applied to the Patients with Narrow Femoral Medullar Canal. Open Orthop. J. 4(1):147–151.

14. Langlais F, Kerboull M, Sedel L, Ling RSM. 2003. The “French paradox.” J. Bone Jt. Surg. - Ser. B 85(1):17–20.

15. Gunn E, Gundapaneni D, Goswami T. 2012. Effect of cement fill ratio in loosening of hip implants. Biomatter 2(2):87–93.

16. Mann KA, Gupta S, Race A, et al. 2004. Cement microcracks in thin-mantle regions after in vitro fatigue loading. J. Arthroplasty 19(5):605–612.

17. Estok DM, Orr TE, Harris WH. 1997. Factors affecting cement strains near the tip of a cemented femoral component. J. Arthroplasty 12(1):40–48.

18. Sas A, Van Camp D, Lauwers B, et al. 2020. Cement augmentation of metastatic lesions in the proximal femur can improve bone strength. J. Mech. Behav. Biomed. Mater. 104(January):103648.

19. Grassi L, Väänänen SP, Yavari SA, et al. 2014. Full-field strain measurement during mechanical testing of the human femur at physiologically relevant strain rates. J. Biomech. Eng. 136(11):1–8.

20. Benca E, Reisinger A, Patsch JM, et al. 2017. Effect of simulated metastatic lesions on the biomechanical behavior of the proximal femur. J. Orthop. Res. 35(11):2407–2414.

21. Bruker. 2010. CT-Analyser Version 1.10 user's guide.

22. Takada M, Grampp S, Ouyang X, et al. 1997. A new trabecular region of interest for femoral dual X-ray absorptiometry: Short-term precision, age-related bone loss, and fracture discrimination compared with current femoral regions of interest. J. Bone Miner. Res. 12(5):832–838.

23. CDC. 2007. Dual Energy X-ray Absorptiometry (DXA) Procedures Manual. Natl. Heal. Nutr. Exam. Surv. (January):1–115.

24. Keyak JH, Kaneko TS, Tehranzadeh J, Skinner HB. 2005. Predicting proximal femoral strength using structural engineering models. Clin. Orthop. Relat. Res. (437):219–228.

25. Abdullah AH, Asri MNM, Alias MS, Giha T. 2010. Finite element analysis of cemented hip arthroplasty: Influence of stem tapers. Proc. Int. MultiConference Eng. Comput. Sci. 2010, IMECS 2010 III:2241–2246.

26. Grassi L, Väänänen SP, Ristinmaa M, et al. 2016. How accurately can subject-specific finite element models predict strains and strength of human femora? Investigation using full-field measurements. J. Biomech. 49(5):802–806.

27. Heller MO, Bergmann G, Kassi JP, et al. 2005. Determination of muscle loading at the hip joint for use in pre-clinical testing. J. Biomech. 38(5):1155–1163.

28. Speirs AD, Heller MO, Duda GN, Taylor WR. 2007. Physiologically based boundary conditions in finite element modelling. J. Biomech. 40(10):2318–2323.

29. Lewis G. 1997. Properties of acrylic bone cement: State of the art review. J. Biomed. Mater. Res. 38(2):155–182.

30. Kutzner KP, Kovacevic MP, Roeder C, et al. 2015. Reconstruction of femoro-acetabular offsets using a short-stem. Int. Orthop. 39(7):1269–1275.

31. Kutzner KP, Freitag T, Donner S, et al. 2017. Outcome of extensive varus and valgus stem alignment in short-stem THA: clinical and radiological analysis using EBRA-FCA. Arch. Orthop. Trauma Surg. 137(3):431–439.

32. Van Oldenrijk J, Molleman J, Klaver M, et al. 2014. Revision rate after short-stem total hip arthroplasty. Acta Orthop. 85(3):250–258.

33. Loppini M, Grappiolo G. 2018. Uncemented short stems in primary total hip arthroplasty: The state of the art. EFORT Open Rev. 3(5):149–159.

34. Speirs AD, Heller MO, Taylor WR, et al. 2007. Influence of changes in stem positioning on femoral loading after THR using a short-stemmed hip implant. Clin. Biomech. 22(4):431–439.

35. Waanders D, Janssen D, Mann KA, Verdonschot N. 2010. The effect of cement creep and cement fatigue damage on the micromechanics of the cement-bone interface. J. Biomech. 43(15):3028–3034.

Fig. 1 The loading and boundary conditions as used in the FE analysis.

Fig. 2 Cement distribution demonstrating that the thickness of the bone cement layer is varying along the length of the stem. The light and dark green color show the bone cement and the implant, respectively. The dashed line shows the resection surface of the implanted femur. Cement distribution is shown at four standardized levels determined by implant size according to two-dimensional templates.

Fig. 3 Force-displacement curves for line-to-line and undersized specimens for one arbitrary specimen pair (pair 8) as determined from the mechanical tests and FE analysis.

Fig. 4 Bar charts (a, c) and Bland-Altman plots (b, d) of strength and stiffness, respectively for FE results against experimental data. Specimen 7R failed prior to testing, hence, 7R was excluded from the data.

Fig. 5 Ordinary least squares regression analyses on the displacement data for specimen 6R at a force of 5 kN.

Fig. 6 Scatter plots (a, c) and Bland-Altman plots (b, d) of FE strength and stiffness, respectively for the undersized cases against line-to-line cases. In the scatter plots the dashed line represents the line y = x. Specimen 7R failed prior to testing, hence, 7L and 7R were excluded from the data.

Table 1. Demographics of donors. NA: data not available.

Specimen	Age [y]	Sex	Body weight [kg]	Side	T-Score	Implant size	Cementing method
1	52	m	NA	L	-1.0	5	Undersized
				R	-0.8	6	Line-to-Line
2	94	f	NA	L	-1.5	2	Undersized
				R	-1.3	3	Line-to-Line
3	91	m	NA	L	-0.1	4	Undersized
				R	-0.2	5	Line-to-Line
4	74	f	NA	L	-2.5	6	Undersized
				R	-1.9	7	Line-to-Line
5	78	f	53.5	L	-1.2	3	Line-to-Line
				R	-1.5	2	Undersized
6	67	f	52.7	L	-2.9	5	Undersized
				R	-2.5	6	Line-to-Line
7	79	f	57.8	L	-2.7	8	Line-to-Line
				R	-2.8	7	Undersized
8	70	m	47.3	L	-2.9	4	Undersized
				R	-1.8	5	Line-to-Line

Table 2. Volume of the bone cement and bone cement distribution. Data indicate as mean ± SD [min – max].

	N	Cement volume (cm^3)	Cement thickness (mm)
Line-to-line	8	26.35 ± 8.35 [17.94 - 42.76]	7.51 ± 1.77 [5.64 - 10.40]
Undersized	8	27.92 ± 8.39 [15.83 - 41.21]	7.82 ± 2.07 [5.41 - 10.76]

Table 3. Strength and stiffness of the different bones, as measured from mechanical tests. Specimen 7R had failed prior to testing, the data from specimen pair 7 were excluded before calculating the mean value. Data indicate as mean ± SD [min – max].

	N	Measured strength (kN)	Measured stiffness (kN/mm)
Line-to-line	3	7.01 ± 1.68 [5.26 - 8.62]	2.42 ± 0.40 [2.01 - 2.80]
Undersized	3	7.80 ± 0.88 [6.81 - 8.48]	2.53 ± 0.50 [1.98 - 2.97]