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RESEARCH ARTICLE

A method to assess primary stability of acetabular components in association with bone defects

Ronja A. Schierjott[1,2] | Georg Hettich[1] | Alexandra Ringkamp[3] | Marc Baxmann[1] | Federico Morosato[4] | Philipp Damm[5] | Thomas M. Grupp[1,2]

[1]Research & Development, Aesculap AG, Am Aesculap-Platz, Tuttlingen, Germany

[2]Department of Orthopaedic Surgery, Physical Medicine & Rehabilitation, Ludwig-Maximilians-University Munich, Munich, Germany

[3]Department of Biomechatronics, Faculty of Mechanical Engineering, Technische Universität Ilmenau, Ilmenau, Germany

[4]Department of Industrial Engineering, School of Engineering and Architecture, Università di Bologna, Bologna, Italy

[5]Julius Wolff Institute, Joint Loading & Musculoskeletal Analysis, Charité–Universitätsmedizin Berlin, Berlin, Germany

Correspondence
Ronja A. Schierjott, Am Aesculap-Platz, 78532 Tuttlingen, Germany.
Email: ronja_alissa.schierjott@aesculap.de

Abstract

The objectives of this study were to develop a simplified acetabular bone defect model based on a representative clinical case, derive four bone defect increments from the simplified defect to establish a step-wise testing procedure, and analyze the impact of bone defect and bone defect filling on primary stability of a press-fit cup in the smallest defined bone defect increment. The original bone defect was approximated with nine reaming procedures and by exclusion of specific procedures, four defect increments were derived. The smallest increment was used in an artificial acetabular test model to test primary stability of a press-fit cup in combination with bone graft substitute (BGS). A primary acetabular test model and a defect model without filling were used as reference. Load was applied in direction of level walking in sinusoidal waveform with an incrementally increasing maximum load (300 N/1000 cycles from 600 to 3000 N). Relative motions (inducible displacement, migration, and total motion) between cup and test model were assessed with an optical measurement system. Original and simplified bone defect volume showed a conformity of 99%. Maximum total motion in the primary setup at 600 N (45.7 ± 5.6 µm) was in a range comparable to tests in human donor specimens (36.0 ± 16.8 µm). Primary stability was reduced by the bone defect, but could mostly be reestablished by BGS-filling. The presented method could be used as platform to test and compare different treatment strategies for increasing bone defect severity in a standardized way.

KEYWORDS

acetabular bone defect model, bone graft substitute, optical measurement, primary stability

1 | INTRODUCTION

Aseptic loosening is one of the main reasons for revision in total hip arthroplasty (THA).[1] Among others, it can be caused by excessive relative motion between implant and bone. It has been shown that relative motions above 150 µm prevent bone ingrowth and cause fibrous tissue formation, whereas relative motions below 40 µm enable osseointegration.[2,3] Absence of osseointegration eventually results in implant loosening and revision surgery, which is often accompanied by bone loss. This makes subsequent implant fixation even more challenging, and may require specific treatment strategies, such as bone grafts or bone graft substitutes (BGS) in combination with cemented or press-fit cups, as well as augments or revision implants. However, there are no standardized test methods yet available to test such specific treatment strategies.

Previous in vitro studies concerning primary implant stability were often focused on the assessment of different hemispherical cups for primary THA.[4–10] Typically, primary stability was assessed by lever-out, pull-/push-out, rotational tests, or in some cases under axial loading with relative motion measurement by linear variable differential transformers (LVDTs) or optical markers.

In revision THA, primary stability was so far analyzed by a limited number of studies in which a simplified acetabular defect was implemented in human donor specimens, Sawbone composite pelvises or surrogate models to test press-fit cups,[11,12] press-fit cups with augments,[13] reconstruction shells,[14,15] or cemented cups in combination with bone grafts or BGS.[16–19] To the authors' knowledge, no study has yet assessed primary stability of press-fit cups in combination with BGS defect filling in a surrogate acetabular test model.

More importantly, in the previous studies, defect shape was implemented by simplified procedures, often by using one acetabular reamer. This leads to simple defect shapes limited to a hemisphere in the acetabular ground[12] or other single spherical segments at the rim of the acetabulum.[13,20] However, clinical cases often show an irregular shape, which requires a less simplified defect implementation.

There is a large variation of acetabular bone defects, and in the previous in vitro testing studies different defect locations and sizes were used depending on the treatment strategy to be tested. This makes the comparison between treatment strategies difficult.

In studies about acetabular bone defects, it was found that bone loss is multidirectional in most cases[21–23] and that bone volume loss in the posterior column and the medial area often appear in combination.[21] Bone loss in the medial area can be present as cavernous defect (AAOS type II) [24] or with the destruction of the medial wall (Paprosky type 2C).[23] Beyond those defects, the cranial roof can be affected, either combined with an intact medial wall (Paprosky type 2B) or with a destroyed medial wall (Paprosky type 3A).[23] In order to improve comparability of in vitro tests and transfer to clinical defect situations, it would be beneficial to use a standardized and realistic defect, which could be enlarged incrementally to test different treatment strategies for increasing bone defect severity.

The aims of this study were to (a) develop a simplified bone defect model based on computed tomography (CT) data of a clinically existing acetabular bone defect, (b) derive four different bone defect increments from the simplified defect to build up a step-wise testing procedure for increasing bone defect severity, and (c) assess the influence of defect and BGS filling on primary stability of a press-fit cup in the smallest defined bone defect increment.

2 | MATERIALS AND METHODS

2.1 | Acetabular test model

An artificial acetabular test model was developed, replicating the main support structures os ilium, os pubis, and os ischii, as well as the incisura acetabuli, oriented towards a previously developed surrogate model[25] (Figure 1A). Dimensions of the acetabulum were based on a clinical CT-data set (Figure 2A), whose use was approved by the Ludwig-Maximilians-University Munich ethics committee (project no. 18-108 UE) and incisura diameter was set to 10 mm.[25] The present model was made of 20 pounds per cubic foot (PCF) (0.32 g/cm^3) solid rigid polyurethane (PU) foam (Sawbones, Malmö, Sweden) with a compressive strength of 8.4 MPa and Young's modulus of 210 MPa to represent slightly weakened bone as expected in revision surgeries, referring to a study of Crosnier et al,[26] who used 30 PCF (0.48 g/cm^3) and 15 PCF (0.24 g/cm^3) foams to simulate two different bone qualities. The test model was placed in an additively manufactured fixation block made of acrylic resin. Recess areas around the test model of 8 mm allowed displacement under load (Figure 1B).

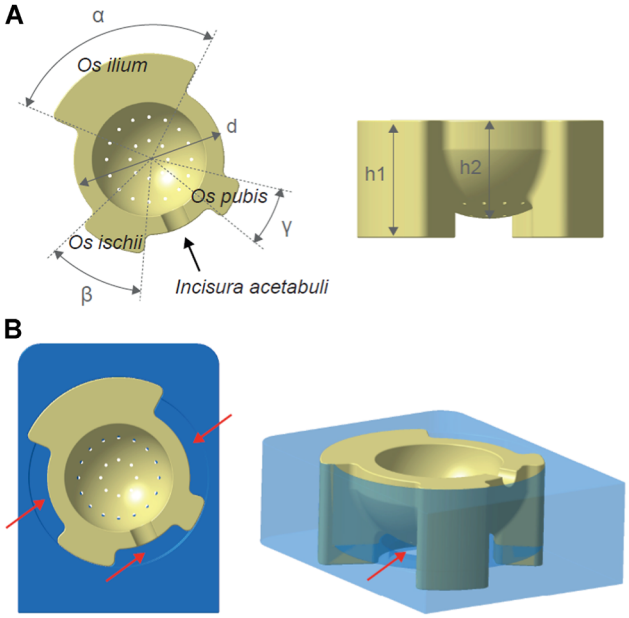

FIGURE 1 Acetabular test model. A, Acetabular model made of 20 PCF polyurethane foam with top and lateral view. Dimensions of os ilium, os pubis, and os ischii ($\alpha = 95°$, $\beta = 26°$, $\gamma = 43°$, $h1 = 40$ mm) and acetabulum ($d = 64$ mm, $h2 = 33.5$ mm) based on the representative clinical computed tomography-data set, and holes in the acetabular ground to allow matrix dissolution in the test series with bone graft substitute. B, Acetabular test model in fixation block made of acrylic resin. Recess areas of 8 mm around the acetabular rim and behind the acetabulum (indicated by red arrows) allow for displacement under load [Color figure can be viewed at wileyonlinelibrary.com]

2.2 | Development of simplified bone defect model and derivation of defect increments

On the basis of the clinically existing defect (Figure 2A), a simplified bone defect model was developed and four defect increments were derived thereof (Figures 3 and S2).

The defect has been quantitatively analyzed based on CT data, alongside with 49 other CT-data sets,[21,27] which was approved by the Ludwig-Maximilians-University Munich ethics committee (project no. 18-108 UE).

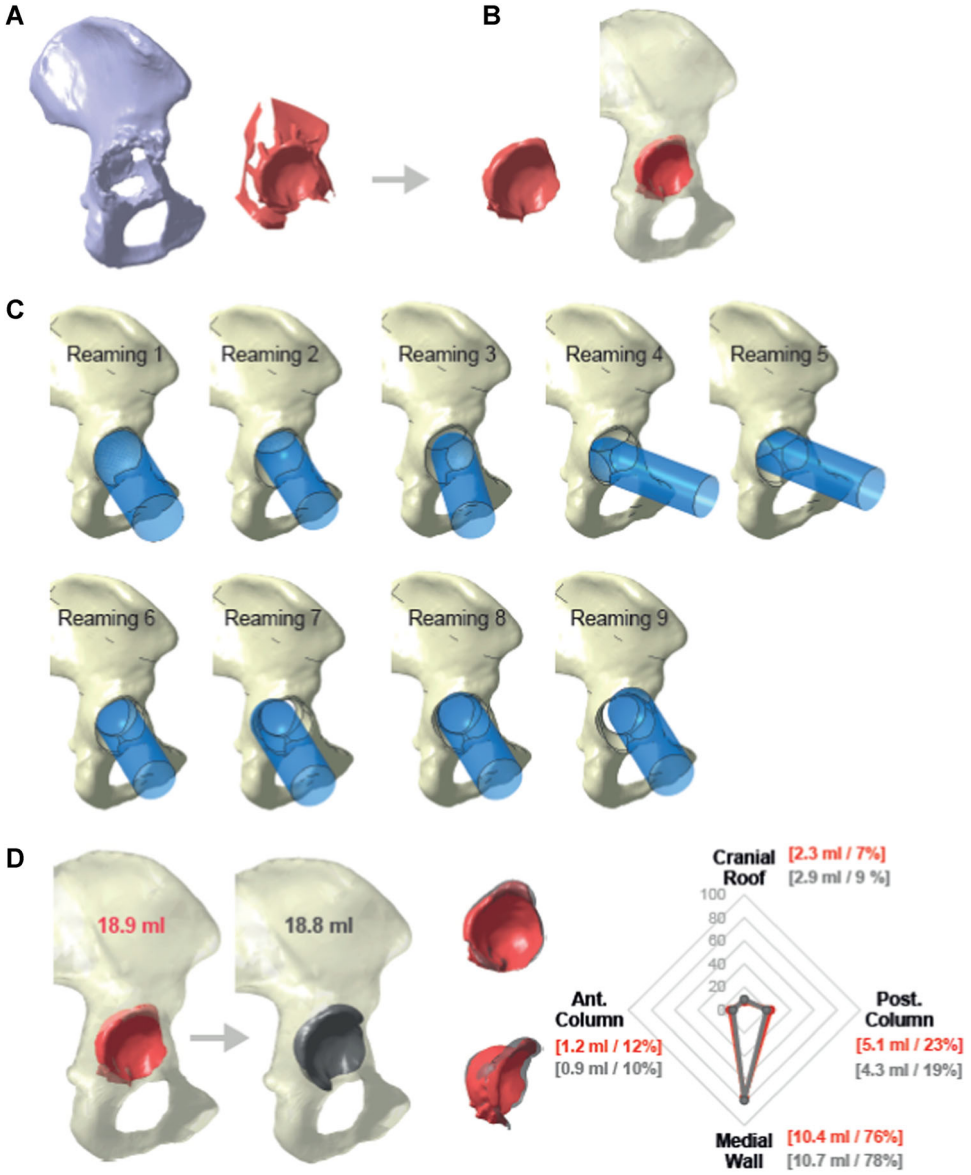

FIGURE 2 Development of simplified bone defect model. A, Three-dimensional model of defect pelvis (light gray) and defect volume (red). B, Defect volume after removal of screw holes and reconstruction inaccuracies. C, Virtual reaming procedures to approximate defect volume with hemispherical reamers. D, Resulting simplified defect volume (dark gray) in comparison with original defect volume (red) presented as spider-plot showing volume loss in the pre-defined defect sectors cranial roof, anterior column, posterior column, and medial wall [Color figure can be viewed at wileyonlinelibrary.com]

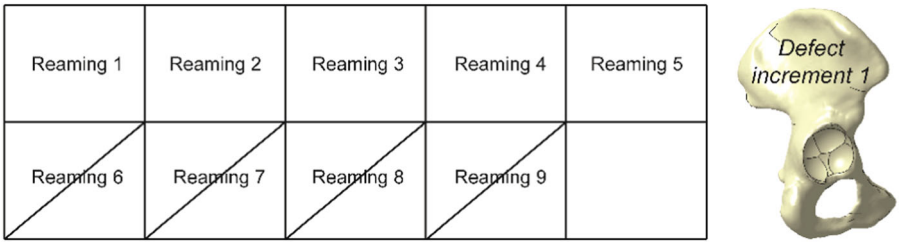

FIGURE 3 Derivation of bone defect increment 1 as a mainly medial contained defect with rim defect in the inferior aspect of the posterior column. Application with reaming procedures 1 to 5 [Color figure can be viewed at wileyonlinelibrary.com]

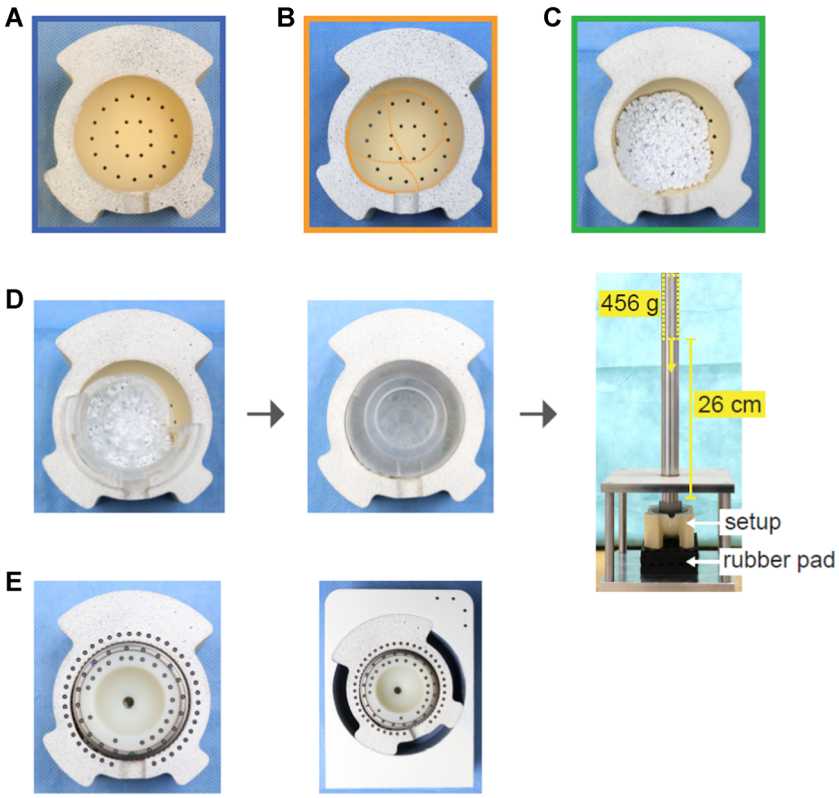

FIGURE 4 The three test series and their preparation. A, Exemplary acetabular test model for test series *Primary*. B, Exemplary model for test series *Empty* with defect contour shown with orange lines. C, Exemplary model for test series *Filled*. D, Preparation of test series *Filled* with implantation of bone graft substitute using a filling template, an impactor, and a drop-weight impaction device. E, Specimen preparation with marker points and positioning in fixation block, exemplary shown for test series *Filled* [Color figure can be viewed at wileyonlinelibrary.com]

Within the data of the quantitative analysis of 50 CT-data sets with acetabular bone defects, it was found that the herein presented defect showed relative bone volume loss close to the 25th percentile in the sectors cranial roof, anterior column and posterior column, and close to the median in the sector medial wall.[21,27] Hence, it represented a comparatively small defect within the 50 analyzed revision cases, which could be treated with bone graft (substitutes), medial and segmental containments, making it a suitable basis for the herein developed acetabular test model. Four senior hip revision surgeons from four different European clinical centers were consulted concerning the choice of a representative defect and confirmed the frequency of this defect type in revision surgery.

After bone volume loss analysis (Figure 2A), shell-like reconstruction inaccuracies were removed, as well as the screw holes, assuming that these were not essential for stability (Figure 2B). In an iterative approach, nine virtual reaming procedures were defined to approximate volume and shape of the original defect with hemispherical acetabular reamers (Figure 2C), which would also enable defect implementation in donor specimens. The resulting simplified defect showed a shape comparable with the original defect (Figure 2D). Total defect volume (bone volume loss) showed a conformity of 99% and distribution of bone volume loss among the four defect sectors cranial roof, anterior column, posterior column, and medial wall[27] was also comparable between original and simplified defect (Figure 2D, spider plot). Detailed information on reamer size, position, and orientation can be found in Figure S1. On the basis of this simplified defect, three additional defect increments were derived by excluding specific reaming procedures (Figure S2) in order to build up a step-wise testing procedure for reconstruction strategies. The first bone defect increment (Figure 3) could be treated with BGS and was used as basis for the acetabular test model in this study. It represents a mainly medial, contained defect with rim damage in the posterior-inferior aspect of approximately one-third of the circumference.

2.3 | Test series and specimen preparation

In the present study, defect *Increment 1* was used in order to test BGS in combination with a press-fit cup. Three test series with $N = 6$ each were conducted (Figure 4A-C): Acetabular test model without defect, treated with press-fit cup (*Primary*), test model with defect, treated with press-fit cup (*Empty*), test model with defect, treated with BGS and press-fit cup (*Filled*). Specimen preparation included application of a random pattern on the acetabular test models to enable later analysis of deformations. A Plasmafit press-fit cup (Aesculap AG, Tuttlingen, Germany) size 48 mm was chosen for all three test series based on virtual planning on the original three-dimensional (3D) defect model in consultation with the clinical advisors. The resulting press-fit with a 48 mm diameter hemispherical reaming was 1.2 mm in relation to the diameter. In the series *Primary* there was circumferential contact between cup and foam, whereas in series *Empty* and *Filled*, rim contact was reduced to 2/3 of the circumference due to the defect in the inferior aspect of the posterior column (Figure 4B). The BGS were β-tricalcium phosphate-hydroxyapatite prototype granules with an edge length of 3.5 mm, which were mixed with polyethylene glycol and glycerin in a ratio of 4:1 by weight. A template was used to fill the defect with the correct and reproducible amount of BGS and a custom-made impactor and a drop-weight-device were used to impact the material. The weight (456 g) was dropped five times from a height of 26 cm (corresponding to an impulse

of 1.0 N·s) to simulate impaction with a standard orthopedic hammer, whereby the acetabular test model was placed on a rubber pad to simulate soft tissue reaction (Figure 4D).

Cups were pressed-in displacement controlled with the material testing machine ZwickRoell Z010 (ZwickRoell GmbH & Co. KG, Ulm, Germany). Specimens were conditioned in purified water for 45 minutes, followed by 24 hours air-cure, and application of marker points on cup and acetabular test model (Figure 4E).

2.4 | Load protocol and stop criteria

The acetabular test model was placed under the servo-hydraulic testing machine MTS 858 Mini Bionix II (MTS, Minneapolis) such that the maximum resultant force during level walking, that is, during contralateral toe-off (Figure 5A) could be applied vertically (Figure 5B). Resultant force direction with respect to the acetabular cup was derived from data provided on www.orthoload.com/orthoload-club/ (OrthoLoad club, Julius-Wolff-Institute Berlin). All basic data have been obtained in a previous study on in vivo loads measured in the femoral prosthesis, which has been independently conducted, funded and ethically approved (EA2/057/09).[28] The data transformed to the acetabular cup coordinate system used within the present study have not yet been published. They are available in an internal database, with access restricted to OrthoLoad club members. Ten patients with instrumented hip implants performed daily activities such as level walking twelve months postoperatively, whereby motion was tracked via an optical tracking system and load in the femoral prosthesis was measured in vivo.[28] In vivo load data and motion data, given in the femoral coordinate system and the lab coordinate system, were transformed to an acetabular cup related coordinate system. Trials were averaged using the "Dynamic time wrapping" approach[29] and scaled.[30]

Mean cup-inclination and anteversion of the 10 patients were 35° ± 6° and 28° ± 7°, respectively. However, with the aim to use a standardized cup position within the Lewinnek safe zone, inclination of 45° and anteversion of 20° were assumed within this study.[31]

Force was applied in a sinusoidal waveform (2 Hz) via a 28 mm ceramic head, with a minimum load of 300 N.[32] Maximum load was

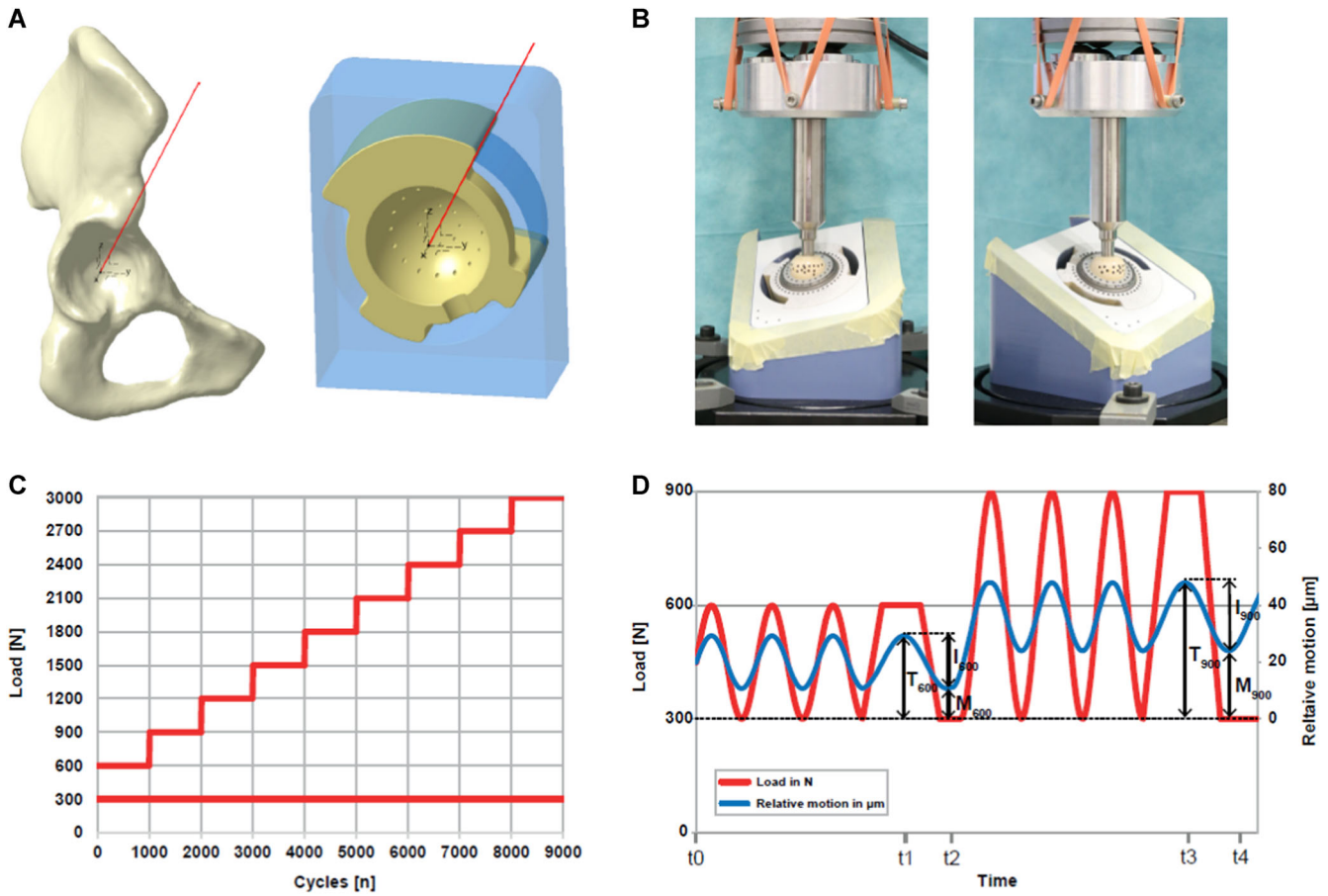

FIGURE 5 Load application and relative motion measurement. A, Direction of load given by the maximum resultant force during level walking, indicated by red line in a native hemipelvis and the acetabular test model. B, Orientation and fixation under the servo-hydraulic testing machine for axial load application. C, Load protocol with nine load stages. Minimum load was 300 N for all load stages and maximum load was increased incrementally with 300 N/1000 cycles from 600 N (load stage 1) to 3000 N (load stage 9). D, Relative motion measurement, schematically shown for the first two load stages (600 and 900 N) with four exemplary load cycles. Load curve (red) overlaid with relative motion curve (blue). Measurements, that is, images were taken at static preload of 300 N (t0) and at the end of each load stage at maximum load (here t1 and t3) and minimum load (here t2 and t4). Inducible displacement was measured between the minimum and maximum load at each load stage (here indicated as I600 and I900). Migration was measured between the preload and the minimum load at each load stage (here indicated as M600 and M900). Total motion was measured between the preload and the maximum load at each load stage (here indicated as T600 and T900) [Color figure can be viewed at wileyonlinelibrary.com]

increased incrementally from 600 N (load stage 1) to 3000 N (load stage 9) with 300 N/1000 cycles[33] (Figure 5C).

Stop criteria was defined as structural failure of the acetabular test model or displacement of the MTS actuator of more than 2.5 mm, which was defined as conservative threshold based on migration values critical for long-term fixation of acetabular cups.[34–36]

2.5 | Relative motion measurement

Relative motions between cup and acetabular test model were assessed in 3D using the optical measurement system GOM Pontos 5 M with two 5MP cameras with 50 mm lenses and the software Aramis Professional 2017 (GOM GmbH, Braunschweig, Germany). Marker points size 0.4 mm (ID 35231) with $N = 20$ on the cup and $N = 38$ on the test model were used, whereof $N = 33$ were visible throughout the tests. Images were taken statically in the beginning at 300 N (t0) and at the end of each load stage at the corresponding maximum (t1, t3, etc.) and minimum (t2, t4, etc.) load (Figure 5D). It was distinguished between inducible displacement, migration, and total motion. Inducible displacement was defined as the displacement between maximum and minimum load in each load stage, migration as the displacement between the preload of 300 N and the minimum load at each load stage, and total motion as the displacement between the preload and the maximum load at each load stage (Figure 5D). Relative motion is given as mean value of all 33 marker points (Figure 6) and as maximum value (highest motion value within all 33 marker points) in Table 1.

2.6 | Resulting cup position measurement

Resulting cup position was measured after dynamic testing as angle between cup surface and acetabular test model surface with the tactile measurement system Prismo Navigator (Carl Zeiss Industrielle Messtechnik GmbH, Oberkochen, Germany) with a measurement inaccuracy of (0.9 + L/350) µm.

2.7 | Statistical analysis

Statistical analysis was performed using nonparametric tests, as the sample size ($N = 6$) was too small to confirm normality. Test for statistical significance between the three test series was performed (Mann-Whitney U test) with level of significance set to $P < .05$.

3 | RESULTS

3.1 | Relative motions

Inducible displacement, migration, and total motion increased with increasing load in all specimens (Figure 6). Series *Empty* was only recorded until load stage 1800 N, as four out of six specimens reached the displacement stop criteria at the following load stage.

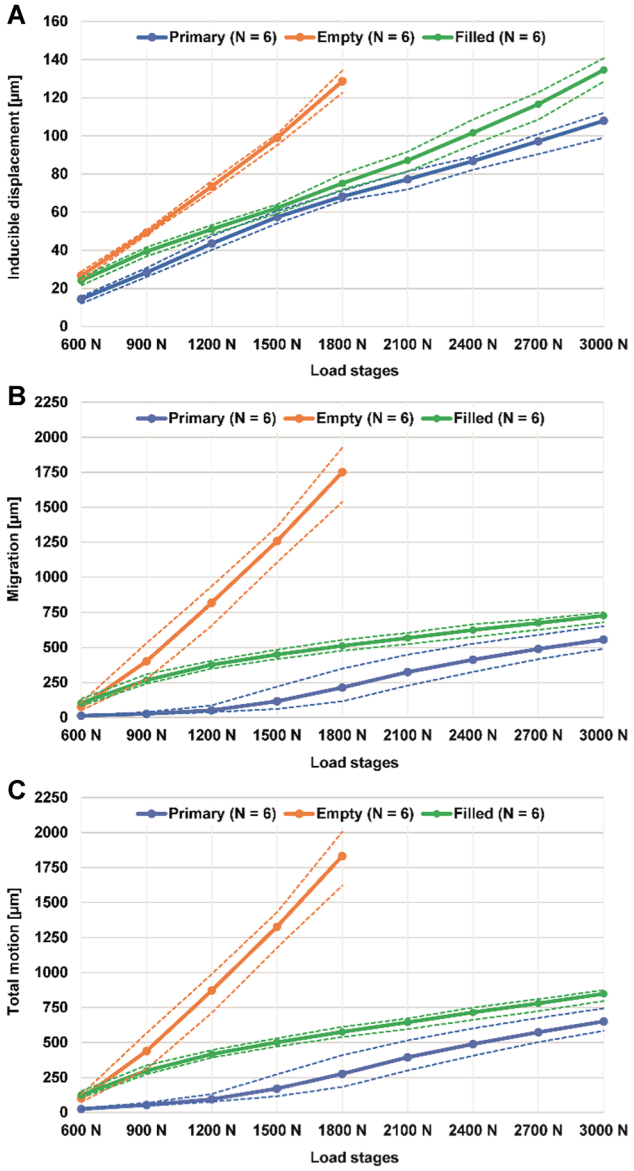

FIGURE 6 Mean relative motions between cup and test model given as mean of $N = 6$ in each test series (solid line) and as minimum and maximum within each test series (dashed lines). A, Inducible displacement. B, Migration. C, Total motion [Color figure can be viewed at wileyonlinelibrary.com]

Mean inducible displacement was highest for *Empty* and lowest for *Primary* in all load stages (Figure 6A). At the last comparable load stage, 1800 N, mean inducible displacement of *Empty* (128.6 ± 4.9 µm) was increased 1.9-fold with respect to *Primary* (68.1 ± 1.9 µm) and in *Filled* (75.1 ± 3.0 µm), it was increased 1.1-fold with respect to *Primary*. Differences between all groups were statistically significant with $P = .002$. Furthermore, *Primary* and *Filled* were comparable in terms of course of curve.

Mean migration was highest for *Empty* and lowest for *Primary* in all load stages, except the first one (Figure 6B). At the last comparable load stage, 1800 N, mean migration of *Empty*

TABLE 1 Inducible displacement, migration, and total motion in each load stage for the three test series, given as maximum motion in the 33 tracking points

	600 N	900 N	1200 N	1500 N	1800 N	2100 N	2400 N	2700 N	3000 N
Maximum inducible displacement in 33 tracking points (mean ± SD) in µm									
Primary	26 ± 3	48 ± 2	71 ± 3	93 ± 5	110 ± 3	127 ± 7	147 ± 5	167 ± 8	186 ± 9
Empty	64 ± 6	128 ± 7	188 ± 6	242 ± 7	292 ± 4	N/A	N/A	N/A	N/A
Filled	50 ± 3	75 ± 9	91 ± 9	106 ± 10	127 ± 7	146 ± 7	171 ± 10	199 ± 11	239 ± 12
Maximum migration in 33 tracking points (mean ± SD) in µm									
Primary	23 ± 4	45 ± 8	79 ± 22	213 ± 161	441 ± 279	730 ± 261	955 ± 239	1138 ± 212	1296 ± 195
Empty	192 ± 54	1077 ± 271	2149 ± 303	3212 ± 279	4311 ± 353	N/A	N/A	N/A	N/A
Filled	271 ± 48	712 ± 66	986 ± 86	1155 ± 120	1279 ± 138	1383 ± 141	1478 ± 146	1563 ± 141	1650 ± 145
Maximum total motion in 33 tracking points (mean ± SD) in µm									
Primary	46 ± 6	90 ± 9	144 ± 18	282 ± 148	500 ± 267	790 ± 257	1022 ± 235	1212 ± 214	1379 ± 199
Empty	252 ± 58	1205 ± 273	2337 ± 301	3451 ± 278	4599 ± 350	N/A	N/A	N/A	N/A
Filled	319 ± 45	787 ± 64	1073 ± 93	1253 ± 128	1390 ± 146	1508 ± 146	1619 ± 154	1727 ± 152	1837 ± 150

Note: Mean values and standard deviations of $N = 6$ in each test series are shown.

(1751.1 ± 154.8 µm) was increased 8.2-fold with respect to *Primary* (213.6 ± 91.7 µm), and in *Filled* (511.4 ± 30.8 µm), it was increased 2.4-fold with respect to *Primary*. Differences between all groups were statistically significant with $P = .002$. *Empty* showed a very steep migration curve, whereas *Filled* showed a steep curve in the beginning with flattening in the following load stages. *Primary* showed almost no migration in the beginning, but started to migrate in load stages 1500 to 1800 N which corresponded to the force applied to press the cups into the *Primary* setups (1841 ± 205 N).

Mean total motion showed characteristics comparable to mean migration with highest values for *Empty* and lowest for *Primary* in load stages 900 to 3000 N (Figure 6C).

3.2 | Resulting cup position

Resulting cup position, measured after dynamic testing, was 1.4° ± 0.3° for *Primary*, 7.3° ± 0.3° for *Empty*, and 1.6° ± 0.3° for *Filled* (Figure 7), which corresponded to a 5.3-fold increase of resulting cup angle in *Empty*, and a 1.1-fold increase in *Filled* in comparison to *Primary*, respectively. Difference was statistically significant between *Primary* and *Empty*, and *Filled* and *Empty* with $P = .002$ each.

4 | DISCUSSION

The objectives of this study were to (a) develop a simplified bone defect model based on a clinically existing acetabular bone defect, (b) build up a procedure to test different treatment strategies in one bone defect model whose severity can be increased incrementally, and (c) compare the stability of a press-fit cup in a primary situation with a revision situation with and without defect filling.

Pre-clinical testing of THA revision treatment strategies has been conducted using donor specimens,[13,15,37,38] Sawbone hemipelvises,[20] and foam models, either as block[32] or with an approximated acetabular shape.[17,25,39] PU foam has been widely used as surrogate for cancellous bone, as it provides more reproducible mechanical properties and can be machined more easily than human cadaver specimens, which both reduces interspecimen variability. It is more readily available, of lower cost and test duration is not critical. Mechanical properties of PU foam are within the range of properties found for cancellous bone,[40,41] although these properties strongly depend on the individual donor, probe extraction site,[42] testing method and specimen preparation used.[43] The trabecular structure of cancellous bone is quite unique and can hardly be replicated by PU foam. The same applies to the anisotropic bone properties prescribed by Wolff's law.[44] Hence, PU foam represents an accepted surrogate material with a lot of advantages over cadaver specimens, but cannot replicate all their unique properties.

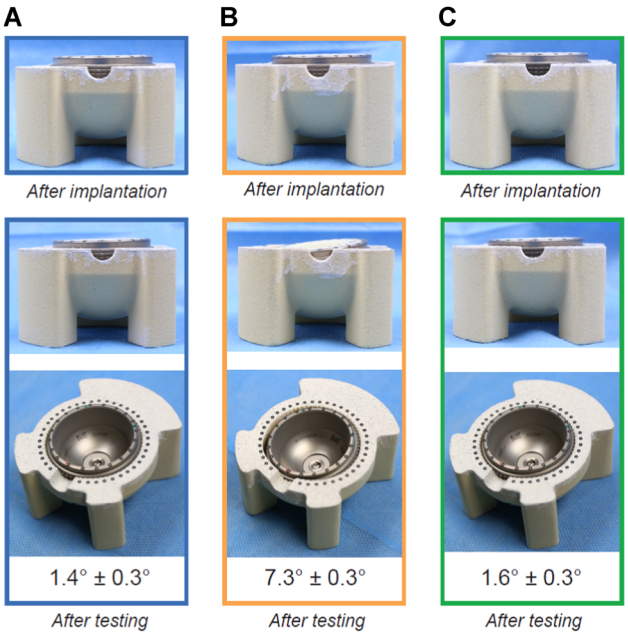

FIGURE 7 Cup positions after implantation and after testing, exemplary shown for one specimen of each test series and with mean ± standard deviation of $N = 6$ each, measured after dynamic testing. A, Primary series. B, Empty series. C, Filled series [Color figure can be viewed at wileyonlinelibrary.com]

In the previous studies, load was applied torsional, tangential (lever-out), dynamically uniaxial, or dynamically with combined axial and rotational load (Table S1). In studies with dynamic load, 20 to 3600 motion cycles were applied, minimum load was 0 to 180 N, maximum load 490 to 3800 N, whereby maximum load was constant throughout the tests in most studies with some exceptions.[17,20,45] In studies with bone defects, these were applied at different locations, either centrally,[15,19,37,45,46] cranially,[11,15] posterior-cranially,[13,25,39] or posteriorly[39] and were implemented in a simplified way, often using only one acetabular reamer.[19,45] Relative motion consists of inducible displacement, migration, and total motion.[47] However, the previously mentioned studies did often not clearly differentiate between these components. Relative motion was measured using LVDTs,[15,32,37] or optical measurement systems.[13,20,38] LVDT measurements are limited to spot checks at the interface[48,49] and the sensor fixation can damage the bone or setup [49,50] and can hence influence it's mechanical properties. Using optical measurement systems, a larger amount of measurement points can be used and due to the fact that the tracking points are only stacked to the surface, the properties of the bone/setup are not influenced. However, this also means that this type of measurement only provides information about relative motions at the surface of the specimen.

In the existing studies, comparison between the treatment strategies is difficult due to the variety of test models (donor specimens, foam models, etc.), load protocols (different directions and magnitudes), types of bone defects (different sizes and locations), interpretations of relative motion (inducible displacement, migration, etc.), and measurement techniques (optical, electromagnetic, etc.) used (Table S1).

Within the present study these points were addressed with the aim to suggest a simplified, but realistic acetabular bone defect, which can be increased incrementally to test different revision treatment strategies based on one defect model. The bone defect was derived from a representative clinical case. It was simplified by nine reaming procedures with standard acetabular reamers such that this method could also be applied in donor specimens. The simplified defect showed a shape comparable to the original defect and a high overall volume conformity of 99% (Figure 2). Four defect increments were derived such that defect severity could be modified by the number of reaming procedures (Figures 3 and S2). Due to the implementation with several reaming procedures, defect shape was irregular and hence close to the clinical case. An acetabular test model, similar to a setup already applied in lever-out studies,[25,39] was used to provide best possible reproducibility and comparability between the test series. The anatomical structure around the acetabulum was mimicked by the main support structures os ilium, os ischii, and os pubis to provide behavior under load as realistic as possible. Dynamic uniaxial loading in direction of level walking was chosen, comparable with other dynamic loading studies.[9,20,51] Load was increased incrementally to investigate the relation of load and relative motion.[9,18–20,45,46] Relative motion between cup and acetabular test model was measured in terms of inducible displacement, migration, and total motion, which has yet only been done by a limited number of studies.[9,52] This enables the distinction between different motion mechanisms and the estimation of their clinical consequence. The smallest defined bone defect increment was implemented in the acetabular test model, representing a mainly contained, medial defect that could be filled by bone graft or BGS. Three test series were conducted comparing the primary situation with a revision situation with and without BGS filling.

Within the test series, it could be seen that mean total relative motion in *Primary* at the first load stage was 24.5 ± 3.8 µm, and the maximum was 45.7 ± 5.6 µm, which was in a range comparable to the relative motions of a press-fit cup measured in human donor specimens with 36.03 ± 16.83 µm.[38] Lowest mean total relative motion was found in *Primary*, as previously documented by Pitto and Schmidt,[15] who assessed primary stability of a Müller, Ganz, and Burch-Schneider cage in a primary and numerous defect situations. In the present study, mean and maximum relative motion increased with increasing load, which is in agreement with cyclic axial loading tests in Sawbone hemipelvises.[20]

This study has several limitations. First, an artificial acetabular model made of PU foam was used to approximate bone structure instead of human donor specimens. Donor specimens represent the most realistic bone model, but are associated with restricted reproducibility and test duration. Second, the presented acetabular bone defect model was based on one individual defect case. The defect was chosen based on a quantitative defect analysis in consultation with four senior hip revision surgeons as a common and rather small defect with main damage in the medial and posterior aspect. There is a wide variation of acetabular bone defects and the herein presented defect model cannot cover the whole range encountered in revision patients, but is rather a first building block for a standardized testing procedure. Depending on the amount and distribution of bone loss in the different sectors of the acetabulum, results for relative motion may be different for other defects, for example, when the cranial roof is concerned with a larger amount of bone loss. Hence, conclusions derived from the specific bone defect model presented in this study do not necessarily apply to all individuals with bone loss. Third, uniaxial loading was applied although most daily activities represent multiaxial loading scenarios and relative motion was found to be higher under multiaxial, than under uniaxial loading.[51] However, application of uniaxial load does not jeopardize comparability between the different test series and multiaxial loading is sophisticated to simulate in vitro, especially when motion tracking is required during the tests. In addition, uniaxial loading does not simulate friction moments. However, during contralateral toe-off, moments are limited to 0.17% BWm[28] and were therefore considered to have only little impact on relative motion. Fourth, relative motion measurement was performed under static load at the end of each load stage such that the temporal course of relative motion within the single load stages could not be investigated. Fifth, angles between cup and acetabular model were measured only after the tests and values on cup orientation before testing could not be provided.

To the authors' knowledge, this is the first study to suggest a bone defect model based on a representative CT-data set of an acetabular bone defect. The defect was simplified with nine reamings

with standard acetabular reamers, such that it could be also implemented in human donor specimens. From the simplified defect, four defect increments were derived such that different treatment strategies could be tested based on one model. Relative motion in terms of inducible displacement, migration, and total motion was measured in the three test series *Primary*, *Empty*, and *Filled* such that the influence of bone defect and defect filling on stability of a press-fit cup could be assessed. The presented method provides a platform to test stability of different treatment strategies based on one simplified, but realistic bone defect model in a standardized way.

Future studies should include validation of the acetabular test model by comparison with human donor specimens, the comparison of BGS with bone grafts, testing of other treatment strategies in higher bone defect increments, and the application of additional directions of load. Furthermore, the acetabular bone defect model should be extended with additional bone defects to cover a broader range of bone defects encountered in revision patients.

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CONFLICT OF INTERESTS

RS, GH, MB, and TG are employees of Aesculap AG Tuttlingen, a manufacturer of orthopedic implants. FM and PD receive institutional support by Aesculap AG. Aesculap AG provided support in the form of salaries for RS, GH, MB, and TG, but did not have any role in design of the study, data analysis, decision to publish or manuscript preparation.

AUTHOR CONTRIBUTIONS

All authors worked on research design, or acquisition, analysis or interpretation of data. RS worked on drafting the paper and GH, AR, MB, FM, PD, and TG critically revised it. All authors have read and approved the final submitted manuscript.

ORCID

Ronja A. Schierjott http://orcid.org/0000-0002-0624-1978

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SUPPORTING INFORMATION

Additional supporting information may be found online in the Supporting Information section.

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