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Does surface roughness influence the primary stability of acetabular cups? A numerical and experimental biomechanical evaluation

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Most acetabular cups implanted today are press-fit impacted cementless. Anchorage begins with the primary stability given by insertion of a slightly oversized cup. This primary stability is key to obtaining bone ingrowth and secondary stability. We tested the hypothesis that primary stability of the cup is related to surface roughness of the implant, using both an experimental and a numerical models to analyze how three levels of surface roughness (micro, macro and combined) affect the primary stability of the cup. We also investigated the effect of differences in diameter between the cup and its substrate, and of insertion force, on the cups' primary stability. The results of our study show that primary stability depends on the surface roughness of the cup. The presence of macro-roughness on the peripheral ring is found to decrease primary stability; there was excessive abrasion of the substrate, damaging it and leading to poor primary stability. Numerical modeling indicates that oversizing the cup compared to its substrate has an impact on primary stability, as has insertion force.

1. Introduction

Total hip arthroplasty (THA) was defined as the "operation of the century" in a paper published in 2007 by Learmonth et al. [1], due to the excellent and long-lasting clinical and radiological results. Long-term stability of acetabular implants depends on their primary stability following implantation [2–6]. This is because long-term stable osseointegration of the porous-coated acetabular cups depends on bone ingrowth within their porous surface [5]. Primary stability is therefore vital to avoid micromotion, which, if excessive, can limit bone ingrowth [7]. This rigid initial fixation of uncemented cups is usually obtained via appropriate impaction and/or screws. Screws could provide strong fixation but involve a potential risk of vascular complications, and their use has declined in the last decade [8]. Impaction currently seems to be the best option for firm fixation. To achieve natural retention of the cup within the acetabulum, the implant must be hemispherical with a flattened dome [2,5,9]. This design maximizes the contact area between the implant and bone and achieves optimal distribution of stresses. However, in vivo assessment of press-fit stability is

relatively imprecise and only based on the surgeon's experience, without any real quantitative evaluation of stability [10]. Thus, to evaluate primary stability in acetabular press-fit cups, mechanical testing and numerical studies were developed and carried out on different substrates such as artificial bone [2,4–6], animal bone [2,11] or human bone [3,12–15]. The tests can be divided into cyclic axial loading [12,13], tangential and rotational stability [2,5,6], and pull-out tests [6]. A complementary approach to analyze primary stability is the finite element (FE) method, which can accommodate large variations in geometry (reaming of the acetabulum) and material properties. FE analysis permits the study of the bone/cup interface (such as contact or strain data) [16–19]. The few studies that backed their numerical model with mechanical testing used cyclic axial loading tests [16,19], not the most accurate representation of a surgeon's hand movements during an operation, to test the stability of the cup immediately after impaction. Yet accurate and reproducible evaluation of the primary stability of press-fit acetabular cups, while complex and dependent on a variety of parameters, is crucial to optimize design.

New press-fit acetabular cups using porous materials of varying degrees of roughness are being developed, and we tested the hypothesis that their primary stability would be impacted by this surface roughness. To evaluate how the primary lateral stability of three press-fit acetabular cups of identical design was affected

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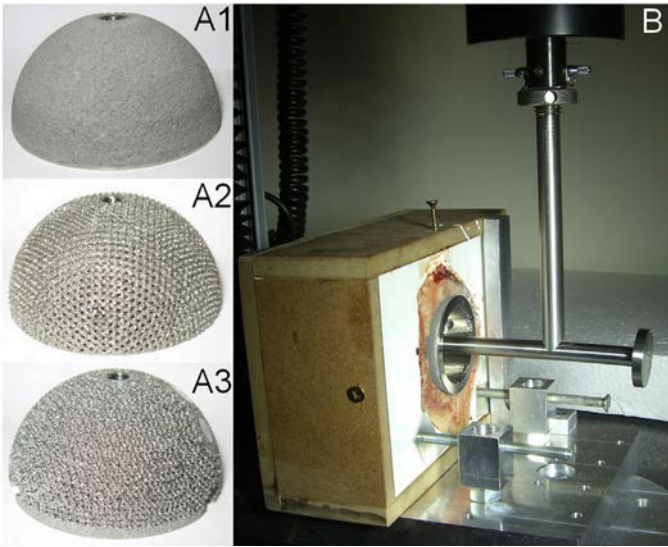


Fig. 1. The cups tested: A1: with smooth surface; A2: with macro-spikes of titanium; A3: with macro-spikes and a smooth equatorial surface. B: tangential stability (TS) test.

by their differing surface roughness, we used first an experimental set-up and second finite-element analysis. We further investigated the influence of the sizing of the cup compared to its substrate and of the insertion force used.

2. Materials and methods

Our experimental study used reproducible materials, starting with an artificial homogeneous material (sawbone) followed by bovine bone, to assess both the effect of the cups' macro-roughness and the effect of the insertion force on their primary stability. A related numerical analysis on sawbone was then performed to obtain contact information not accessible via experiments and which might elucidate the contact and interaction mechanisms between an inserted press-fit cup and its substrate, as well as how differences in cup diameter might affect primary stability.

2.1. Experimental part

Three geometrically identical press-fit cups, of differing degrees of surface roughness (Fig. 1A), were compared for primary stability. We tested a standard press-fit cup (A1), considered as the micro-roughness cup, with a conglomeration of titanium balls (diameter 300 μm) covered with a hydroxyapatite coating (70 μm thickness), and two new-generation cups: one with macro-roughness (A2) and the other with combined macro- and micro-roughness (A3). The first (A2) has 1 mm high titanium macro-spikes spaced 0.7 mm, and the second (A3) is a combination of A1 and A2: smooth (similar A1) around an equatorial ring (5 mm wide) and abrasive (spikes 1 mm high) everywhere else. Moreover, the cups with macro-roughness (A2 and A3) offer surface porosity within their surface, to enhance the osseointegration of the prosthesis during secondary stability. Six cups in each of the three materials were obtained from Adler Ortho® (Milano, Italy). External diameters were 54 mm for A1 and A3 and 55 mm for A2.

Two different substrate materials were used. Sawbones® (Malmö, Sweden) provided polyurethane foam blocks of $130 \times 130 \times 40 \text{ mm}^3$, with a density of 0.32 g/cm³, simulating cancellous bone in a reproducible, clean and artificial material. Then, specimens of proximal bovine humerus, more similar to human bone with the presence of fluids and already used for

analysis of cup impaction [20], were obtained and cut with a handsaw into blocks of approximately $80 \times 140 \times 50 \text{ mm}^3$. On the advice of the implant designers, all blocks were under-reamed 1 mm compared to the external size of the cups, with surgical reamers (holes of 53 mm diameter for A1 and A3, and 54 mm diameter for A2). Blocks of bovine bone were frozen at -20°C immediately after reaming. The day before the experiments, the bovine bones were defrosted at room temperature and placed in wooden boxes of size $130 \times 180 \times 70 \text{ mm}^3$ filled with Fascast polyurethane resin (Axson Technologies®). The bone surrounding the reamed holes was far enough from the resin not to be affected by the thermal effect of hardening. Cups' insertion was done using the testing machine detailed below, that does not reproduce the dynamic characteristics of the clinical environment. But we opted for a controlled insertion force to obtain test repeatability. As a first experimental campaign, A1 and A2 cups were implanted with a force of 1800 N, as per the literature [2,19]. Next, to investigate the influence of insertion force on primary stability, the A2 cups were subjected to two different insertion forces, first 1800 and then 4000 N, to enhance cup implantation (just beneath the surface). A3 was then compared to A2 via a second experimental campaign where an insertion force of 4000 N was used to give a more realistic representation of the appearance of the cup after insertion.

Two tests for tangential stability (TS) and pull-out (PO) were designed and performed using the materials testing system Instron (INSTRON 5566A). An experimental setup specifically manufactured for this study was designed so as to hold every substrate on the test platform during both tests. All cups underwent preloading of 4 N. In the TS test, cups were pulled down at an angle of 90° using a metallic rod (the impactor) threaded into the cup (Fig. 1B), in order to produce a tangential load on the cup (the load is applied 60 mm from the cup edge). The objective here was to obtain the closest match with the surgeon's hand movement during a THA. The PO test involved pulling the cup out vertically, thus with the same orientation as for insertion but in the opposite direction. The crosshead was moved at a rate of 1 mm/min for each test, and the maximum load required to extract each cup from its substrate was recorded (load precision $\pm 0.5\%$). Each test was repeated three times for each type of cup (A1, A2 and A3) on each support (sawbone and bovine bone), for both PO and TS. In addition, two insertion forces were tested on A2, for all configurations. A total of 48 tests were thus performed.

2.2. Numerical part

Using the finite-element code ABAQUS (ABAQUS V6.11; Simulia Corp., Providence, RI, USA), we created a 3D model of a block of sawbone measuring $130 \times 130 \times 40 \text{ mm}^3$ with a central hemisphere hole of 54 mm diameter, identical to that used for A2 in the experiments. The goal was to obtain information on the contact between the cup and its substrate. The block was discretized into 20,375 tetrahedral linear elements. The cup was modeled as a rigid hemisphere with an external diameter of 54.5, 55 or 55.5 mm, which represents an oversizing of 0.5, 1 and 1.5 mm, respectively. The block of sawbone was assumed to have linear isotropic elastoplasticity: elastic properties obtained from the manufacturer Sawbone® with Young modulus $E = 284 \text{ MPa}$ and Poisson coefficient $\nu = 0.3$, and plastic properties obtained from Calvert et al. [21]. Displacements were prescribed to zero on the base of the block and on its lateral faces. The cup was placed on the block in contact with the periphery of the hole. A reference point (RP), situated 60 mm from the cup edge, was used to apply the load, as in the experiments. The experimental protocol was numerically reproduced with quasi-static resolution:

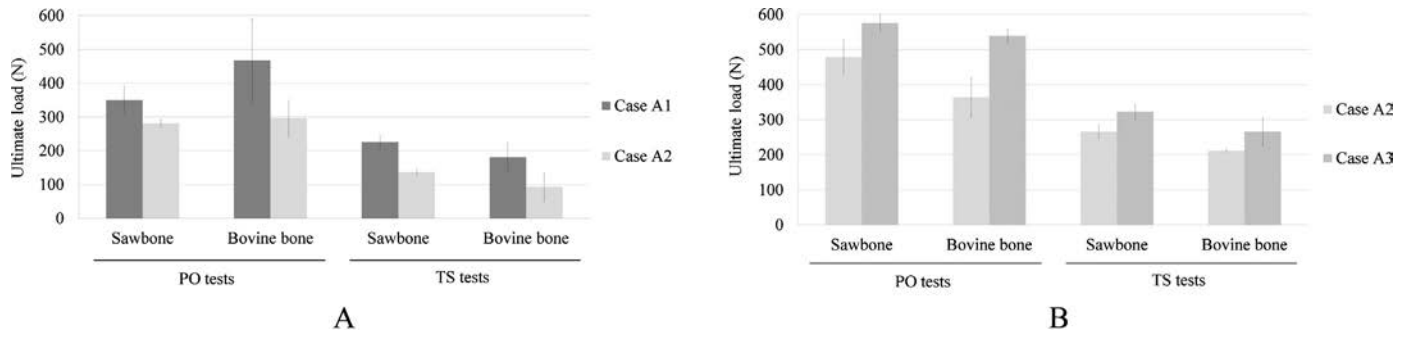


Fig. 2. Histogram summarizing ultimate loads sustained by the cups during pull-out and tangential stability tests. A: comparison of the first pair of cups (A1 and A2) during the first experimental campaign, insertion at 1800 N; and B: comparison of the second pair of cups (A2 and A3) during the second experimental campaign, insertion at 4000 N.

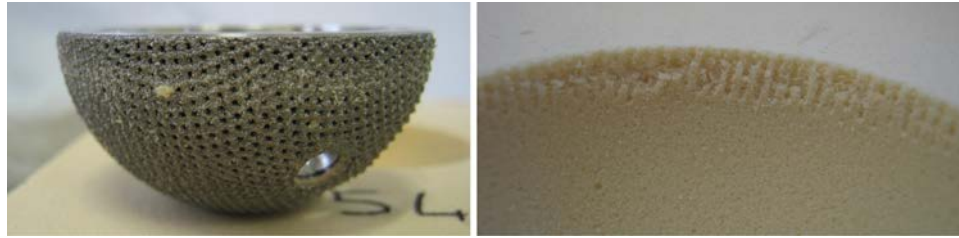


Fig. 3. Contact appearance of a cup with macro-spikes of titanium (A2) and a sawbone after testing.

- *First step—insertion:* at the cup/substrate interface, Coulomb friction law was taken into account, with a friction coefficient of 0.25. A downward displacement of 5.5 mm was applied on the reference point (RP) of the cup to realize cup insertion. Under this model geometry, such a displacement corresponds to an approximate force of 3000 N.
- *Second step—tests:* TS was realized by applying a radial displacement of 3 mm on the RP of the cup. For the PO test, a vertical displacement of 2 mm was applied. These two tests enabled the stability forces to be determined.

Friction coefficient and numerical insertion force were chosen consistent with the experimental results, based on the loads withstood by the cups during the experimental tangential stability test.

Contact zones and elastic and plastic strain distributions on the support were recorded after cup insertion, as well as the evolution of the stability forces *versus* displacement during the numerical TS test.

3. Results

Table 1 summarizes the stability data from the two different tests, for the three types of cups implanted in the two different substrates. The effect of surface roughness is illustrated in Fig. 2. In the first experimental campaign, with insertion at 1800 N, A1 (the standard cup) withstood greater ultimate loads than A2 (with macro-spikes of titanium). In the second experimental campaign, with insertion at 4000 N, A3 (with a smooth equatorial surface) withstood greater ultimate loads than A2 (with macro-spikes of titanium). This was confirmed, thanks to the use of an artificial support (polyurethane foam), during naked-eye analysis of the hole after testing, which revealed that the contact between the cup and its substrate is only equatorial and that the macro-roughness cup (A2) is very abrasive (Fig. 3). The insertion force can be compared through rows 2 and 3; A2 withstood greater ultimate loads with an increased insertion force, as also shown in Fig. 4. The FE model analysis of the contact area between the cup and its substrate is

reported in Fig. 5. The upper part of Fig. 5A shows local contact areas (in green) between the cup and the sawbone after insertion, for the three different diameters tested. Local plasticity of the sawbone after insertion is illustrated in the lower part of Fig. 5A. The greater the oversizing (left), the smaller the contact zone and the greater the plasticity values are. Fig. 5B shows the evolution of the stability forces withstood by the cups of different diameters *versus* displacement during the numerical TS test. Under identical conditions, i.e. substrate hole reamed at 54 mm diameter and fixed insertion force and friction, the 55 mm diameter cup withstood the greatest stability loads.

4. Discussion

When a patient who has undergone THA begins his rehabilitation, the osseointegration of the acetabular cup is not yet achieved, and the implant can be prone to rotational movements due to tangential forces applied by the femoral component [4]. During this phase, lasting approximately 3 months [22], the stability of the cup is called primary stability. The cup is held only mechanically by the elastic recoil of the substrate. Secondary stability consists in the osseointegration of the prosthesis, with bone healing. However, to

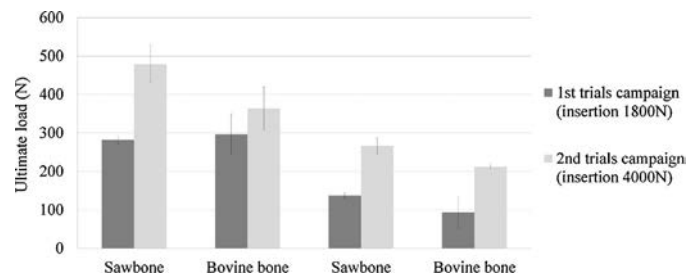


Fig. 4. Histogram summarizing the results of the ultimate loads of A2 for all tested conditions: pull-out and tangential stability tests, on sawbone and bovine bone, and for both insertion forces, 1800 and 4000 N.

Table 1

Tangential (TS) and pull-out (PO) forces for the whole cups, the two different substrates and the two tests.

		Sawbone		Bovine bone	
		TS (N)	PO (N)	TS (N)	PO (N)
First experimental campaign (insertion 1800 N)	A1	226.4 ± 19	350.5 ± 40.1	181.1 ± 41.5	467.4 ± 125.3
	A2	137 ± 7.4	281.7 ± 11.2	93.3 ± 40.4	296.2 ± 53.6
Second experimental campaign (insertion 4000 N)	A2	266 ± 19.8	479 ± 48.5	211 ± 4.3	364 ± 55.4
	A3	323.4 ± 21.9	576.6 ± 23.5	265.8 ± 39.9	539.5 ± 19.5

obtain good secondary stability, primary stability must be optimal. The new-generation hemispherical cups tested here (A2 and A3) offer a surface macro-roughness (titanium spikes) that appears promising to penetrate bone and thus potentially enhance both primary and secondary stability (promoting osseointegration). While secondary stability can be clinically evaluated in vivo with radiographic follow-up (bone healing), primary stability could only be objectively evaluated by mechanical testing. Our reproducible experimental study aimed to evaluate the effect of surface roughness on primary stability. Our findings make it clear that the equatorial surface of the acetabular cup should not be overly abrasive: contact is only peripheral, and macro-spikes of titanium at the equator destroy the holding substrate, resulting in poor primary stability. Primary stability is also found to increase with increasing insertion force, but this is limited by the risk of acetabular fracture. Finally, the sizing of the cups compared to the substrate hole is found to have an impact on primary stability; a compromise has to be made between the elastic recoil of the substrate and the size of the contact zone.

The effect of the roughness of the cups on their primary stability was assessed via mechanical testing. Visual examination of the contact area after the tests showed that the macro-roughness cup (A2) had left significant abrasions on its stability ring (Fig. 3). As can be seen from Table 1, this type of cup withstood lower ultimate loads than the micro-roughness cups (A1). Contrary to our initial hypothesis that macro-roughness would enhance primary stability, its presence on the equatorial surface of the cup actually leads to the destruction of the substrate during insertion, which prevents the substrate from exerting elastic recoil essential for stability. This lack of compressive loads on the cup explains the poorer A2 results obtained in the TS and PO tests compared to the standard A1 cup (Fig. 2A). In order to avoid this equatorial destruction, a new cup was developed (A3), combining A1 and A2:

micro-roughness on an equatorial ring (5 mm wide) and macro-roughness (1 mm high titanium macro-spikes) everywhere else. This new cup was mechanically compared to A2 during the second experimental campaign (Table 1, summarized in histogram in Fig. 2B), to determine whether adding a smooth surface increases primary stability. Results are encouraging; overall, A3 withstood greater than A2 (Fig. 2), confirming that macro-roughness on the equatorial surface decreases primary stability, destroying the holding substrate.

We also evaluated the impact of the oversizing of a cup on its primary stability, using numerical modeling. Modeling gave us qualitative information about the contact distribution and local stresses after insertion, enabling comparisons among the cups. The first conclusion of the model was that the contact between the cup and its support after insertion of an oversized cup is only established on a narrow surface at the equator (Fig. 5A). This was confirmed by examining the damaged area in the mechanical study (Fig. 3). It was also observed by Olory et al. [5], who added that the larger the contact area, the better the primary stability is. Thus, it was decided to increase contact after insertion, by reducing the difference in diameter (Fig. 5A, top). Modeling then showed a trend: for a given insertion force, the more oversized the cup, the smaller the contact area is; full contact is achieved when the cup and hole sizes are identical. We noted that with a 0.5 mm difference in diameter, lower than the 1 mm advised by the implant designers, the contact between the cup and its support was better established. However, in that configuration, modeling showed that the resulting compression force applied to the cup after insertion is lower than in a configuration with a difference of 1 mm, indicating lower stability during the bending test (Fig. 5B). In contrast, implanting with a 1.5 mm difference in diameter leads to a decrease in the contact area, as well as in the resulting compressive forces, and

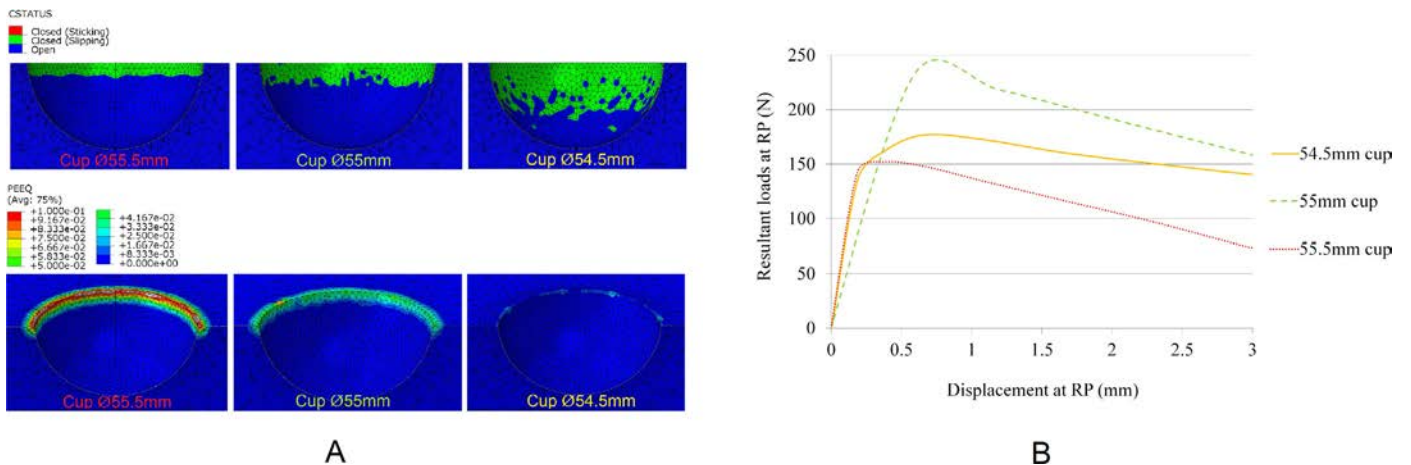


Fig. 5. A: numerical contact areas in green (top) and plasticity (bottom) of the sawbone after insertion of cups of diameters 54.5, 55 and 55.5 mm. B: numerical results of the evolution of the resulting forces vs displacements during the bending test, according to the diameter of the cup, for a reaming of 54 mm and fixed insertion force and friction.

thus to lower primary stability. Consequently, for a given insertion force, it seems that a 1 mm difference in diameter ensures the best cup stability (Fig. 5). Moreover, in terms of the plasticity of the support (Fig. 5A, bottom), the greater the difference in diameter (left), the greater the plastic deformations are, and thus the greater the risk of fracture during insertion. Adler et al. [2] showed that under-reaming of 1 and 2 mm increased the cups' primary stability, respectively by 20 and 60%, which partly contradicts the results of our numerical calculation. However, the authors manually impacted the cups until contact, and thus insertion forces were not identical. We experimentally observed the effect of insertion force on A2 cups, using two different forces: 1800 and 4000 N (Fig. 4). We observed that increasing the insertion force results in increased ultimate restraining forces. This confirms that greater insertion force leads to better contact between the cup and the substrate, and thus to better primary stability. Numerical modeling confirmed these results and also showed that the greater the insertion force, the greater the plasticity of the support is, approaching the ultimate stress of the material. Naturally, the insertion force that can be used is limited by the risk of fracture of the support, and this insertion force needs to be carefully chosen to avoid the collapse of the acetabulum [23].

To the authors' knowledge, there are no explicit minimum values for the restraining forces required to obtain sufficient primary stability. There is thus no way to be sure that before osseointegration, the cup will not be subjected to micromotion, which will produce fibrous tissues. In our study, three press-fit cups differing in surface roughness were compared for primary stability. Numerical modeling revealed that after insertion of an oversized cup, the contact surface between the cup and the bone is only peripheral. Mechanical trials showed the cup with micro-roughness to have better primary stability. The macro-roughness of cups A2 and A3 destroyed the equatorial stability ring on the substrate; reducing roughness on the equatorial ring of the cup enhances its primary stability. Concerning insertion force, our findings suggest that a compromise needs to be made. While the force that can be applied is limited by the risk of fracture of the support (numerical modeling), greater force can also increase stability (mechanical testing) because contact between the cup and its support is maximized. Finally, the numerical model indicates that oversizing the cup has an impact on primary stability, and here too a compromise needs to be made. For a given insertion force, excessive oversizing increases the plasticity damage to the substrate hole, leading to an impaired contact area and thus to poorer primary stability.

There are differences between our experimental conditions and the clinical environment. However, the aim of our study was to compare and evaluate different configurations rather than to obtain physiological values for primary stability. One limitation of our study is the supports: we tried to reproduce clinical practice by reaming them with surgical reamers, but the cups were inserted into created and not existing holes (like the acetabulum). Thus, we were unable to reproduce the cortical part of the acetabulum, which may affect contact between the cup and the support and thus the values obtained for loads withstood by the cups. However, as written above, our goal here was to compare different configurations rather than to determine physiological values. We worked on sawbone and bovine bone samples in order to avoid the variations in mechanical properties encountered with human bone samples. The mechanical behavior of the samples used can therefore be assumed reproducible. Concerning insertion, it was performed via the machine so as to ensure reproducibility, but relatively slowly (1 mm/min), which did not allow the viscoelastic stiffening that can occur with high-speed impact. This too may be a factor in the values obtained for loads withstood by the cups; however, as we reproduced the same insertion protocol for all tests, results are comparable among the configurations. The 48 tests

realized include only three repeats of each condition, which limits the conclusions that can reliably be drawn. Moreover, due to the limited number of implants, only diameters of 55 and 54 mm were tested; it would be interesting to test both smaller and larger cups, to determine whether the impact of under-reaming is similar for other sizes.

To conclude, the principal recommendation here is to remove macro-roughness on the equatorial contact ring, as tested here with A3, to avoid destruction of the substrate, which leads to reduced elastic recoil and decreased stability. Nevertheless, some macro-roughness should be maintained for secondary stability, to enhance osseointegration, thereby improving anchorage of the prosthesis.

Conflicts of interest statement

None declared.

Ethical approval

Not required.

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