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Anisotropic bone remodeling of a biomimetic metal-on-metal hip resurfacing implant

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ABSTRACT

Hip resurfacing (HR) is a highly attractive option for young and active patients. Some surgeons have advocated cementing the metaphyseal stem of the femoral component to improve fixation and survivorship of HR. However, extending component fixation to the metaphysis may promote femoral head strain shielding, which in turn may reduce survival of the femoral component. Replacing the metallic metaphyseal stem by a composite material with bone-matching properties could help to alleviate this phenomenon. This study uses finite element analysis to examine the strain state in the femoral head for three types of implant fixation: an unfixed metallic stem, an osseointegrated biomimetic stem and a cemented metallic stem. Bone remodeling is also simulated to evaluate long-term bone resorption due to strain shielding. Results show that the unfixed stem causes strain shielding in the femoral head, and that cementing the stem increases strain shielding. The biomimetic stem does not eliminate the strain shielding effect, but reduces it significantly versus the metallic cemented version. The current finite element study suggests that an osseointegrated metaphyseal stem made of biomimetic material in hip resurfacing implants could become an interesting alternative when fixation extension is desired.

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1. Introduction

After a rather disappointing beginning in the 1970s, modern hip resurfacing (HR) implants with metal-on-metal (MoM) bearing have regained interest as a viable alternative to total hip arthroplasty (THA) in young and active patients. Modern implants are made of cobalt–chrome alloys (Co–Cr) with controlled diametral clearance that favours hydrodynamic lubrication of the joint. In most cases, the acetabular cup is press-fitted and the femoral component is cemented (no cement on the alignment stem) and the survivorship of HR is more encouraging [1].

Although attractive because it preserves bone mass of the proximal femur, HR cannot be offered to all patients. Those suffering from femoral head deficiencies such as large cysts, necrotic bone or extensive avascular necrosis may not be considered for HR because of lack of proper fixation of the femoral component. Possible failure modes are femoral neck fracture [2] and aseptic loosening. Metal allergy and hypersensitivity to metallic wear debris are also a concern [3]. Risk factor for failure of HR includes small femoral

components (<48 mm) and presence of large bone cysts (>1 cm) [4]. A study recently showed that out of the target patient population of young and active patients, only 46% would be candidate for the procedure [5].

In two recent studies [6,7], Amstutz et al. recommended cementing the stem in cases where femoral head deficiencies (presence of bone cysts, necrotic bone, osteopenia, etc.) are present to improve implant stability. This approach has been shown to be effective, with a survivorship of 98.2% at 5 years [6] on hips with small femoral components (<48 mm) or with large femoral defects (>1 cm). Extending the fixation in the metaphysis would therefore seem to stabilise the implant and compensate for the lack of good bone quality in the femoral head. However, finite element (FE) studies have shown that fixing the full length of metallic stems of commercially available resurfacing implants, either by using cement or as a result of osseointegration, led to decreases in stresses and strains in the femoral head trabecular bone [8–10]. In this case, FE revealed stress shielding specifically in the region underneath the spherical part of the implant, which is not transparent to X-rays and cannot be evaluated clinically. Such an effect could eventually result in significant bone resorption, which would cause the failure of the implant. To this date however, with only short term follow up, clinical studies have not shown adverse effects to fixation of the stem [6] aside from neck narrowing [11]. Longer follow-up is

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Fig. 1. Durom resurfacing implant (a) and geometrical model of unfixed stem fixation (b).

needed to determine whether this cementing technique is detrimental to long-term survivorship of the implant.

More than two decades ago, it was proposed that the stress shielding problem created by fixation of the metallic stem of commercial implants could be alleviated if a material with bone-matching properties were used [12]. A biomimetic material of carbon fibre in a polyamide matrix was previously used to create a hip stem for THA [13,14]; FE studies showed that this stem reduced stress shielding in the proximal femur when compared with a more traditional metallic stem [15,16]. We hypothesize, in cases of HR where extension of fixation to the metaphysis is desired, an osseointegrated implant with a metaphyseal stem made from such a bone-matching material would provide extended fixation, therefore improving implant stability without increasing strain shielding in the femoral head.

The first objective of this study is therefore to determine whether extending the implant fixation to the metaphysis actually leads to increased strain shielding in the femoral head in hip resurfacing, and in the case it does, whether that increase is significant enough to promote detrimental bone remodeling. The second objective is to determine the potential of a bone-matching or biomimetic material as a replacement to metallic stems to solve the strain shielding problem expected in cases of hip resurfacing implants with extended fixation. FE method is used for the analysis of the stress and strain distributions in the bone with different types of hip resurfacing implants. An anisotropic bone remodeling model is used to evaluate long-term bone resorption or apposition.

2. Materials and methods

The new biomimetic material proposed is a composite made of continuous carbon fibres (diameter $\approx 12 \mu\text{m}$) and a polyamide 12 resin matrix (CF/PA12). The material has been tested for biocompatibility *in vitro* and *in vivo* in a rabbit study [17] and good bone apposition with minimal inflammation (haematoxylin eosin (H&E) staining) with no osteolysis when exposed to purposely produced debris from grit blasting was seen in a rat model [18]. It was also optimised in previous studies [14] to closely match cortical bone compressive properties, as can be seen in Table 1. Its density also closely matches that of cortical bone.

For comparison purposes with literature data, the implant design is based on a commercially available model, Zimmer's Durom implant (see Fig. 1). Three different implants fixation types are studied. Standard cementing techniques [7] for resurfacing implants dictate that the short metaphyseal stem be non-cemented and only used as an alignment aid during implantation. The first type is an unfixed stem implant which will serve as a control for the bone remodeling results (unfixed stem model). The second type is a cemented stem Durom implant, with the cement mantle covering

Personalized bone density and elasticity modulus distributions

Fig. 2. Apparent bone density and elastic modulus distributions obtained from CT data (cortical and trabecular bone of the proximal femur are shown).

the full length of the stem (cemented stem model). The third type is a Durom implant with a biomimetic composite stem, osseointegrated along its full length (biomimetic stem model). For all three types of fixation, the same geometry is used (i.e. the Durom geometry). The virtual implant is cemented as per recommendations of the manufacturer (1-mm thick cement mantle underneath the spherical part of the implant to cover the femoral head exposed trabecular bone surface). Therefore, all references to fixation conditions are applicable to the stem fixation only. All implants were positioned at 5° valgus versus the native neck-diaphysis angle; final positioning of the implants was verified by two experienced surgeons (co-authors ML and PAV). A model of the healthy unimplanted femur is also used for comparison purposes.

2.1. The finite element model (FEM)

The FEM uses a personalised femur obtained from CT-scan of a patient who received the implant (52 mm diameter implant). The CT-scan images were segmented and a 3D geometric reconstruction was made in the CATIA V5 CAD software. Heterogeneous patient-specific mechanical properties were then applied to the trabecular bone region. Material properties are based on literature data; a relation used by Rho [19] was applied to obtain the density and elasticity modulus distribution in the trabecular bone ranging between 0.5 and 6 GPa (see Fig. 2). These distributions were found to be in good agreement with experimental data from Morgan and Keaveny [20]. Cortical bone elasticity modulus is set at 17 GPa [21,22]. Both bone materials are modeled as linear elastic isotropic materials.

The cement mantle is modeled using Young modulus of 2.07 GPa [23]. The composite material is modeled using an averaged value of 25 GPa for Young modulus. All bone–metal interfaces are modeled with frictional contact elements ($\mu = 0.4$ [10]); bone–cement and bone–osseointegrated implant are modeled with bonded contacts. Cement–implant interfaces were also modeled with bonded contacts (i.e. a perfectly bonded interface was assumed).

The models are meshed with 3-mm quadratic tetrahedrons based on a mesh convergence study (mesh is shown in Fig. 3a). A less refined mesh (5-mm) was deemed sufficient for the distal part of the cortical bone, as it is far from the region of interest of this study. The resulting models comprise between 122 653 and 131 800 elements (135 889 and 153 108 nodes). Convergence of the contact elements was also verified to ensure realistic contact pressure and load transfer across model parts. The model was compared with previously reported numerical stress and strain distributions of the resurfaced hip, and agreed well with published works [10,24,25].

Table 1
Comparison of material properties from different orthopaedic and bone materials.

Material/tissue	Density (g/cm ³)	Modulus (GPa)	Strength (MPa)	Poisson's ratio
Trabecular bone	0.03–0.12	0.04–1.0	1.0–7.0	0.01–0.35
Cortical bone	1.6–2.0	12–20	150	0.28–0.45
Ti–6Al–4V	4.4–4.7	106	780–1050	0.33
Composite (compression)	1.2–1.6	5–14	53–220	0.3
Composite (tension)		12–30	70–250	0.36

Fig. 3. Finite element model used in the current study (a) mesh, (b) box plot analysis zones (anterior view) and (c) box plot analysis zones (posterior view).

Fig. 4. Graphical representation of load cases used in the bone remodeling simulations (a) heel-strike, (b) abduction and (c) adduction.

Three load cases are applied to simulate a full day worth of normal physical activities [26]. The load cases are presented in Table 2 and Fig. 4. They represent heel-strike, abduction and adduction.

2.2. The anisotropic bone remodeling model

To evaluate long-term bone resorption or apposition with each of the three fixation types, Doblaré's bone remodeling model was applied [27]. This model allows for the evaluation of bone density changes and bone trabeculae reorientation (i.e. bone anisotropy at the continuum level). The bone remodeling model is based on continuum damage theory (CDM) and the oriented damage theory first evoked by Codebois and Sideroff [28]. In the case of bone material, the porosity level is used as a level of damage: for 0% porosity, bone is in an undamaged state and its mechanical properties are at their highest possible values, whereas at 100% porosity, bone has completely resorbed. Bone anisotropy is assessed by means of the fabric tensor $\hat{\mathbf{H}}$ (tensor whose three principal components take the relative values of the material principal moduli of elasticity,

identity tensor for isotropic materials and ovaloid for orthotropic materials such as bone) first developed by Cowin [29]. Fabric tensor and porosity are combined to obtain the remodeling tensor $\mathbf{H}$, the model's internal variable, as per Eq. (1):

$$\mathbf{H}^2 = \left(\frac{B\rho^\beta}{E_{\max}} \right)^{1/2} \hat{\mathbf{H}} \quad (1)$$

where B and β are the coefficients of the relationship used to calculate the elasticity modulus from the apparent density and $E_{\max}$ is the maximal possible value of elasticity modulus.

As with other more traditional bone remodeling models, the continuum stress state obtained from the FEM of a load case is then used to calculate the remodeling stimulus at bone tissue level, using the number of cycles per day as a weight factor for the effect of each load case. The remodeling stimulus $\mathbf{Y}$ is a function of tensor $\mathbf{H}$, is anisotropic and is defined as per Eq. (2):

$$\mathbf{Y} = \frac{\partial \psi(\boldsymbol{\epsilon}, \mathbf{H})}{\partial \mathbf{H}} \quad (2)$$

Table 2

Numerical values and orientation of load cases (LC) used in the bone remodeling simulations.

LC	Cycles/day	Hip contact force			Abductors		
		Force	Orientation		Force	Orientation	
			(°XZ)	(°YZ)		(°XZ)	(°YZ)
A	6000	2317	24	6	703	28	15
B	2000	1158	–15	35	351	–8	9
C	2000	1548	56	–20	468	35	16

Axe Z is parallel to diaphyseal axis.

Axe X is perpendicular to axe Z and is in the femoral frontal plane (plan XZ is the femoral frontal plane).

Axe Y is perpendicular to the other two planes (plan YZ is a latero-sagittal plan).

Angle XZ is positive in anterior direction.

Angle YZ is positive in medial direction.

Table 3

Numerical values of the parameters used in the bone remodeling model.

Parameter	Numerical value (definition)
ψ_t^*	50 MPa (daily reference stimulus at tissue level)
m	4 (exponent of the daily stress stimulus, used to calculate tissue level stimulus from continuum stimulus)
c	0.02 $\mu\text{m/day}$ (constant remodeling velocity both for resorption and apposition)
w	25% (half-width of the dead zone)
Δt	1 day (time step per simulation run)
k	100% (percentage of active bone surface)
ω	0.1 (stimulus anisotropy factor: 0 is a purely isotropic stimulus, 1 is a purely anisotropic stimulus)

where ψ is the free energy function and ϵ is the strain tensor.

The relative influence of spherical (hydrostatic pressure, stress that causes volume change but no distortion) versus octahedral (deviatoric stress that causes distortion but no volume change) components is adjusted via the anisotropy factor ω . The stimulus is compared with a reference value of daily stimulus at tissue level (ψ_t^*) that includes a dead zone, and tensor $\mathbf{H}$ is then updated accordingly in an iterative loading scheme. The three load cases presented in Table 2 and Fig. 4 are applied sequentially with a weight factor (number of cycles per day) to obtain the remodeling stimulus. The mechanical properties (density and elasticity moduli) are then updated and the cycle is repeated until density change reaches a negligible value. The numerical parameters used in the bone remodeling process are described in Table 3. The reader is referred to papers published by Doblaré [27,30,31] for further details concerning this model.

Strain shielding was preferred to stress shielding to compare the models because uniaxial yield strain of bone is uniform within an anatomic site [20,32] and is also believed to be isotropic [32]: this makes principal strains a useable site-independent “yield criterion” for bone material. The maximal compressive strain (ϵ_3) is the highest strain value in magnitude in most of the region of interest of the models and will be used to report results, as opposed to maximal tensile strain that has a lower absolute value range. For comparison purposes, the proximal femur was split into 8 zones labelled as A to H (4 anterior and 4 posterior, 2 of each in the femoral neck and 2 of each in the femoral head, see Fig. 3b and c). Local strains ($\epsilon_3^{\text{compared model}}$) were evaluated at a series of equally spaced points located within the bone volume (position x_i, y_i, z_i). They are reported as an increase/decrease in percentage ($\delta\epsilon_3^{\text{comparison}}$) vs. the reference model ($\epsilon_3^{\text{reference model}}$), as per Eq. (3). These variations in percentages are presented in a box plot graph. The line at 0% change represents a strain level similar to the reference model. A box below this line represents a state of decreased strain relative to the reference model; conversely, a box above represents a state of increased strain relative to the reference model. The boxes present the 25th,

median, and 75th percentiles of the distributions, with the ends of the whiskers at 1.5 times the interval between the 25th and 75th percentiles (99.3% of the data if it is normally distributed).

$$\partial\epsilon_3^{\text{comparison}}(x_i, y_i, z_i)$$

$$= \frac{\epsilon_3^{\text{compared model}}(x_i, y_i, z_i) - \epsilon_3^{\text{reference model}}(x_i, y_i, z_i)}{\epsilon_3^{\text{reference model}}(x_i, y_i, z_i)} \times 100 \quad (3)$$

3. Results

The bone remodeling simulations ended for all three models after 19 iterations (6 sequences of 3 load cases) when the maximal absolute density change reached below 0.01 g/cm³ and the model had converged. The resulting trabecular remodeled bone apparent density is shown in Fig. 5 (cortical bone is not shown, as its density remained constant at the maximal value of 2.031 g/cm³). The density in the latero-superior part of the femoral head is greater for the unfixed and biomimetic stems than the cemented stem. A zone of increased density is observed at the tip of the stem for the biomimetic (0.61 g/cm³) and cemented (0.72 g/cm³) stems, which is not present for the unfixed stem. In the center of the femoral neck, the unfixed model shows a zone where density is smaller (0.58 g/cm³) than in both fixed models.

The principal compressive strain (ϵ_3) distributions for the unfixed, biomimetic and cemented stems are shown in Fig. 6, both before and after the bone remodeling process (heel-strike load case is shown). Before bone remodeling, the supero-lateral part of the femoral head is more strained for the unfixed stem model than it is for the cemented stem (–590 $\mu\epsilon$ vs. –320 $\mu\epsilon$ respectively), with the biomimetic stem featuring an intermediate level of strain (–535 $\mu\epsilon$). The center of the femoral head is more strained in the unfixed stem model (–780 $\mu\epsilon$) than in both the biomimetic (–645 $\mu\epsilon$) and cemented stem models (–615 $\mu\epsilon$). At the tip of the stem, the unfixed stem is less strained (–370 $\mu\epsilon$) than the biomimetic stem (–430 $\mu\epsilon$) or the cemented stem (–500 $\mu\epsilon$), indicating a change in load transfer mechanism. As expected, for both models with a fixed stem (biomimetic and cemented), part of the load is transferred distally through the stem.

After bone remodeling, strain distributions show higher strains for all three models when compared with their unremodeled counterparts. The supero-lateral part of the femoral head remains more strained for the unfixed stem (–1260 $\mu\epsilon$) and the biomimetic stem (–1150 $\mu\epsilon$) than the cemented stem (–780 $\mu\epsilon$). The same observation is made for the center of the femoral head (–1550 $\mu\epsilon$ and –1415 $\mu\epsilon$ for the unfixed and biomimetic stems respectively vs. –1380 $\mu\epsilon$ for the cemented stem respectively). Strains at stem tip for the unfixed stem model (–600 $\mu\epsilon$) are lower than for both fixed stems (–890 $\mu\epsilon$ and –990 $\mu\epsilon$ for the biomimetic and cemented stems respectively).

The principal compressive strain (ϵ_3) differences before bone remodeling are presented in percentages of the unfixed, biomimetic and cemented stems versus the healthy femur on the box-plot graph of Fig. 7 (heel-strike load case is shown). This graph allows for a quantified comparison of the strain distributions presented in Fig. 6, using the healthy femur model as a reference. Zones A to D represent the four quadrants of the femoral head. Zones A and C (supero-lateral part of the femoral head) show strain shielding for all three stem models: the unfixed stem has medians of –6.4 and –22.4%, the biomimetic stem has medians of –21.0% and –34.5% and the cemented stem is the most strain shielded with medians of –33.2% and –47.4%. Zones B and D (infero-medial part of the femoral head) show the unfixed model to be slightly over-strained relative to the healthy femur with medians of +12.7% and +3.5%. The biomimetic and cemented stems are strain shielded with medians

Fig. 5. Apparent bone density (ρ) after the bone remodeling process (only trabecular bone of the proximal femoral bone is shown).

Fig. 6. Maximal compressive strain (ϵ_3) distributions for cemented, biomimetic and unfixed stems, before and after bone remodeling process (cortical and trabecular bone of the proximal femoral bone are shown).

Fig. 7. Box-plot representation of strain differences in percentage between unfixed, biomimetic and cemented stems with healthy femur as reference, before the bone remodeling process.

of -1.3% and -8.4% for zone B and -10.5% and -15.8% for zone D. The four zones in the femoral neck (zones E–H) do not show a clear trend between the three models; however, all medians are positive, indicating that the neck is more strained with either type of implant than the healthy femur.

The principal compressive strain (ϵ_3) differences before and after bone remodeling are presented in percentages of the biomimetic and cemented stems versus the unfixed stem on the box-plot graph of Fig. 8 (heel-strike load case is shown). Before bone remodeling, Fig. 8 indicates that zones A and C (supero-lateral part of the femoral head) of the cemented stem model are the most strain shielded, with medians of -28.1% and -32.3% respectively. The biomimetic stem model has better results with medians of -15.6% and -15.2% for the same two zones A and C. Zones B and D (infero-medial part of the femoral head) are strain shielded at -14.1% and -20.2% for the biomimetic and cemented stems respectively. Zones E–H represent the femoral neck; zone E is strain shielded for the biomimetic stem (-10.2% for median value) and the cemented stem (-13.3% for median value). The other three zones show no clear trend, indicating similar states of strain to the unfixed model.

After bone remodeling, all four quadrants of the femoral head (A–D) in both fixed models are strain shielded when compared with

Fig. 8. Box-plot representation of strain differences in percentage between cemented/biomimetic stems and unfixed stem as reference, before and after the bone remodeling process.

the unfixed model. Fig. 8 also shows that the level of strain shielding is reduced (-30.2% to -24.8% in zones A and C of the cemented stem) as a result of the bone remodeling process. In all cases, the biomimetic stem is closer to the unfixed stem than to the cemented stem. In the femoral neck (E–H), only zone E is observed to be still slightly strain shielded. Results for Zones F–H show that both fixed models lead to strains higher than the unfixed model, indicating that the unfixed model is strain shielded when compared with both fixed models.

4. Discussion

Metaphyseal fixation in femoral component of HR may improve short and mid-term implant survival and allow resurfacing implant to be used in patients with poorer femoral head bone quality [7]. However, cementing the stem increases strain shielding and may compromise long-term performance and survival of the implant. The use of a biomimetic stem in HR could provide an implant with distal fixation producing minimal stress changes in the femoral head. By performing FEMs of a commercially available resurfacing implant with three different fixation types (unfixed metallic stem, biomimetic material osseointegrated stem and cemented metallic stem) it was found that cementing the stem increases strain shielding in the femoral head when compared with the unfixed stem. The results presented in the current study indicate that the superior aspect of the femoral head has strain values reduced by as much as 30% if the stem is cemented, indicating that this zone is at risk of bone resorption. The use of the biomimetic material showed that strain values were reduced by 15% (i.e. a 50% reduction of the strain shielding effect relative to the cemented metallic stem) at the same location, indicating that while the risk of bone resorption is reduced, it is not eliminated. The bone remodeling simulations also indicated density changes due to the change in load transfer mechanism.

One of the main limitations of this study is the fact that it was conducted with only one patient-specific femur and optimally positioned implants; inter-patient variability cannot be considered accounted for in our study. Furthermore, this particular femur may not be an adequate candidate to properly replicate a case in which stem fixation is desired, as it does not have any femoral head deficiency. Finite element bone remodeling simulations also have their

limitations. They can be used effectively in comparing models, but their validity has yet to be demonstrated clinically. Another main limitation of the current study is the modeling method for interfaces: cement–implant, bone–implant and bone–cement interfaces were modeled as perfectly bonded along the entire length of the stem, which is not representative of the clinical situation. Also, the curing conditions of bone–cement affects local interface and bone stresses [33,34]. Load transfer to the surrounding bone would be affected by these effects that are not taken into account in the current study. Furthermore, experimental validation should be conducted on the biomimetic stem to evaluate if its mechanical behaviour can be approximated by an isotropic linear elastic material.

Cementing the stem has been tested clinically, and several studies have found this technique to be effective on patients with poor femoral head bone quality, at least after short periods of follow-up [35,36]. Other finite element studies have shown that cementing the stem could lead to detrimental stress shielding in the femoral head [8,10,24], although the technique does not appear to have any clinically observable detrimental effect [6].

Fixation of a metallic stem by osseointegration rather than cementation has been tested clinically by Lilikakis et al. [37] using a commercially available implant (the Cormet 2000). The implant with all internal surfaces, including the stem, was coated with an hydroxyapatite layer to facilitate osseointegration. It was found that survivorship of the femoral component after 2 years was 98.6%, but neck narrowing was a common radiographic finding. Another clinical study by Katrana et al. [38] found that cemented femoral components led to more neck narrowing than uncemented femoral components, again indicating that implants with cemented stems have load transfer mechanism different from implants with uncemented stems, and that neck narrowing could be an indication of that change in load transfer mechanism. Neck narrowing has also been observed clinically, but has not so far been correlated with clinical symptoms or femoral neck fracture [11]. A recent clinical study showed a 60% occurrence of neck narrowing in a series of 57 hips [39]; valgus positioning of the femoral component was significantly associated with higher diminution of neck diameter, leading the authors to propose altered load paths as the underlying cause of neck narrowing.

The results presented in the current study indicate that a change in load transfer mechanism occurs when the stem of the resurfacing implant is fixed, regardless of whether this fixed stem is metallic or made of biomimetic composite material. The strain increase at the tip of the stem and in the distal part of the femur is reproduced in both models of fixed stems. The use of a more distal fixation allows for load transfer through this part of the femur rather than through the femoral head, which was precisely the goal sought by fixating the stem. The main difference in the strain pattern calculated is the higher (presumably healthier) strain levels in the superior part of the femoral head obtained for the biomimetic stem compared to the cemented metallic stem case. This indicates that an osseointegrated biomimetic stem with bone matching properties should perform better in the long term than a cemented metallic stem.

Because the bone remodeling model used in this study does not allow for bone shape changes, it is not possible to know if the density changes observed in the femoral neck would translate to neck narrowing, or if fixation of the stem would affect its importance. The possibility that the patient whose femur was modeled was not at risk of having neck narrowing cannot be ruled out either. As the aetiology and risk factors of this particular complication are not currently known, it is not possible to assess whether its origin is mostly mechanical (strain shielding) or is related to osteolysis (wear debris). Osteolytic lesions on the femoral neck have been reported in the literature [6], and cementing the stem has not been shown to have an influence on its incidence, although cemented

implants were shown to lead to more occurrences of neck narrowing than their uncemented counterparts [38].

5. Conclusion

The current study is a preliminary finite element analysis showing that cementing the metaphyseal stem of HR femoral components increases strain shielding in the femoral head trabecular bone, particularly in the supero-lateral zone situated underneath the spherical part of the implant, and that the biomimetic stem reduces the phenomenon significantly. An osseointegrated stem made of the new biomimetic composite material can provide the kind of metaphyseal fixation sought by cementing of the stem, without increasing strain shielding as much as a cemented metallic stem does. The bone remodeling simulations suggest that after a few months strain distribution in the femoral head of this new biomimetic implant would be similar to the strain distribution in an unfixed stem metallic implant, indicating that a biomimetic stem, i.e. one with bone matching mechanical properties represents an interesting alternative when fixation extension is desired.

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Conflict of interest

Christiane Caouette and Martin N. Bureau declare that they have no conflict of interests or funding that might bias their work. Pascal-André Vendittoli is consultant for Wright Medical, Stryker and Zimmer. Martin Lavigne is consultant for Wright Medical and Zimmer. Pascal-André Vendittoli and Martin Lavigne have research grants from Zimmer, Stryker, Biomet, Depuy, Smith & Nephew and Wright Medical. Natalia Nuño has a research grant from Zimmer.

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