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1 Influence of the biomechanical environment on the  
2 femoral stem insertion and vibrational behavior: a  
3 3-D finite element study

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10 **Abstract**

11 The long-term success of cementless surgery strongly depends on the im-  
12 plant primary stability. The femoral stem initial fixation relies on multiple  
13 geometrical and material factors, but their influence on the biomechanical  
14 phenomena occurring during the implant insertion is still poorly understood,  
15 as they are difficult to quantify *in vivo*. The aim of the present study is

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16 to evaluate the relationship between the resonance frequencies of the bone-  
17 implant-ancillary system and the stability of the femoral stem under various  
18 biomechanical environments. The interference fit  $IF$ , the trabecular bone  
19 Young's modulus  $E_t$  and the bone-implant contact friction coefficient  $\mu$  are  
20 varied to investigate their influence on the implant insertion phenomena and  
21 on the system vibration behavior. The results exhibit for all the configura-  
22 tions, a non-linear increase of the bone-implant contact throughout femoral  
23 stem insertion, until the proximal contact is reached. While the pull-out  
24 force increases with  $E_t$ ,  $IF$  and  $\mu$ , the bone-implant contact ratio decreases,  
25 which shows that a compromise on the set of parameters could be found in  
26 order to achieve the largest bone-implant contact while maintaining suffi-  
27 cient pull-out force. The modal analysis on the range [2-7] kHz shows that  
28 the resonance frequencies of the bone-implant-ancillary system increase with  
29 the bone-implant contact ratio and the trabecular bone Young's modulus,  
30 with a sensitivity that varies over the modes. Both the pull-out forces and  
31 the vibration behavior are consistent with previous experimental studies.  
32 This study demonstrates the potential of using vibration methods to guide  
33 the surgeons for optimizing implant stability in various patients and surgical  
34 configurations.

35 **Keywords** femoral stem, finite element analysis, interference fit, bone-implant  
36 contact, resonance frequency, primary stability

# 1 Introduction

Each year, around one million Total Hip Arthroplasties (THA) are performed worldwide, which makes it one of the most common surgeries (Pivec et al. 2012; Sloan et al. 2018). However, surgical failures still occur and up to 10% of revision surgeries are necessary within the ten years after implantation (Ulrich et al. 2008; Corbett et al. 2010), leading to additional costs for the healthcare system and higher pain and risks for the patient (Bayliss et al. 2017). The primary stability of cementless implants is crucial for the surgical success (Khanuja et al. 2011) since aseptic loosening or periprosthetic fractures are among the most frequent causes of failures (Abdel et al. 2016).

Both the Acetabular Cup Implant (ACI) and the Femoral Stem (FS) are inserted into the host bone cavity through successive hammer impacts. The initial fixation is achieved thanks to press-fit phenomena at the bone-implant interface due to an undersized host bone cavity previously reamed by the surgeon (Taylor et al. 1995; Kim et al. 2001). The surgical issue is to maximize the primary stability without increasing the risk of periprosthetic fractures, taking into account the inter-individual variability of bone quality and of the anatomical features. On the one hand, micromotions higher than 150  $\mu\text{m}$  should be avoided, since it has been shown that higher values prevent bone ingrowth (Engh et al. 1992) and therefore osseointegration and implant long-term stability. Moreover, a lack of mechanical loading of the bone tissue because of an imperfect bone-implant interface can cause stress-shielding effects (Raffa et al. 2019a, 2021; Hériveaux et al. 2022), which may lead to the formation of fibrous-tissue (Kuiper and Huiskes 1997) and also inhibit optimal bone ingrowth (Khanuja et al. 2011; Engh et al. 2003; Herrera et al. 2007).

61 In particular, several studies commonly observed proximal atrophy for the FS due  
62 to such phenomena (Engh et al. 1992; Herrera et al. 2007). On the other hand,  
63 intra-operative fractures occur around 14 times more often with cementless stems  
64 than with cemented (Abdel et al. 2016; Sidler-Maier and Waddell 2015) and lead  
65 to revision arthroplasties (Fitzgerald et al. 1988). In particular, due to the nature  
66 of the fixation and the shape of the FS implants, the risk of calcar cracks and shaft  
67 fractures is significantly increased for cementless implants (Lamb et al. 2019).

68 The compromise between an optimal fixation and the minimization of fracture  
69 risk may be difficult to find for the surgeon as it depends on multiple factors such  
70 as: i) the FS design (Folgado et al. 2009; Reimeringer et al. 2013; Monea et al.  
71 2014; Russell et al. 2016; Dopico-González et al. 2010), ii) the patient bone quality  
72 (Wong et al. 2005; Shultz et al. 2006), iii) the shape and size of the host bone cavity  
73 (Shultz et al. 2006; Abdul-Kadir et al. 2008), iv) the implant coating and porosity  
74 (Folgado et al. 2009; Monea et al. 2014; Ovesy et al. 2018) and v) the geometry  
75 and mass of the surgical hammer (Bishop et al. 2022). Numerical simulations  
76 are therefore valuable to understand the phenomena occurring during implant  
77 insertion and in particular to analyze the influence of the biomechanical parameters  
78 independently, which is difficult in clinical conditions. While for the ACI, the  
79 primary stability may be assessed through the estimation of the pull-out force  
80 because of the hemi-spherical shape and the localization of the residual stresses  
81 at the ACI rim (Michel et al. 2017; Immel et al. 2021), the Bone-Implant Contact  
82 (*BIC*) ratio and distribution is of particular interest when studying the FS stability,  
83 due to the physiological loads and the implant geometry (Monea et al. 2014; Russell  
84 et al. 2016; Reimeringer and Nuño 2016). However, according to the studies, the  
85 contact ratio achieved after surgery has been reported to vary between 20% and

86 95% (Monea et al. 2014). Moreover, there is a lack of information concerning the  
87 optimal interference fit for the FS (Abdul-Kadir et al. 2008), whereas it has been  
88 studied experimentally and numerically for the ACI (Immel et al. 2021; Michel  
89 et al. 2017), in order to provide the appropriate range of micromotion and polar  
90 gap conducting to successful bone remodeling. Besides, most of the studies on the  
91 FS focus on the implant stability during daily activities such as stair climbing or  
92 gait loading (Reimeringer et al. 2013; Dopico-González et al. 2010; Abdul-Kadir  
93 et al. 2008; Dickinson et al. 2011), that is, when the FS is fully seated. Such  
94 approaches give limited insight with respect to the development of intra-operative  
95 methods to guide the surgeon in the detection of the optimal insertion end-point,  
96 with the ultimate goal to maximize the primary stability and prevent the risk  
97 of peri-prosthetic fractures. To reach this long-term objective, the biomechanical  
98 phenomena occurring during the FS insertion should be better understood.

99 Currently, the surgeons detect the insertion end-point by using their proprio-  
100 ception and in particular by listening to the sound of the impact, which is a highly  
101 subjective approach. Several experimental studies have explored the use of vibra-  
102 tion methods to monitor the FS insertion (Pastrav et al. 2009b; Mulier et al. 2008;  
103 Leuridan et al. 2021; Poudrel et al. 2022). A change of the modal features and  
104 in particular, a shift of the resonance frequencies, has been evidenced as a func-  
105 tion of the implant insertion depth (Poudrel et al. 2022). However, the sensitivity  
106 of vibrational analysis with regard to the *BIC* ratio or the bone quality, which  
107 is relevant for implant insertion optimization, could not be identified, since it is  
108 difficult to measure experimentally. In the literature, the change of the resonance  
109 frequency with bone-implant contact conditions has been studied once the implant  
110 was fully inserted, either for measuring primary stability (Pastrav et al. 2009a) or

111 osseointegration (Pérez and Seral-García 2013). Therefore, the change of contact  
112 is artificial and these studies do not provide the relation between the frequency  
113 change and the implant insertion characteristics.

114 The present work aims to determine a relationship between the resonance fre-  
115 quencies of the insertion tool, named “ancillary” in what follows, and the implant  
116 stability. To do so, a quasi-static 3-D finite element modelling of a FS insertion into  
117 a human femur is proposed and a modal analysis is performed at different steps  
118 during the insertion. The FS primary stability is evaluated through an estimation  
119 of the pull-out force (Shultz et al. 2006; Tijou et al. 2018) and of the bone-implant  
120 contact ratio (Monea et al. 2014). The bone stiffness, the bone-implant friction  
121 coefficient and the interference fit are varied in order to determine whether the  
122 resonance frequency of the bone-implant-ancillary system depends on the FS sta-  
123 bility. The originality of the present work is to determine whether a tool capable  
124 of measuring the vibrations of the ancillary at different insertion steps could be  
125 used in different patients and with different femoral stems to detect the implant  
126 insertion end-point.

127 The paper is organized as follows: after a presentation of the finite element  
128 model, the results for the reference case are shown, for both mechanical and vibra-  
129 tional features. Then, the influence of each parameter on the primary stability is  
130 studied separately. In addition, the sensitivity of the vibration behavior with re-  
131 gard to the trabecular bone Young’s Modulus  $E_t$ , and to the bone-implant contact  
132 is presented.

## 133 2 Material and methods

### 134 2.1 Geometry and mesh

135 The geometrical configuration is described schematically in Fig. 1 where the an-  
136 cillary, the FS and the femoral bone, cut at its distal end (Pérez and Seral-García  
137 2013; Abdul-Kadir et al. 2008) are shown.

138 The ancillary was modeled as a cylinder with the same shape and dimensions  
139 as in (Poudrel et al. 2022), and bonded to the stem for both the insertion stage and  
140 the modal analysis, similarly to the experimental studies (Tijou et al. 2018; Poudrel  
141 et al. 2022). The FS geometry was created from the STL file of a scan of the FS  
142 RMIS implant of size 9 (CERAFIT RMIS, Ceraver, Roissy, France). The geometry  
143 for the human femur was retrieved from a free online database with an appropriate  
144 size to accommodate the stem. The cortical layer was created by extrusion using  
145 Meshmixer (Autodesk, San Rafael, CA, USA), assuming a constant and uniform  
146 thickness of cortical bone tissue of 3 mm, as what was done in (Immel et al.  
147 2021) for an ACI's study. The thickness of the cortical layer was chosen based on  
148 experimental measures on bone mimicking phantoms used in (Tijou et al. 2018) and  
149 (Poudrel et al. 2022) (ORTHObones, 3B Scientific, Hamburg, Germany), to allow  
150 further comparison. The femoral bone cavity was created in ANSYS workbench  
151 (v.20, ANSYS Inc., Canonsburg, PA, USA) by boolean subtraction of the rasp  
152 volume. The rasp corresponds to the FS, undersized with a scale factor such as  
153 the diameter difference between the rasp and the implant, called the interference  
154 fit  $IF$ , was 200  $\mu\text{m}$  in the reference case. In order to obtain different  $IF$ , the FS  
155 scale factor was modified. The choice was made to modify the stem geometry  
156 rather than the bone cavity, in order to keep the same mesh for the bone.

157 The mesh was generated in ANSYS Workbench software (v.20, ANSYS Inc.,  
 158 Canonsburg, PA, USA) and consisted in 427,523 tetrahedral elements, which leads  
 159 to a system of equations with 1,958,289 degrees of freedom. The mesh was finer  
 160 around the bone implant contact (Fig. 1) and a convergence study on the bone  
 161 and implant element size  $h_e$  and the load step increment  $\Delta l_s$  was conducted for the  
 162 reference case. Both mesh and load step increment were refined until the implant  
 163 displacement  $U_I$  and  $BIC$  ratio, defined by the bone surface in contact with the  
 164 FS with regard to the total bone cavity surface, local relative errors were less than  
 165 5%.

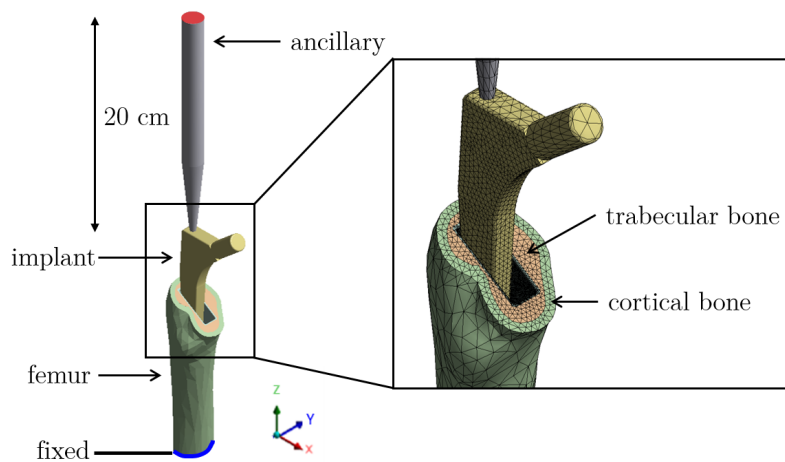


Figure 1: Image of the femur geometry with the FS implant and ancillary and zoom on the mesh close to bone-implant contact. Red elements indicate the region where the load is applied and blue elements indicate the fixed boundary condition.

## 166 2.2 Material, interface modeling and parametric study

167 In order to describe the stem insertion process, the system should be studied under  
 168 the large displacement assumption. In this context, all materials were assumed to

169 have isotropic hypoelastic behavior which may be defined by two elastic constants:  
170 the Young modulus and the Poisson coefficient. Here, all domains were assumed  
171 to be homogeneous and have a same Poisson ratio  $\nu = 0.3$ . The ancillary and the  
172 FS implant were assumed to be made of stainless steel and titanium alloy (Ti-Al6-  
173 V4), respectively. The femoral bone was assumed to be composed of trabecular  
174 bone surrounded with a 3 mm-thick layer of cortical bone. The Young's moduli of  
175 the four domains which constitute the mechanical system are shown in Table 1. A  
176 wide range of trabecular bone Young's modulus  $E_t$  was investigated, according to  
177 physiological values found in the literature (Brown and Ferguson 1980; Bayraktar  
178 et al. 2004; Dickinson et al. 2011). The reference value  $E_t^* = 0.2$  GPa was chosen  
179 according to results on bone mimicking phantoms (Leuridan et al. 2017). As it was  
180 previously shown that changes in the trabecular bone modulus have a greater effect  
181 on proximal micromotions and strains than cortical properties changes (Wong et al.  
182 2005), the cortical bone Young's modulus  $E_c$  was assumed to be fixed and equal  
183 to 18 GPa (Katsamanis and Raftopoulos 1990; Bayraktar et al. 2004).

184 To model different press-fit conditions at the bone-implant interface, interfer-  
185 ence fit  $IF$  values comprised between 100  $\mu\text{m}$  and 400  $\mu\text{m}$  were considered based  
186 on results of experimental (Gebert et al. 2009) and numerical studies (Shultz et al.  
187 2006; Abdul-Kadir et al. 2008; Rothstock et al. 2010). The reference value  $IF^*$   
188 is not well known in the literature, especially because of the surgeon variability  
189 for cavity reaming (Konow et al. 2022) and the specific geometry of the FS. The  
190 reference value  $IF^* = 200$   $\mu\text{m}$  was chosen with a variation between 100 and 400  
191  $\mu\text{m}$  in order to achieve an acceptable value of bone-implant contact ratio at the  
192 end of FS insertion.

193 Moreover, the frictional contact between the bone and the FS implant was

194 modeled using Coulomb’s law (Wriggers 2006), where

$$f_s = |F_t| - \mu|F_n| \leq 0 \quad (1)$$

195  $|F_t|$  and  $|F_n|$  are the norms of tangential and normal components of interface trac-  
196 tion vector, respectively;  $\mu$  is the friction coefficient; and  $f_s$  is a slip criterion which  
197 is negative ( $f_s < 0$ ) when no sliding occurs (sticking) and null ( $f_s = 0$ ) in case  
198 of sliding. The constant friction coefficient  $\mu$  was comprised between 0 and 0.5  
199 for the parametric study in order to consider different types of implant coating  
200 surfaces and different physiological bone-implant contact conditions (Shirazi-Adl  
201 et al. 1993; Dammak et al. 1997; Damm et al. 2015). A reference value of  $\mu^* =$   
202 0.3 was chosen (Immel et al. 2021). The bone cavity was decomposed into four  
203 distinct regions of interest represented in Fig. 2. The two opposite faces paral-  
204 lel to the frontal plane and the two opposite faces parallel to the sagittal plane  
205 constituted the frontal *BIC* (in red in Fig. 2) and the sagittal *BIC* (in blue in  
206 Fig. 2), respectively. For solving the contact problem, the Augmented Lagrangian  
207 Algorithm, which generally leads to better conditioning and convergence with rea-  
208 sonable number of iteration, was employed (Pettersen et al. 2009; Dopico-González  
209 et al. 2010; Reimeringer et al. 2013; Immel et al. 2021).

## 210 **2.3 Loading, boundary conditions and solver settings**

### 211 **2.3.1 Femoral stem quasi-static insertion**

212 All simulations were carried out under quasi static loading conditions (i.e., exclud-  
213 ing inertia and viscosity effects) and taking into account large deformation effects.

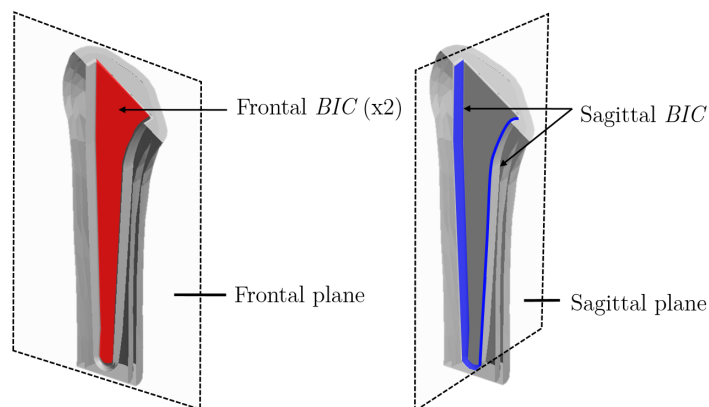


Figure 2: Images of the bone cavity with the *BIC* zones definition regarding the anatomical planes. The frontal and sagittal *BIC* are indicated in red and blue, respectively.

Parameter	Material	Symbol	Density ( $\text{g}/\text{cm}^{-3}$ )	Range	Reference value
Ancillary	Stainless steel	-	7.85	-	200 GPa
Implant	TiAl6Al4V	-	4.4	-	113 GPa
Outer bone	Cortical bone	$E_c$	1.64	-	18 GPa
Inner bone	Trabecular bone	$E_t$	0.27	[0.1-0.6] GPa	0.2 GPa
Interference fit	-	$IF$	-	[100-400] $\mu\text{m}$	200 $\mu\text{m}$
Friction coefficient	-	$\mu$	-	[0.2-0.5]	0.3

Table 1: Material properties of the four subdomains as well as ranges and reference values of the studied parameters: the trabecular bone Young’s modulus  $E_t$  (Brown and Ferguson (1980); Bayraktar et al. (2004); Dickinson et al. (2011); Leuridan et al. (2017)), the interference fit  $IF$  (Abdul-Kadir et al. (2008); Shultz et al. (2006); Rothstock et al. (2010); Gebert et al. (2009)) and the friction coefficient  $\mu$  (Shirazi-Adl et al. (1993); Dammak et al. (1997); Damm et al. (2015); Immel et al. (2021)).

214 A direct solver was used. For all the configurations, the distal region of the femur,  
 215 which is highlighted in blue in Fig. 1, was assumed to be fixed (Taylor et al. 1995).

The simulation was divided into two stages: i) the implant insertion and ii) the implant pull-out to quantify the primary stability. The imposed vertical load  $F_z$  during both the insertion and pull-out stages can be described as a function of

the load step  $l_s$  by:

$$F_z = \begin{cases} f_0/l_{s_1} \times l_s & \text{for } 0 \leq l_s \leq l_{s_1} \\ -2f_0(l_s + l_{s_1})/(l_{s_2} - l_{s_1}) + f_0 & \text{for } l_{s_1} \leq l_s \leq l_{s_2} \end{cases}$$

216 where  $l_{s_1}$  is defined as the load step when the force  $F_z$  starts decreasing and  $l_{s_2}$   
 217 corresponds to the last load step of the simulation, when the contact is lost. The  
 218 load increment between two computations was taken equal to  $\Delta l_s = f_0/50$  during  
 219 the implant insertion in order to ensure Newton-Raphson convergence. To simulate  
 220 the insertion process, a quasi-static vertical load  $F_z$  was applied uniformly on the  
 221 top surface of the ancillary (see Fig. 1) from 0 to  $f_0 = 2.5$  kN, corresponding to  
 222 the load steps  $l_s = 1$  to  $l_{s_1} = 103$ , for all the configurations. The vertical load  
 223 applied for the simulation of the implant insertion phase will be denoted  $F_I$  bellow,  
 224 in order to distinguish between the insertion and pull-out phases. The value of  
 225 the imposed force was chosen based on the experiments of (Raffa et al. 2019b)  
 226 on the ACI. The second stage of the simulation, from  $l_{s_1} = 103$  to  $l_{s_2} = 330$ ,  
 227 corresponds to the implant pull-out from the bone which enables to estimate the  
 228 implant pull-out force  $F_p$ .

### 229 2.3.2 Bone-stem-ancillary system vibration analysis

230 At each of ten equally distributed load steps during the FS insertion, pre-stressed  
 231 modal analysis was performed on the entire bone-implant-ancillary system, sim-  
 232 ilarly at what was experimentally done in (Poudrel et al. 2022). For the modal  
 233 analysis, the bone-implant interface was assumed to be fully bonded. Only the  
 234 modes within the frequency range [2-7] kHz were considered in order to allow

235 comparison with a previous experimental study which demonstrated the sensitiv-  
236 ity of several modes regarding implant insertion within this range (Poudrel et al.  
237 2022). Both the resonance frequencies and associated mode shapes were analyzed  
238 throughout the implant insertion.

## 239 **2.4 Femoral stem insertion monitoring and primary sta-** 240 **bility evaluation**

241 The insertion of the FS implant was monitored by analyzing i) the implant vertical  
242 displacement  $U_I$ , ii) the *BIC* ratio variation with respect to the insertion force  $F_I$   
243 and iii) the variation of the bone-stem-ancillary system's resonance frequencies  
244 with regard to the *BIC* ratio (Pastrav et al. 2009b; Poudrel et al. 2022).

245 The FS primary stability was evaluated by i) the value of the pull-out force  $F_p$   
246 (Shultz et al. 2006) as what has previously been done for the ACI study (Raffa  
247 et al. 2019b; Immel et al. 2021; Doyle et al. 2020) and ii) the *BIC* ratio at the end  
248 of insertion (Monea et al. 2014), that is for  $F_I = 2.5$  kN. The pull-out force  $F_p$  is  
249 defined as the value at the happening of the detachment of the implant from the  
250 bone, identified by a slope discontinuity with a brutal decrease of the *BIC* ratio.

# 251 **3 Results**

## 252 **3.1 Reference case**

### 253 **3.1.1 Bone-implant contact characteristics during implant insertion**

254 A typical variation of the macroscopic *BIC* evolution during the implant insertion  
255 and pull-out stages is represented in Fig. 3 for the reference case. Four distinct

256 phases may be identified:

- 257 1. contact positioning ( $ls \in [0, 60]$ , highlighted in grey): at first,  $BIC = 0$   
258 as there is no contact between the bone and the implant. As soon as the  
259 contact is established ( $ls = 28$ ), the  $BIC$  suddenly increases until reaching  
260 a "plateau" around 40%. This constant value of the  $BIC$  corresponds to  
261 a regime in which the non-sliding contact is dominant at the bone-implant  
262 interface ( $|F_t| < \mu|F_n|$ ).
- 263 2. implant insertion ( $ls \in [60, 103]$ ): this phase corresponds to a regime in which  
264 the sliding contact is dominant at the bone-implant interface ( $|F_t| = \mu|F_n|$ ).  
265 A detailed description of the  $BIC$  evolution during insertion is given in Fig.  
266 4.
- 267 3. implant pull-out ( $ls \in [103, 308]$ ): slight decrease of the  $BIC$  until a sudden  
268 drop to 0% : the slope discontinuity gives the load step corresponding to the  
269 pull-out force  $F_p$ , (here  $ls = 305$ ).
- 270 4. in this last phase, the implant is totally removed from the bone cavity.

271 Figure 4 studies the macroscopic  $BIC$  evolution during FS implant insertion  
272 for the reference case, considering the different regions of interest in the cavity  
273 presented in Fig. 2. The red and blue curves correspond to the frontal  $BIC$  and  
274 the sagittal  $BIC$ , respectively. Four phases of the  $BIC$  can be identified: 1) contact  
275 positioning (**A**), 2) frontal  $BIC$  increase (**B**), 3) sagittal  $BIC$  increase (**C**) and 4)  
276 total  $BIC$  convergence (**D**). At the end of the FS insertion, that is for  $F_l = 2.5$  kN,  
277 the  $BIC$  reaches 75% of the total bone cavity surface and no further significant  
278 contact increase is observed from 2 kN.

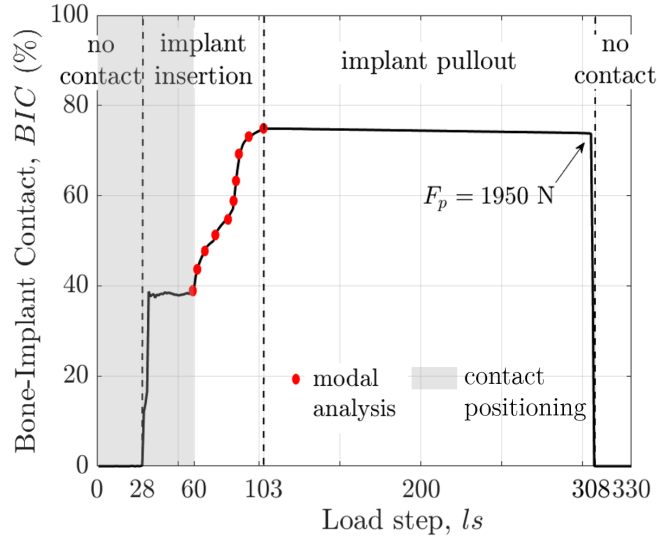


Figure 3: Variation of the  $BIC$  ratio as a function of the load step  $ls$  during implant insertion and pull-out phases for the reference case  $E^*$ ,  $\mu^*$ ,  $IF^*$ . The red circles indicate the load steps where modal analyses are performed during the insertion phase.

### 279 3.1.2 Modal characteristics during femoral stem insertion

280 The mode shapes and the resonance frequencies in the frequency range [2-7] kHz  
 281 are represented in Fig. 5 with the same scale factor (0.005) used for the graphical  
 282 representation of all the deformed shapes. The bone is not shown for a better  
 283 visualization. Eight modes are represented for which the vibration of the ancillary  
 284 is of bending nature with two or three nodes. The mode shapes for which the  
 285 maximum displacement of the ancillary is largely inferior to the maximum dis-  
 286 placement of the bone or the femoral stem are not represented since they would  
 287 not be measurable experimentally.

288 Figure 6 shows the variation of the resonance frequency of the eight modes as  
 289 a function of the  $BIC$  (Fig. 6a) and the implant relative displacement compared  
 290 to the bone (Fig. 6b). Only  $BIC$  values higher than around 38% are shown since

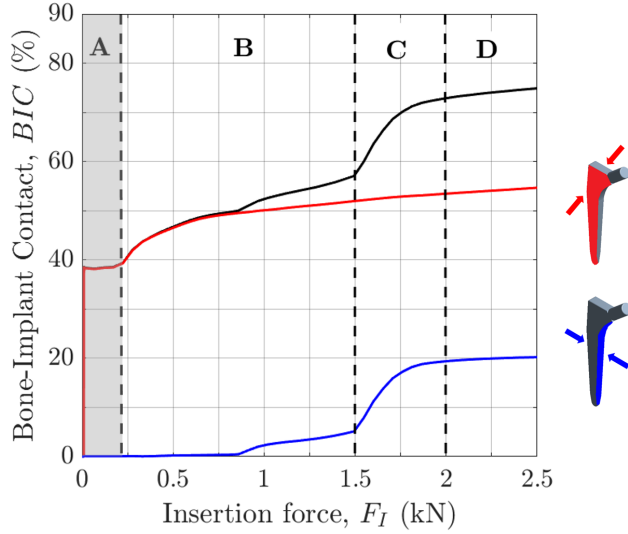


Figure 4: Variation of the  $BIC$  ratio considering the full bone cavity (black line), the bone cavity faces parallel to the sagittal plane (blue line) and the bone cavity faces parallel to the frontal plane (red line), as a function of the insertion force  $F_I$  for the reference case  $E^*$ ,  $\mu^*$ ,  $IF^*$ .

291 this level is achieved as soon as the implant is manually positioned in the bone  
 292 cavity. The three modes  $f_3$ ,  $f_4$  and  $f_5$  obtained from the simulations correspond  
 293 to the modes  $2Y$ ,  $2X$  and  $2Y_b$  measured experimentally in (Poudrel et al. 2022) in  
 294 terms of bending shapes and of values of the resonance frequency. The resonance  
 295 frequencies of the modes  $2Y$  and  $2Y_b$  increase until reaching a value close to 3.8  
 296 kHz and 4.5 kHz, respectively, which is consistent with our experimental study.  
 297 Moreover the mode  $2X$  shows a smaller resonance frequency variation, especially  
 298 for  $BIC > 55\%$ , which is in good agreement with what was obtained experimen-  
 299 tally in (Poudrel et al. 2022). Therefore, in what follows, only these three modes  
 300 will be selected for the study on the sensitivity to the trabecular bone Young's  
 301 modulus  $E_t$ .

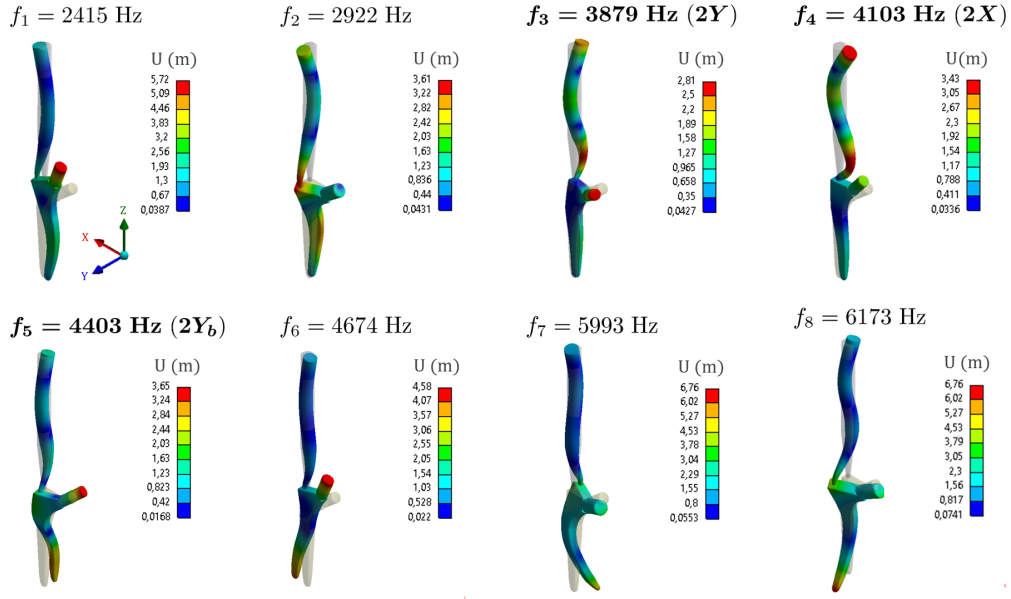


Figure 5: Mode shapes and corresponding resonance frequencies in the range [2-7] kHz obtained at the end of the FS insertion ( $F_I = 2.5$  kHz), for the reference case  $E^*$ ,  $\mu^*$ ,  $IF^*$ . The modes 2Y, 2X and 2Y<sub>b</sub>, highlighted in bold, corresponds to the experimental modes identified in (Poudrel et al. 2022).

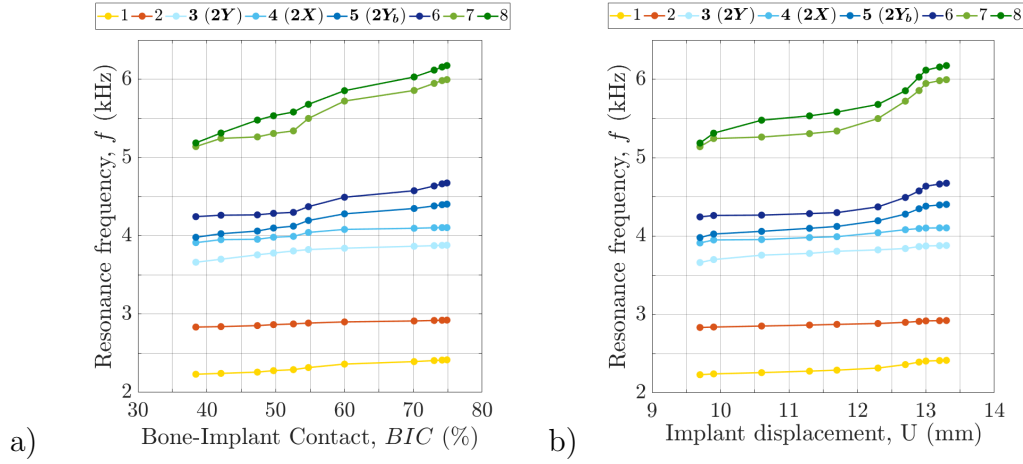


Figure 6: Variation of the bone-implant-ancillary system resonance frequencies  $f$  in the range [2-7] kHz as a function of (a) the  $BIC$  ratio and (b) the implant displacement  $U_I$ . The modes 2Y, 2X and 2Y<sub>b</sub> are highlighted in bold and correspond to the experimental modes identified in (Poudrel et al. 2022).

## 302 **3.2 Implant insertion monitoring under different biome-** 303 **chanical parameters**

### 304 **3.2.1 Effect of trabecular bone Young's modulus**

305 Figure 7 shows the influence of the trabecular bone Young's modulus  $E_t$  on the  
306 variation of the implant displacement  $U_I$  and on the  $BIC$  as a function of the  
307 insertion force  $F_I$ . Both the implant displacement  $U_I$  and the  $BIC$  increase when  
308  $E_t$  decreases. For the three smallest values of  $E_t$  ( $E_t = 0.1$  GPa,  $E_t^* = 0.2$   
309 GPa and  $E_t = 0.3$  GPa), the curves corresponding to the variation of the implant  
310 displacement  $U_I$  as a function of  $F_I$  exhibit a slope discontinuity, which corresponds  
311 to the jump of the  $BIC$  evolution explained by the establishment of the sagittal  
312  $BIC$  (see Fig. 4 and Fig. 7b). The value of  $F_I$  where this discontinuity occurs  
313 increases as a function of the Young's modulus  $E_t$ . At the end of the insertion  
314 stage ( $F_I = 2.5$  kN), the  $BIC$  varies from 50% to 82% depending on the value of  
315 the trabecular bone Young's modulus, which corresponds to a difference of implant  
316 displacement  $U_I$  close to 2.5 mm.

### 317 **3.2.2 Effect of friction coefficient**

318 Figure 8 shows the influence of the friction coefficient  $\mu$  on the variation of the  
319 implant displacement  $U_I$  (a) and the  $BIC$  (b) as a function of the insertion force  
320  $F_I$ . The implant displacement  $U_I$  and the  $BIC$  are higher for smaller values of  
321 the friction coefficient  $\mu$ . Except for the highest value of the friction coefficient  
322 ( $\mu = 0.5$ ), the curves corresponding to the variation of the implant displacement  
323  $U_I$  as a function of  $F_I$  exhibit a slope discontinuity, which corresponds to the jump  
324 of the  $BIC$  evolution explained by the establishment of the sagittal  $BIC$  (see Fig.

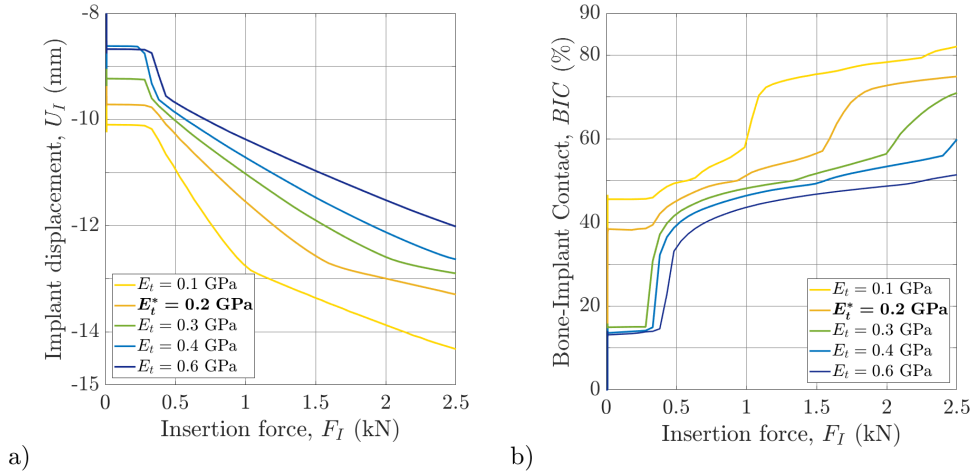


Figure 7: Variation of (a) the implant displacement  $U_I$  and (b) the  $BIC$  ratio as a function of the insertion force  $F_I$  for different values of the trabecular bone Young's modulus  $E_t$ . The reference case  $E_t^*$  is highlighted in bold.

325 4 and Fig. 8b). At the end of insertion ( $F_I = 2.5$  kN), the difference of implant  
 326 displacement  $U_I$  reaches 1 mm for friction coefficient  $\mu$  comprised in the range  
 327 [0.2-0.5]. It is worth noting that this small difference in the displacement, difficult  
 328 to measure in the clinic, corresponds to a variation of  $BIC$  of nearly 20%.

### 329 3.2.3 Effect of interference fit

330 Figure 9 shows the influence of the interference fit  $IF$  on the variation of the  
 331 implant displacement  $U_I$  (a) and the  $BIC$  (b) as a function of the insertion force  
 332  $F_I$ . The values of implant displacement  $U_I$  and of the  $BIC$  for any insertion force  
 333  $F_I$  decrease as a function of the interference fit  $IF$ . For the three smallest values  
 334 of  $IF$  ( $IF = 100$   $\mu\text{m}$ ,  $IF = 150$   $\mu\text{m}$ ,  $IF = 200$   $\mu\text{m}$ ), the curves corresponding to  
 335 the variation of the implant displacement  $U_I$  as a function of  $F_I$  exhibit a slope  
 336 discontinuity, which corresponds to the jump of the  $BIC$  evolution explained by  
 337 the establishment of the sagittal  $BIC$  (see Fig. 4 and Fig. 8b). For values of  $IF$

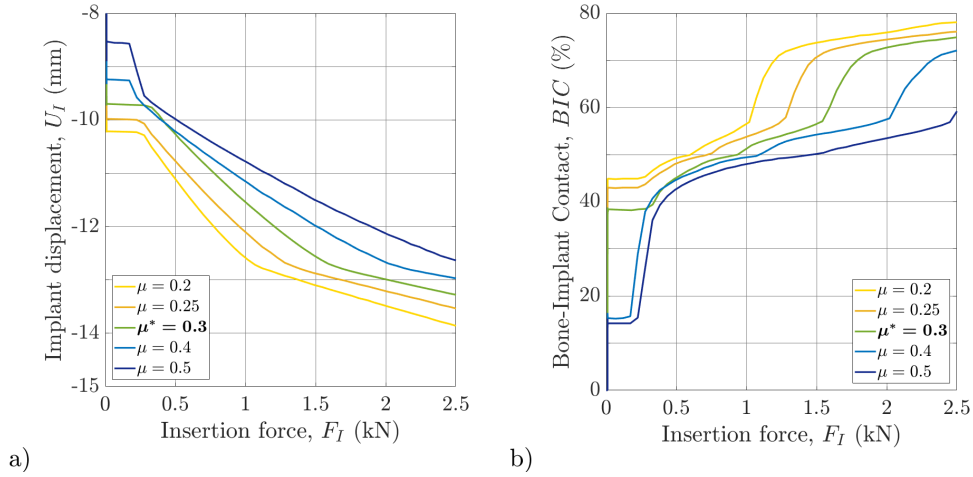


Figure 8: Variation of the implant displacement  $U_I$  (a) and the  $BIC$  ratio (b) as a function of the insertion force  $F_I$  for different values of the friction coefficient  $\mu$ . The reference case  $\mu^*$  is highlighted in bold.

338 between 100  $\mu\text{m}$  and 400  $\mu\text{m}$ , the values of the implant displacements  $U_I$  at the  
 339 end of the insertion ( $F_I = 2.5$  kN) are equal to 15 mm and 9 mm, respectively,  
 340 while the corresponding  $BIC$  ratio varies from 80% to 60%. Increasing the value  
 341 of the interference fit leads to lower implant displacement  $U_I$  and lower  $BIC$  value  
 342 reached at the end of the insertion, which may be explained by an increase of the  
 343 normal forces due to contact pressure. For the range of variations considered in  
 344 the present study for  $\mu$  and  $IF$ , the effect of the interference fit on the implant  
 345 displacement is lower than that of the friction coefficient.

### 346 3.3 Variation of the FS implant stability

347 Figure 10 shows the variation of the pull-out force  $F_p$  and of the  $BIC$  ratio at  
 348 the end of insertion stage ( $F_I = 2.5$  kN) as a function of (a) the trabecular bone  
 349 Young's modulus  $E_t$ , (b) the friction coefficient  $\mu$  and (c) the interference fit  $IF$ .  
 350 For each parametric study, the results are given with all other parameters fixed at

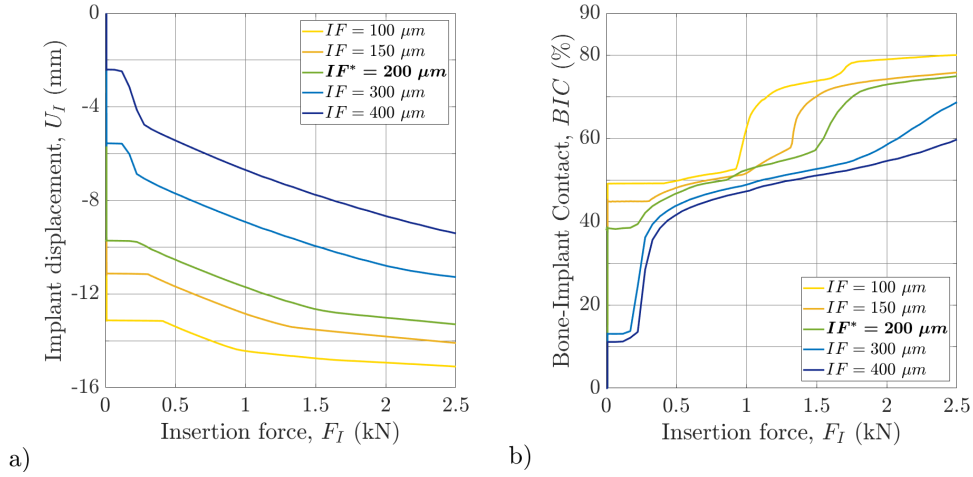


Figure 9: Variation of the implant displacement  $U_I$  (a) and the  $BIC$  ratio (b) as a function of the insertion force  $F_I$  for different values of the interference fit  $IF$ . The reference case  $IF^*$  is highlighted in bold.

351 the reference value, noted by \*. Figure 10 shows an increase of the pull-out force  
 352 with the parameters  $E_t$ ,  $\mu$  or  $IF$ . Conversely, the  $BIC$  ratio decreases as a function  
 353 of  $E_t$ ,  $\mu$ ,  $IF$ . The pull-out force  $F_p$  has a strong non-linear behavior as a function  
 354 of  $IF$ , with a strong increase for  $IF$  values between 100  $\mu\text{m}$  and 150  $\mu\text{m}$  and a  
 355 significantly lower increase for  $IF > 150 \mu\text{m}$ . A good compromise is obtained for  
 356 the reference case with the maximization of both the  $BIC$  and the pull-out force  
 357  $F_p$  reaching 75% and 1.95 kN, respectively. The values of the pull-out force  $F_p$   
 358 found in Fig. 10 are again in good agreement with experimental results obtained  
 359 on bone mimicking phantoms (Tijou et al. 2018). An optimal range of values can  
 360 be defined for  $\mu$ , and  $IF$  (an optimal value of  $E_t$  is not searched here as it is  
 361 not a parameter controllable by the surgeon), which optimize the FS stability by  
 362 maximizing both the pull-out force  $F_p$  and the  $BIC$  ratio. For instance, friction  
 363 coefficient values between 0.25 and 0.35 and  $IF$  values between 150  $\mu\text{m}$  and 250  
 364  $\mu\text{m}$  provide FS stability such as  $F_p > 1.9 \text{ kN}$  and  $BIC > 70\%$ .

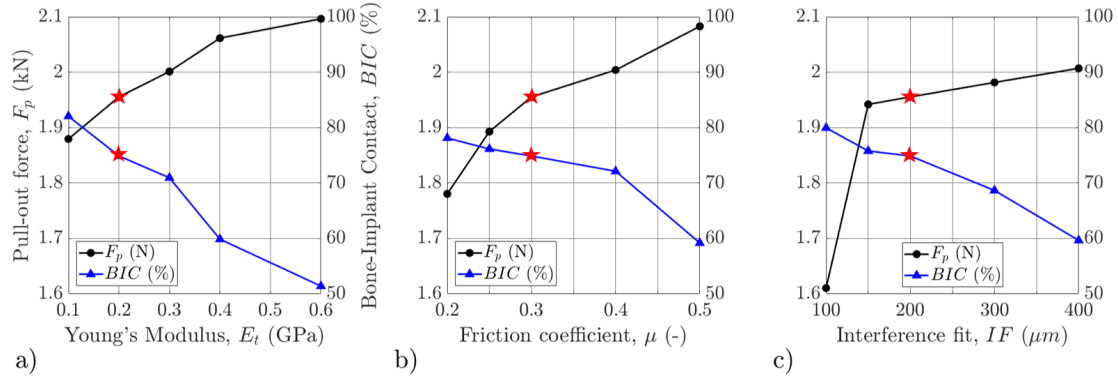


Figure 10: Variation of the pull-out force  $F_p$  and the  $BIC$  ratio measured at the end of the FS insertion ( $F_I = 2.5$  kN) as a function of the trabecular bone Young's modulus  $E_t$  (a), the friction coefficient  $\mu$  (b) and the interference fit  $IF$  (c). The red stars indicate the values obtained for the reference case  $E_t^*$ ,  $\mu^*$  and  $IF^*$ .

### 3.4 FS insertion monitoring by means of modal analysis

Figures 11, 12 and 13 show the variation of the resonance frequencies of modes  $2Y$  (a),  $2X$  (b) and  $2Y_b$  (c) as a function of the  $BIC$  ratio evolution during implant insertion for different values of trabecular bone Young's modulus  $E_t$ , friction coefficient  $\mu$  and interference fit  $IF$ , respectively. Each marker is the result of the modal analysis performed during implant insertion, noted by red circles in Fig. 3. For all values of the bone stiffness, of the friction coefficient and of the interference fit, the resonance frequencies of the modes  $2Y$  and  $2Y_b$  increase with the  $BIC$  ratio in the ranges considered herein. The mode  $2X$  appears to be less sensitive than the two others modes to the implant insertion. Moreover, for a constant value of the  $BIC$  ratio, while the bone stiffness influences the resonance frequencies (see Fig. 11), they do not depend on the friction coefficient (see Fig. 12) and on the interference fit (see Fig. 13). In particular, for a given  $BIC$  ratio, the value of the resonance frequencies of mode  $2Y$  and  $2Y_b$  increases when the bone stiffness increases (see Fig. 11). The mode  $2X$  does not seem to be influenced by trabecular bone Young's

380 modulus  $E_t$  higher than 0.3 GPa. These results are consistent with the boundary  
 381 conditions considered between the bone and the implant during the modal analy-  
 382 ses since the friction coefficient only influences the insertion phenomena (implant  
 383 displacement and  $BIC$  ratio, see Fig. 8) and not the modal analysis during which  
 384 the bone-implant interface is considered to be linear and fully bounded. There-  
 385 fore, considering the same level of  $BIC$  ratio, which correspond to similar implant  
 386 positions, the system resonance frequencies are not significantly different.

387 More specifically, an approximately linear relation between the resonance fre-  
 388 quency and the  $BIC$  ratio is obtained for each mode and each value of  $E_t$ . The  
 389 sensitivity of each mode to the  $BIC$  may be assessed by the slope of the linear  
 390 regression curve for each value of  $E_t$ , which is indicated in Table 2 together with  
 391 the corresponding correlation coefficient  $R^2$ . Considering the reference case, the  
 392 most sensitive mode is the mode  $2Y_b$ , with a mean sensitivity of 14.03 Hz/%. This  
 393 sensitivity is similar for the different values of  $E_t$  within the range [0.1 0.6] GPa.  
 394 The least sensitive mode is  $2X$ , with a slope inferior to 2 Hz/% for  $E_t = 0.3$  GPa,  
 395  $E_t = 0.4$  GPa and  $E_t = 0.6$  GPa. However, the sensitivity of  $2X$  is higher for the  
 396 smallest trabecular bone Young's modulus, reaching up to 12.95 Hz/% for  $E_t = 0.1$   
 397 GPa, which may be explained by large differences of implant position for the same  
 398 level of  $BIC$  ratio. In addition, the sensitivity of the resonance frequency to the  
 399 trabecular bone Young's modulus  $E_t$  is evaluated at a given  $BIC$  ratio, indicated  
 400 by the vertical dashed line at  $BIC = 50\%$  in Fig. 11. The resonance frequency of  
 401 each mode obtained at  $BIC = 50\%$  is noted  $f_{BIC=50\%}$  and the values are shown  
 402 in Table 2 for the different values of  $E_t$ . The mode  $2Y_b$  is the most sensitive to  
 403 the trabecular bone Young's modulus  $E_t$ , with a variation of more than 1.1 kHz  
 404 obtained for values of  $E_t$  varying between 0.1 GPa and 0.6 GPa. The modes  $2Y$

405 and  $2X$  are more sensitive to changes of the trabecular bone Young's modulus  
 406 when small values of  $E_t$  are considered. In particular, the resonance frequency of  
 407 the mode  $2X$  is nearly constant for  $E_t > 0.3$  GPa, with a frequency variation lower  
 408 than 30 Hz between  $E_t = 0.3$  GPa and  $E_t = 0.6$  GPa.

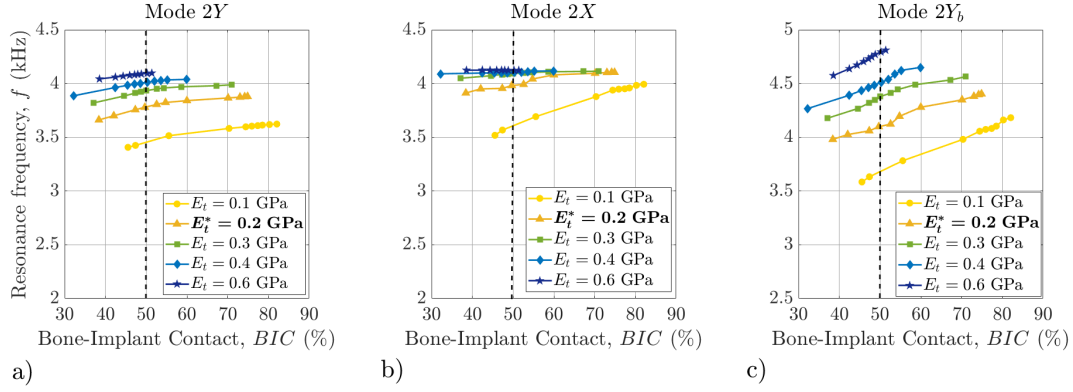


Figure 11: Variation of the bone-stem-ancillary system resonance frequencies  $f$  corresponding to the mode  $2Y$  (a), the mode  $2X$  (b) and the mode  $2Y_b$  (c) as a function of the  $BIC$  ratio for different values of the trabecular bone Young's modulus  $E_t$ . The vertical dashed line at  $BIC = 50\%$  is used to determine the sensitivity of  $f$  to  $E_t$  for a constant value of the  $BIC$  ratio by assessing  $f_{BIC=50\%}$ , which corresponds to the value of  $f$  for  $BIC = 50\%$  (see Table 2).

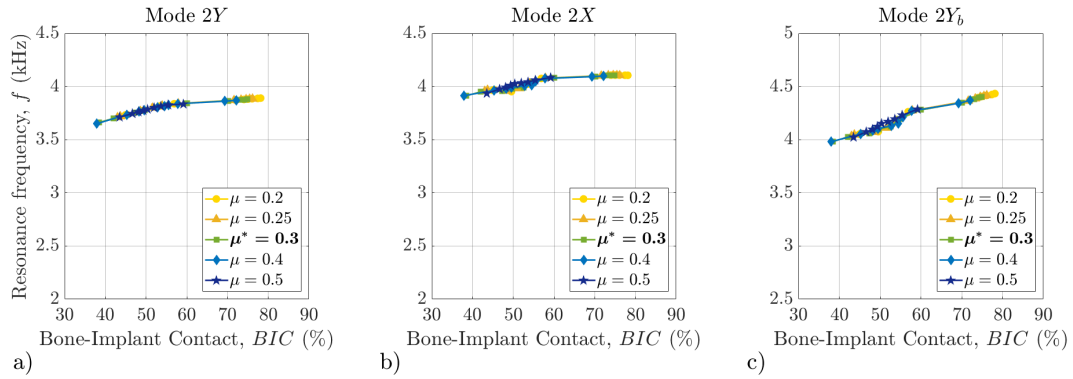


Figure 12: Variation of the bone-stem-ancillary system resonance frequencies  $f$  corresponding to the mode  $2Y$  (a), the mode  $2X$  (b) and the mode  $2Y_b$  (c) as a function of the  $BIC$  ratio for different values of the friction coefficient  $\mu$ .

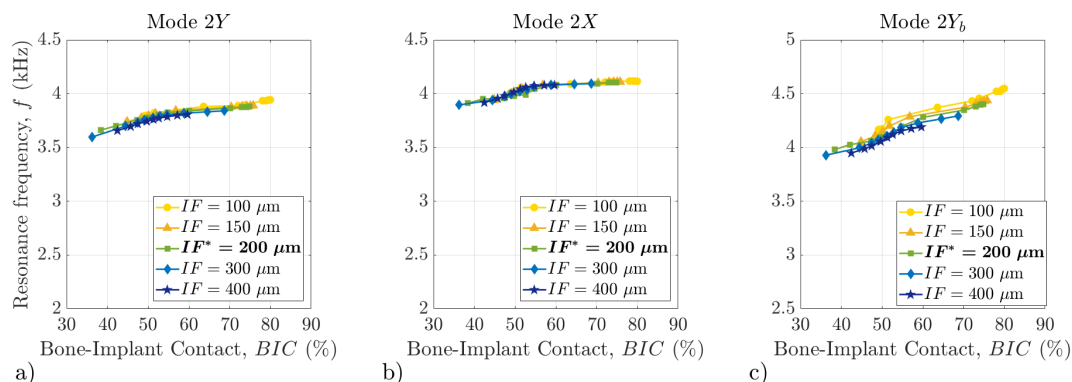


Figure 13: Variation of the bone-stem-ancillary system resonance frequencies  $f$  corresponding to the mode  $2Y$  (a), the mode  $2X$  (b) and the mode  $2Y_b$  (c) as a function of the  $BIC$  ratio for different values of the interference fit  $IF$ .

$E_t$ (GPa)		0.1	0.2	0.3	0.4	0.6	Mean	Std
slope (Hz/%)	$2Y$	5.81	5.28	4.53	5.75	4.34	5.14	0.68
	$2X$	12.95	5.26	1.80	1.04	-0.17	4.13	5.30
	$2Y_b$	12.95	11.86	11.47	14.82	19.07	14.03	3.10
$R^2$	$2Y$	0.97	0.90	0.83	0.95	0.99	0.93	0.06
	$2X$	0.99	0.93	0.85	0.87	0.57	0.84	0.16
	$2Y_b$	0.99	0.98	0.95	0.98	0.99	0.98	0.02
$f_{BIC=50\%}$ (Hz)	$2Y$	3453.6	3781.9	3933.3	4018.8	4093.5	3856.2	253.1
	$2X$	3680.5	3982.4	4093.5	4101.7	4123.0	3996.2	184.8
	$2Y_b$	3607.2	4101.3	4377.4	4507.2	4790.1	4276.6	448.8

Table 2: Slope and determination coefficient ( $R^2$ ) corresponding to the linear regression analysis of the variation of the resonance frequency as a function of the  $BIC$  ratio for the modes  $2Y$ ,  $2X$  and  $2Y_b$ . Values of the resonance frequency of the modes  $2Y$ ,  $2X$  and  $2Y_b$  evaluated at  $BIC = 50\%$  for different trabecular bone Young's modulus  $E_t$  (see dashed lines in Fig. 11).

## 409 4 Discussion

410 This work aims to provide more insight into the influence of various biomechan-  
411 ical parameters on the FS insertion and primary stability, which could lead to  
412 eventually develop a vibration-based method to guide surgeons during the inser-  
413 tion procedure. The contact ratio and the pull-out force depend on the trabecular  
414 bone Young's modulus  $E_t$ , the friction coefficient  $\mu$  and the interference fit  $IF$ . The

415 resonance frequencies of the bone-implant-ancillary are shown to be sensitive to  
416 the bone-implant contact *BIC* ratio and to the trabecular bone Young's modulus  
417  $E_t$ . The corresponding mode shapes and the variation of the resonance frequencies  
418 during the FS insertion are in good agreement with experimental measurements  
419 performed in a previous work by our group (Poudrel et al. 2022).

## 420 4.1 Bone-implant contact evolution

421 During the FS insertion, the *BIC* evolution is found to increase non-linearly with  
422 the insertion force for all configurations, which is in good agreement with the  
423 dynamic finite element study of (Monea et al. 2014) on the evolution and dis-  
424 tribution of the bone-implant contact during and after the FS insertion. This  
425 non-linear increase of the *BIC* ratio may be explained by the particular shape of  
426 the FS implant : the *BIC* at the sagittal bone cavity faces is established after  
427 the *BIC* at the frontal bone cavity faces (see Fig. 4). In particular, the contact  
428 reached in the calcar zone, which is part of the sagittal *BIC* is of great interest  
429 for the surgeons because this femur region is subjected to periprosthetic fractures  
430 (Abdel et al. 2016) and stress-shielding effects (McCarthy et al. 1991; Joshi et al.  
431 2000; Khanuja et al. 2011). The contact evolution obtained for the reference case  
432 indicates that continuing the stem insertion into the cavity for insertion forces  $F_I$   
433 higher than 2 kN does not lead to a significant increase in *BIC* ratio (see Fig. 4).  
434 This observation validates the choice of  $IF^* = 200 \mu\text{m}$  as a reference value and  
435 confirms the need for the surgeons to detect the insertion end-point in order to  
436 avoid stress concentration which could lead to periprosthetic bone fractures (Kim  
437 et al. 2001) and to surgical failure.

## 438 4.2 Femoral stem pull-out

439 The simulated pull-out forces are comprised between 1611 N and 2096 N, which is  
440 in good agreement with experimental values obtained on bone mimicking phantoms  
441 in (Tijou et al. 2018), where the pull-out forces were found to be between 1000 and  
442 2300 N according to the stability configuration. The pull-out force is found to be an  
443 increasing function with respect of the bone’s stiffness, the friction coefficient and  
444 the interference fit. It is worth noting that these results are new findings, because  
445 in the literature, the other studies on FS’s biomechanical behavior rather focused  
446 on the micromotion than the pull-out force evaluation. The fact that the pull-out  
447 force of the femoral stem increases as a function of various parameters (interference  
448 fit, bone stiffness, friction) may be helpful to improve the implant design (friction  
449 coefficient) and the insertion procedure (interference fit) in order to maximize the  
450 implant stability. Moreover, to maximize the FS primary stability, the *BIC* ratio  
451 should also be considered in addition to the pull-out force, as discussed in Section  
452 4.3.

453 In previous studies on the ACI, it was shown that the pull-out force increases  
454 with these same parameters until an optimal value and then decreases for higher  
455 values (Raffa et al. 2019b; Immel et al. 2021). This behavior may be explained  
456 by the fact that highly concentrated contact stresses appearing at the equatorial  
457 rim of ACI strongly increase with the increasing variation of  $(E_t, \mu, IF)$ , which  
458 may prevent the implant to be completely inserted into the bone cavity under the  
459 physiological driving force, leading to a worse pull-out force. However, in the case  
460 of FS considered herein, as the bone-implant contact surface is much larger, the  
461 contact force is not sufficiently strong, even for the biggest values of  $(E_t, \mu, IF)$ ,

462 to prevent the complete insertion of the implant into the bone cavity under the  
463 considered imposed force. Consequently, the pull-out force of FS was found to  
464 monotonously increase with the aforementioned parameters. A simplified cone-  
465 shape implant model from (Shultz et al. 2006) confirms our finding concerning the  
466 pull-out force, which is found to increase with the interference fit or the friction  
467 coefficient, with values between 8000 N and 9000 N. The higher pull-out forces  $F_p$   
468 observed in (Shultz et al. 2006) than in Fig. 10 may be due to the use of cortical  
469 bone only, which has a Young's modulus 100 times higher than the trabecular bone  
470 material used in the present finite element model. This interpretation is coherent  
471 with the increase of the pull-out force with the trabecular bone Young's modulus  
472  $E_t$  shown in Fig. 10a.

### 473 **4.3 Optimal configurations for maximizing FS primary sta-** 474 **bility**

475 While the polar gap and the pull-out force are key determinants of the ACI stability  
476 (Immel et al. 2021; Raffa et al. 2019b; Doyle et al. 2020), the FS insertion success  
477 mainly depends on the *BIC* ratio and the stress distribution in the host bone  
478 (Kim et al. 2001; Monea et al. 2014; Russell et al. 2016; Reimeringer and Nuño  
479 2016). Due to the anatomical distribution of muscular and joint loads, optimizing  
480 the FS pull-out force in the case of the FS is less crucial for the surgical success  
481 than minimizing micromotion at the bone-implant interface, which influence the  
482 osseointegration and the bone remodelling phenomena (Engh et al. 1992; Herrera  
483 et al. 2007; Folgado et al. 2009). Therefore, the effect of the bone quality, of  
484 the friction coefficient at the bone-implant interface and of the stem geometry

485 has been widely studied in the literature for the optimization of the long-term  
486 stability evaluated from the micromotion retrieved during daily activities such  
487 as stair climbing or walking (Dopico-González et al. 2010; Dickinson et al. 2011;  
488 Reimeringer et al. 2013; Russell et al. 2016). However, the pull-out force remains  
489 useful to evaluate quantitatively the primary stability (Tijou et al. 2018) in order  
490 to identify factors influencing the immediate fixation strength as bone property  
491 (Shultz et al. 2006) or implant design.

492 Concerning the effect of bone quality, it has been shown in the literature that  
493 both the reduction of the elastic modulus of cortical and trabecular bone caused  
494 an increase of the initial micromotion and the bone strain at the bone-implant  
495 interface (Wong et al. 2005). As shown in Fig. 10a, the initial fixation strength  
496 decreases when  $E_t$  decreases, which may be explained by lower interface stresses  
497 even if the global bone-implant contact is higher. This result proves the impor-  
498 tance to not only consider the pull-out force but also the level of the *BIC* ratio  
499 when studying FS stability. This same results also emphasizes the necessity to  
500 consider the bone quality parameter when comparing with experimental results.  
501 Concerning the influence of the friction coefficient, it was found in the literature  
502 that higher levels of coefficient of friction are beneficial to primary fixation (Shultz  
503 et al. 2006), which is in good agreement with our results presented in Fig. 10b  
504 where the pull-out force increases from 1780 N to 2090 N when the friction coeffi-  
505 cient increases from  $\mu = 0.2$  to  $\mu = 0.5$ . However, a friction coefficient of  $\mu = 0.5$   
506 should be avoided since the *BIC* ratio at the end of FS insertion is not converged  
507 (see Fig. 8b), indicating that the maximum contact level at the sagittal faces and  
508 in particular in the calcar zone is not achieved.

509 While the studies on the ACI confirm the existence of an optimal interference

510 fit for maximizing primary stability (Immel et al. 2021), the existence of such op-  
511 timal configuration is less obvious for the FS. Due to the geometry of the FS, the  
512 effective value of the interference fit resulting from the cavity reaming is difficult  
513 to control, which leads to a huge variability on the cavity preparation between  
514 surgeons (Konow et al. 2022). Several studies on FS micromotion optimization  
515 found significantly lower optimal interference fit values than for the ACI (Abdul-  
516 Kadir et al. 2008; Pettersen et al. 2009; Russell et al. 2016). The results found  
517 herein indicate that an interference fit under 150  $\mu\text{m}$  significantly decreases the  
518 pull-out force without providing a significantly higher bone-implant contact ratio  
519 (see Fig. 10c). The behavior of the pull-out force with regard to the interfer-  
520 ence fit is in good agreement with a similar study considering both elastic and  
521 viscoelastic bone properties and a simplified cylindrical stem, where an interfer-  
522 ence fit “threshold” was evidenced beyond which no additional gains in push-out  
523 load are achieved (Shultz et al. 2006). However, other studies in the literature  
524 predict smaller optimal interference fit levels, around 50  $\mu\text{m}$ , to achieve good pri-  
525 mary stability fixation while avoiding femoral canal fracture (Abdul-Kadir et al.  
526 2008; Pettersen et al. 2009). In clinic, these levels of interference fit are difficult  
527 to reach with manual reaming and experimental results show that the interference  
528 fit corresponding to such micromotion obtained from FE analysis are more likely  
529 to be of 1-2  $\mu\text{m}$  (Abdul-Kadir et al. 2008).

#### 530 **4.4 Modal analysis: a tool to monitor FS insertion**

531 The resonance frequencies of the bone-stem-ancillary system and their evolution  
532 with the implant insertion into the bone are found to be in good agreement with

533 previous experimental results obtained with a bone mimicking phantom (Poudrel  
534 et al. 2022). In this previous study (Poudrel et al. 2022), two bending modes  
535 vibrating in a plane parallel to the sagittal one, namely the modes  $2Y$  and  $2Y_b$ ,  
536 with a mean frequency of 2875 Hz and 3496 Hz over the specimens, were found  
537 to be sensitive to the implant insertion depth into the bone cavity. Another mode  
538 shape named  $2X$  and oscillating parallel to the frontal plane was also measured.  
539 Its resonance frequency was around 3129 Hz and did not change significantly dur-  
540 ing implant insertion. The number of modes obtained numerically is found to be  
541 higher than experimentally (see Fig. 5), which can be explained by the experi-  
542 mental method employed to retrieve the mode shapes, based on the measure of  
543 accelerations all along the ancillary axis, which were recorded consecutively to an  
544 excitation by an hammer impact. Therefore, only the modes with the most impor-  
545 tant ancillary vibration amplitude were recorded in the previous study (Poudrel  
546 et al. 2022).

547 For the reference case ( $E_t^* = 0.2$  GPa,  $\mu^* = 0.3$ ,  $IF^* = 200$   $\mu\text{m}$ ), the numer-  
548 ical results corresponding to modes  $2Y$ ,  $2Y_b$  and  $2X$  (highlighted in bold in Fig.  
549 5) were identified based on their mode shape and resonance frequency, which are  
550 both coherent with experimental data, even if the frequencies are slightly higher.  
551 However, the boundary conditions, the bone material properties, the implant coat-  
552 ing and friction coefficient, and the effective experimental interference fit may be  
553 slightly different in the numerical and experimental studies, which may explain  
554 such difference. However, the variation of the resonance frequencies with the im-  
555 plant displacement (see Fig. 6b) is in good agreement with experimental data  
556 (Poudrel et al. 2022). In particular, the numerical model confirms the sensitivity  
557 of the modes  $2Y$  and  $2Y_b$  to the *BIC* ratio (see Fig. 6a), which could not have

558 been measured in the experimental study (Poudrel et al. 2022). Moreover, the  
559 resonance frequency of the mode  $2X$  is found not to increase for  $BIC > 55\%$ ,  
560 which is a value likely to be reached during the first steps of implant insertion and  
561 corroborates the non-sensitivity of the mode  $2X$  observed experimentally (Poudrel  
562 et al. 2022). Table 2 shows that the most sensitive mode for insertion monitoring  
563 is  $2Y_b$  with a variation of 14 Hz per % $BIC$  increase. Such a frequency increase dur-  
564 ing implant insertion could be easily measured experimentally since the frequency  
565 resolution of the experimental device developed in Poudrel et al. (2022) is 4 Hz.  
566 In addition, a lower increase of the eight resonance frequencies in the range [2-7]  
567 kHz is observed at the end of FS insertion than at the beginning, and in particular  
568 for  $U_I > 13$  mm (see Fig 6b), which is explained by the fact that the increase of  
569  $BIC$  ratio is lower at the end of the insertion (see Fig. 4). Experimentally, the  
570 modes  $2Y$  and  $2Y_b$  increased as a function of the insertion step, until they reached  
571 a threshold and stayed constant (Poudrel et al. 2022). The insertion step corre-  
572 sponding to the convergence was in good agreement with the convergence of the  
573 implant displacement and with the results obtained by another implant insertion  
574 monitoring method based on the impact force analysis (Poudrel et al. 2022). This  
575 result confirms that such a frequency convergence may be of interest to detect  
576 insertion end-point, which corresponds to the moment when continuing to push  
577 the implant into the bone will not lead to any better implant fixation or higher  
578  $BIC$  increase.

579 Different numerical studies considered the resonance frequency shift with con-  
580 tact conditions in order to simulate osseointegration and therefore investigate the  
581 mode sensitivity to the long-term stability (Pastrav et al. 2009a; Pérez and Seral-  
582 García 2013). Similarly as in our study, the resonance frequency shift with the

583 bone-implant contact was shown to depend on the mode number and on the loca-  
584 tion of the  $BIC$  : proximal, distal or central (Pastrav et al. 2009a). The highest  
585 frequency shifts were observed for proximal contact increase. In the perspective  
586 of the development of a surgical tool, being able to estimate the amount of bone-  
587 implant contact thanks to the measure of the bone-implant-ancillary system reso-  
588 nance frequencies is of great interest. In particular, it can prevent periprosthetic  
589 bone fractures by avoiding non-necessary hammer impacts if the  $BIC$  ratio has  
590 already converged (Fig. 4). The results shown in Fig. 6 highlight that two modes  
591 are particularly sensitive to the bone-implant contact ratio, namely the modes 7  
592 and 8, which were not identified in the previous experimental study (Poudrel et al.  
593 2022). An optimization of the sensor positioning on the ancillary, according to the  
594 mode shape, could be relevant to be able to measure these modes experimentally.

595 The sensitivity of the resonance frequency to  $E_t$  is found to depend on the  
596 mode of vibration (see Fig. 11). For the same level of  $BIC$ , while the frequencies  
597 of the modes  $2Y$  and  $2Y_b$  increase with the trabecular bone Young's modulus  
598 (Fig. 11a and 11c), the frequency of the mode  $2X$  is constant for high bone  
599 rigidity ( $E_t > 0.2$  GPa) (Fig. 11b). This variation of sensitivity depending on  
600 the mode number was also observed in a study focusing on the stem material  
601 properties (Pérez and Seral-García 2013), where the frequency shift observed for  
602 various implant materials varied according to the mode number. A global increase  
603 of the resonance frequencies is observed for stiffer materials considering the same  
604 level of bone-implant contact, which is in good agreement with our findings. This  
605 observation may be of interest for the surgeons in order to retrieve information on  
606 the patient bone quality or to adapt the surgical protocol. Moreover, as shown in  
607 Fig. 12 and 13, the resonance frequencies of mode  $2Y$ ,  $2X$  and  $2Y_b$  do not depend

608 on the friction coefficient and the interference fit, respectively, while they increase  
609 as a function of the *BIC* ratio. This behavior is explained by the sensitivity of  
610 the resonance frequencies to the bone-implant system stiffness, which depends  
611 on the quantity of bone-implant contact. Although the friction coefficient and  
612 the interference fit influences the implant insertion (and in particular the implant  
613 displacement and bone-implant contact, see Fig. 8), the values of  $IF$  and  $\mu$  do  
614 not affect the measurement of the resonance frequencies for a given *BIC* value,  
615 as the bone-implant interface was assumed to be fully bonded and linear for the  
616 modal analysis. As a conclusion, this analysis confirms that the vibration method  
617 studied herein, and in particular the monitoring of the resonance frequencies, may  
618 lead to an estimation of the *BIC* ratio, provided that bone quality is known, which  
619 is currently difficult to measure by the surgeon.

## 620 4.5 Limitations of the FE model

621 This study has several limitations. First, only a single type of FS and ancillary  
622 geometry was studied due to the numerous parameters considered and to computa-  
623 tion time related issues. However, the FS design considered in this work -straight  
624 stem with a cervico diphyseal angle of  $132^\circ$ , a medial curvature adapted to different  
625 morphologies to avoid post-operative varus ([uncemented femoral stem](#))- is largely  
626 employed by the surgeons for uncemented procedures. Therefore, the results ob-  
627 tained herein are of interest for a majority of uncemented procedures. However,  
628 the influence of other shapes of implant could be considered in a future work,  
629 based on studies about the effect of the implant geometry on the level of stress  
630 and micromotion under cyclic loading ([Dopico-González et al. 2010](#); [Reimeringer](#)

631 [et al. 2013](#); [Russell et al. 2016](#)) or on the bone remodeling ([Folgado et al. 2009](#)).

632 Several recent reports also provide insight on the contribution of FS design to the

633 risk of periprosthetic fractures ([Khanuja et al. 2011](#); [Carli et al. 2017](#)). A change

634 of femur geometry should also be considered in order to represent the patient vari-

635 ation ([Pettersen et al. 2009](#)). Eventually, although considering half of the femur

636 with the fixation of the distal end is commonly employed when studying femoral

637 stem insertion ([Taylor et al. 1995](#); [Pérez and Seral-García 2013](#); [Abdul-Kadir et al.](#)

638 [2008](#); [Tijou et al. 2018](#); [Poudrel et al. 2022](#)), such configuration is likely to affect

639 the values of the bone-implant-ancillary system resonance frequencies, compared

640 to a real anatomic configuration. Since the resonance frequencies depend on the

641 system rigidity, the length of the femur as well as the boundary condition at the

642 distal part may affect their quantitative values. However, the vibration behavior

643 with regard to the implant insertion should not be affected since these geometrical

644 and boundary conditions do not vary throughout the implant insertion procedure.

645 In addition, the same geometry of the femur as well as the same distal boundary

646 condition were used for the parametric study, which allows to compare the results

647 across the configurations. Note that in the context of acetabular cup insertion,

648 we showed that adding soft tissue around the bone receiving the implant does

649 not significantly modify the results obtained using the instrumented hammer ([Ti-](#)

650 [jou et al. 2018](#); [Bosc et al. 2018](#)). Eventually, the configuration of the numerical

651 model represents the experimental one of [Poudrel et al. \(2022\)](#) and the results ob-

652 tained using the two approaches could be compared. Although the results cannot

653 be directly translated to the clinical practice, the model may help to understand

654 the vibration behavior of the bone-implant-ancillary system during the implant

655 insertion procedure.

656 Second, we assumed that bone is homogeneous, elastic and isotropic. However,  
657 the bone properties are inhomogenous, as shown by several authors who used CT  
658 scans to recover the spatial distribution of the bone density (Ovesy et al. 2018).  
659 Even if the same computational procedure could have been followed, the aim was  
660 to determine the effect of the global bone stiffness on the FS insertion. The effect  
661 of bone anisotropic properties have been investigated for the ACI stability (Nguyen  
662 et al. 2017), but was not used in the present study herein in order to simplify the  
663 configuration for comparison of the effect of the parameters of interest.

664 Third, other studies have also pointed out the debonding effects (Immel et al.  
665 2020) or bone damage at the interface (Ovesy et al. 2020) and their influence  
666 on micromotion and pull-out force. However, even if trabecular bone damage  
667 may occur during the insertion process, an experimental study showed that bone  
668 damage has no impact on the pull-out force (Bishop et al. 2014). As our study  
669 accounts for primary stability, this effect was neglected.

670 Fourth, a uniform thickness of cortical bone was considered. As the implant is  
671 only in contact with trabecular bone, it is assumed that the homogeneous thickness  
672 does not influence the results. A similar study on the ACI confirms this assumption  
673 with results showing that the influence of the cortical bone stiffness on the pull-  
674 out force is small compared to the other parameters (trabecular bone Young's  
675 modulus, friction coefficient and interference fit) (Immel et al. 2021).

676 Fifth, quasi-static analyses were carried out whereas the surgeons insert the  
677 implant by successive hammer impacts. In order to take into account viscoelastic  
678 effects of the bone, it would be interesting to consider dynamic simulations with  
679 implant impaction (Monea et al. 2014; Michel et al. 2017). Note that the compu-  
680 tation cost required for such simulations would be considerably higher than the

681 one of the present study.

## 682 **5 Conclusion**

683 The present study brings new outcomes on the dependence of the vibration be-  
684 havior of the bone-implant-ancillary system on the implant environment, which  
685 allow to open new path to monitor implant seating in terms of implant displace-  
686 ment and bone-implant contact. To the best of our knowledge, this is the first  
687 numerical study proposing modal analysis of the bone-implant-ancillary system at  
688 different FS insertion steps and with different biomechanical environments. Based  
689 on the results of this study, the vibration method could be used in patients with  
690 different bone qualities and for different surgical procedures defined by various  
691 values of friction coefficient and interference fit. The sensitivity of the resonance  
692 frequencies to the bone-implant contact ratio and to the bone quality may be used  
693 to develop a quantitative method to monitor FS insertion by means of vibration  
694 measurement on the ancillary, which is easily accessible during the surgical pro-  
695 cedure. In addition, this study emphasizes the necessity to find a compromise on  
696 the interference fit and friction coefficient values, in order to maximize both the  
697 pull-out force and the *BIC* ratio. While the pull-out force is an increasing func-  
698 tion of the interference fit, the trabecular bone Young's modulus and the friction  
699 coefficient, the *BIC* ratio is a decreasing one. Both the stability and the vibration  
700 features, that is, the pull-out forces and the resonance frequencies, respectively,  
701 are consistent with previous experimental results. However, the proposed *in sil-*  
702 *ico* model should be improved in order to be more representative of the surgical  
703 impaction technique by simulating dynamic impacts.

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## 707 **7 Declarations**

### 708 **7.1 Ethic approval**

709 **Conflict of interest.** The authors declare that they have no financial or non-  
710 financial interests that are directly or indirectly related to the work submitted for  
711 publication.

### 712 **7.2 Authors' contribution**

713 G.H, V.-H.N and G.R. conceived the study and were in charge of overall direction  
714 and planning. A.-S.P, and V.-H.N. designed the model and the computational  
715 framework. A.-S.P. performed the calculations. All authors participated to the  
716 analysis of the data. A.-S.P. wrote the main manuscript text with inputs from all  
717 authors. All authors reviewed the manuscript.

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## 724 **7.4 Availability of data and materials**

725 The data that support the findings within this study are available from the corre-  
726 sponding author upon reasonable request.

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