Modal frequency and shape curvature as a measure of implant fixation: a computer study on the acetabular cup

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Abstract

Modal parameters are often investigated in order to assess the initial fixation of an implant. Most of studies are focused on the natural frequencies and frequency response function. Usually the femoral stem is tested although the acetabular cup fixation is important as well. The results of implant stability assessment are inconsistent and seem to suggest that frequency as a stability indicator is not sufficiently sensitive. In this study the sensitivity of the modal properties to changes in the bone-implant interface was investigated with the help of the finite element method (FEM). A novel fixation index based on modal shape curvature was investigated as a potential measure of the implant fixation. Modal frequencies are sensitive to interface changes in some manner, but suffer from insensitivity to local changes at bone-implant interface. The sensitivity up to 44 % of natural frequencies to stiffness change due insertion steps was observed. The tested damage indicators are able to detect localized small changes in peripheral stiffness (5 % stiffness reduction) with 95 % confidence under the noise up to 1%. The modal shapes and their curvatures have a great potential to be a robust fixation indicator.

Keywords: Acetabular Cup, Initial Fixation, Finite Element Method, Modal Shapes, Damage Index

1. Introduction

It is generally known that insufficient initial fixation of press-fitted acetabular implants leads to an excessive implant movement and consequent fibrous layer formation [43], which can cause

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4 aseptic loosening of the implant [15, 14]. On the other hand, excessive mechanical press-fit fix-
5 ation leads to an inadequate stress load in the peripheral bone. This may result in damage to the
6 surrounding bone and subsequent implant failure [7]. Achieving the optimal fixation level is a
7 fundamental factor for the long term outcome of the implant longevity. Initial implant fixation
8 assessment is often based on destructive mechanical tests. Implant fixation can be measured
9 by pull-out [37], lever-out [27] or torsional [10] tests, which are all unfortunately restricted to
10 be used in an in vitro setting. Alternatively, medical imaging techniques can provide qualita-
11 tive information about the implant state, but cannot quantify the amount of the initial fixation.
12 Other major drawbacks of imaging techniques are; the emission of harmful rays, cumbersome
13 to be used per-operatively and diffraction phenomena due to presence of metal materials [16].
14 Alternative concepts to assess implant fixation based on vibrational methods have been widely
15 investigated by many authors (changes in natural frequencies or changes in the magnitude of
16 frequency response functions (FRF) [6, 12, 21]). [An instrumented impact hammer for measuring](#)
17 [the implant stability has been developed and extensively tested. The impact hammer benefits](#)
18 [from time-domain feature from force cell mounted on the hammer while does not require the](#)
19 [response from bone itself \[34, 4, 26, 33, 31\].](#) Most of these studies are mainly concerned about
20 the hip stem, although acetabular component failure rates are similar or even slightly higher.
21 Moreover the qualitative assessment of the initial stability by auditory or other cues is more
22 difficult than for the stem [24, 22, 38]. The monitoring of initial implant fixation by tracking
23 frequency resonance shifts works well for dental implant [28, 29, 30], however the clinical im-
24 plementation in total hip replacement (THR) is still a challenge. Monitoring of initial fixation
25 has shown promising results on artificial and cadaveric bones and even on limited per operative
26 trials on patients [12, 42, 32, 33, 18]. Challenges are still abundant however; patient-specific or
27 operational factors may thwart the interpretation of changes in the FRFs. Additionally there is
28 still no consistent opinion whether the lower or higher frequency band is more sensitive to initial
29 fixation.

30 To investigate isolated parameters that influence the vibrational behaviour of the bone-implant
31 system, FEM models have been developed. Pastrav et al investigated the influence of contact
32 conditions on the resonance frequencies of a hip stem by FEM. They found that change at con-
33 tact area correlates with frequency shifts and more complex modes are more sensitive to changes
34 in contact area [40, 21]. The higher sensitivity of more complex modes has been supported by

35 Qi *et al* [44].

36 Modal analysis is the most widely used method to assess implant fixation, but only the natural
37 frequencies are commonly studied despite the fact that they are rather global and would not re-
38 flect adequately local changes in the bone-implant interface. Corresponding mode shapes are
39 more interesting because of their spatial dependence which potentially is more influenced by lo-
40 cal changes in the bone-implant interface. Moreover, the curvature of mode shape is often used in
41 industrial engineering to localize and quantify structural damage [8, 56, 39]. A curvature-based
42 damage index (DI) can be calculated based on the difference or ratio of two states, a healthy and
43 damaged state. The DI approximates the strain energy changes due to local changes in struc-
44 tural stiffness. Since the initial fixation of an implant is directly related to the interface stiffness,
45 characterizing the implant stability using a form of this DI could be of interest. Based on the
46 state-of-the art knowledge of the current methods dealing with monitoring of implant fixation,
47 this study aims to investigate computationally the sensitivity of the frequencies, mode shapes and
48 their curvatures to bone-acetabular implant interface changes. To the best of our knowledge, such
49 study has never been published and brings a fresh look to the usability of a vibration approach in
50 monitoring the fixation of implants

51 **2. Materials and Methods**

52 *2.1. FE Models*

53 The numerical model was composed of a validated finite element of an artificial pelvic bone.
54 The finite element model of the bone was validated with respect to its modal properties. The
55 material properties of the model were optimized based on the experimental modal analysis and
56 large scale gradient based optimization scheme with accuracy error below 1% [17]. The implant
57 model was based on a type SF-58 acetabular implant (Beznoska, Czech Republic). An isotropic,
58 homogeneous material model was assumed for the implant with a Young's modulus of 110 [GPa]
59 (Ti6Al4V), a Poisson's ratio of 0.3 [-] and a density of $4500 \text{ kg} \cdot \text{m}^{-3}$. The bone-implant interface
60 was modelled with simple interface conditions without friction or contact definitions. The bone
61 and implant were kinematically constrained in order to prevent relative movement between them.
62 The change at the interface was generated by different shared contact areas (SCA) at different
63 implantation depths. The depth of reamed cavity was varied from 4.2 to -2.3 mm and was defined
64 as a polar gap (Figure 1). The range was carefully set-up so that at the lowest value the implant

65 does not go through the medial wall while at the same time the implant only made contact with
 66 completely uncovered trabecular bone. The highest range 4.2 mm was selected so that the cortical
 67 contact area was more than 1000 mm^2 . No Dirichlet boundary conditions were assumed to ensure
 68 no influence of boundary conditions. The hemispherical cup was positioned at 45° abduction and
 25° anteversion [51]. The following simplifications are taken into account:

Figure 1: An illustrative cut of pelvic bone-acetabular implant interface with polar gap imposed. A positive/negative polar gap refers to the depth of reamed cavity that is lower/higher than the depth of acetabular notch.

69

- 70 • Simplified interface model was described by changing of contact area without relative
 71 displacement between the bone and implant.
- 72 • No friction or stiffening effects were considered in the model.
- 73 • No damping properties were taken into account.

74 2.2. Modal Analysis

75 A Krylov-Schur based method for solving the generalized eigenvalue problem was used from
 76 the library SLEPc [19]. Thirty-five eigenvalues and corresponding mode shapes were computed
 77 in a frequency range of 0–8kHz by a spectral shift-invert approach. The factorization solver used
 78 in the transformation was of a MUMPS type [2]. The full numerical scheme was implemented

in Python/C with parallel execution of assembling and solving the eigenvalue problem using the PETSc/SLEPc library [3]. The convergence of eigenvalues was verified on a parametrized mesh according to [17]. A mesh convergence analysis was carried out on the implant mesh. The maximum difference between eigenvalues was set to 5%. This threshold was satisfied by an average element size of 1 mm. Displacement field was approximated by a piecewise quadratic shape function. The mesh was created using the CGAL-3D library [53]. In order to compare two mode shapes quantitatively, the Modal Assurance Criterion (MAC) was used. The MAC is a standard similarity criterion commonly used in experimental and numerical modal analysis to match two mode shapes [1, 48]. The MAC is defined as:

$$MAC(\psi_1, \psi_2) = \frac{|\psi_1^T \psi_2|^2}{\psi_1^T \psi_1 \psi_2^T \psi_2} \quad (1)$$

Where $\cdot^T$ stands for the transpose and ψ_1, ψ_2 the real mass normalized mode shape vectors. To take into account the minor changes of the mesh for every insertion step, the mode shape vectors were interpolated on a template mesh with radial basis interpolation method [57]. The interpolation error was established by testing different mesh densities. The interpolation error was set to 10^{-6} and the MAC error to 1%. These criteria were satisfied with average element size 1 mm obtained from convergence analysis.

2.3. Damage Indices

With all mode shapes computed, a curvature based method commonly used in structural health monitoring for damage identification was applied in order to quantify the changes of mode shapes for points at the bone-implant peripheral interface. The curvature was measured on the peripheral rim of acetabular implant at finite element nodes (Figures 1 and 2). The curvature κ of the i^{th} mode shape ψ_i defined as second order derivation of ψ with respect to spatial coordinates:

$$\kappa_i = \frac{\partial^2 \psi_i}{\partial \mathbf{x}^2}, \mathbf{x} = \{x, y, z\} \quad (2)$$

was computed with second order forward differentiation. Three Damage Indices (DI-A, DI-B and DI-C) were used. The damage indices were sensitive to stiffness changes commonly occurring as a result from damage to a structure. Due to the reduced stiffness of a damaged object, the local curvature of the mode shape increases. The damage indices compare quantities derived from the mode shapes of the bone-implant construct between different states of the bone-implant

construct. The undamaged state of the bone-implant construct was defined as configuration with the implant at an insertion level with depth -2.3 mm. This state represents the ideal fixation level of the implant. The damaged states of the bone-implant constructs were defined as the rest of the insertion levels. The *damaged* and *undamaged* states are indicated by superscripts $\cdot^d$, $\cdot^u$ respectively. The curvatures κ are expressed as a function of peripheral angle ϕ defined in Figure 2. The variable N introduces the number of modal shapes used to compute the DI and has value 10 unless otherwise specified.

The first damage indicator (DI-A) [5] was defined as the difference between the mode shape curvatures κ of a *damaged* and *undamaged* state of the bone-implant construct:

$$\delta_A(\phi) = \frac{1}{N} \sum_{i=1}^N \left\| \left[1 + \frac{\Delta \kappa_{\phi,i}}{\max(\Delta \kappa_i) - \min(\Delta \kappa_i)} \right]^2 \right\| \quad (3)$$

with curvature difference $\Delta \kappa_{\phi,i} = \kappa_{\phi,i}^d - \kappa_{\phi,i}^u$. The DI-A was normalized with respect to the max/min values of the mode shape curvature difference. The second damage index DI-B [13] was based on the fractional relation of the modal energy U between the *damaged* and *undamaged* state of the bone-implant construct.

$$\delta_B(\phi) = \frac{\sum_{i=1}^N \frac{U_{i,\phi}^d}{U_i^d}}{\sum_{i=1}^N \frac{U_{i,\phi}^u}{U_i^u}} \quad (4)$$

where $U_{i,\phi}$ is proportional to the modal strain energy of i^{th} mode shape [52]:

$$U_{i,\phi} \propto INTER^{\phi \leftarrow bc^e} \int_{\Omega_e} [\kappa_x^2 + \kappa_y^2 + \kappa_z^2] d\Omega_e \quad (5)$$

where $\kappa_x, \kappa_y, \kappa_z$ express the partial curvatures with respect to coordinates. The modal energy of i^{th} mode shape is computed over elements on the peripheral rim of the implant. In order to compare it with nodal valued indexes DI-A, DI-C, elemental modal strain energy $U_{e,i}$ was interpolated to a nodal space at barycentric coordinates (bc^e) of e^{th} element. The total modal strain energy U_i of i^{th} modal shape is a sum over the elements. The third damage index DI-C was based on the norm of the difference of the mode shape curvature κ corresponding to a *damaged* and *undamaged* state:

$$\delta_C(\phi) = \frac{1}{N} \sum_{i=1}^N \left\| (\kappa_{i,\phi}^d)^2 - (\kappa_{i,\phi}^u)^2 \right\| \quad (6)$$

where the $\kappa_{i,\phi}$ is the mode shape curvature of i^{th} shape at peripheral angle ϕ . For every DI the average over the modal shapes N was calculated. Subsequently, the DI for every mode was

129 normalized as follows:

$$\hat{\delta}(\phi) = \frac{\delta(\phi) - \mu_{\delta}}{\sigma_{\delta}} \quad (7)$$

130 where μ and σ are the mean over angle ϕ and standard deviation respectively. For values $\hat{\delta}(\phi) > 2$
 131 the presence of the damage at measured angle ϕ was presumed. This threshold value corresponds
 132 to a probability at the 97.725% confidence level (assuming a normal distribution) that damage
 133 was located at the identified location [9, 54, 13, 45, 46, 50].

134 2.4. Computational Experiment with the Damage Indices

135 In order to investigate the potential of the defined DIs to distinguish between *damaged* and
 136 *undamaged* states, firstly an additional computational experiment was performed before testing
 137 the DIs on the pelvic bone model. A simplified FEM model was created consisting only of the
 138 acetabular cup itself. In order to simulate the localized damage, the so called 'damage segment' is
 139 introduced in Figure 2. The damage segment contains the reduced Young's modulus to simulate
 140 the damage.

141 The parametrized hemispherical cup was created using the CGAL library and was meshed
 142 using linear tetrahedral elements with an average element size of 1 mm (the thickness of cup was
 143 4 mm). The displacement of the cup was approximated by quadratic basis functions.

144 2.4.1. Sensitivity of DI to Change of Stiffness

145 The stiffness of the cup was then changed locally at several angles ϕ by modifying the modu-
 146 lus of elasticity at these angles. Angle ϕ and Youngs modulus E were parametrized in a following
 147 way to assess the DI performance:

- 148 • Parameter E : steps 5, 25 and 50% at the indicated locations,
- 149 • Parameter ϕ : varying from 0 to 300 ° with step 60 ° (6 location in total)

150 The segment width d was 5°. A total of 18 simulations were performed.

151 2.4.2. Sensitivity of DI to a Noise

152 In order to assess the DIs performance with noisy data, the modal shapes were contaminated
 153 by noise controlled by variable σ according to:

$$\hat{\psi} = \psi \left[1 + \alpha \left(0, \frac{\sigma}{100} \right) \right] \quad (8)$$

154 where ψ is a modal shapes computed from FE simulation and α is a random uniform variable
 155 spanning from 0 to σ percent of the average deformation of the modal shape. The level of σ var-
 156 ied from 0 to 5% with steps of 0.5%. A Monte Carlo method was used to estimate the probability
 157 of correct identification of damage at all simulation configurations presented in section 2.4.1. A
 158 95% confidence level was chosen and the maximal noise level that can exist with respect to
 159 this probability was computed. 10,000 occurrences were sampled from the uniform distribution
 during the Monte Carlo simulation. A total of 1,800,000 simulations were performed.

Figure 2: Parametrized acetabular cup with damage segment (green color), which has the position measured by angle ϕ [°] with respect to global coordinate system O and arc-length d [°].

160

3. Results

3.1. Simplified Acetabular Model

Figure 3 shows the ten first flexible mode shapes of the acetabular cup. Due to symmetry, the first six modes are coupled. Modes 1-6 and 8-10 are predominantly a combination of bending and bellowing deformational behaviour, mode 7 is characterized by torsional deformation. Higher modes typically showed more complex behaviour, with a combination of bending, bellowing, longitudinal and torsional deformation. The first ten natural frequencies of the cup

Figure 3: Modal shapes of acetabular cup

corresponding to these mode shapes are shown in Table 1. The first natural frequencies and corresponding modal shape are similar to the results reported by [20]. The frequency range of the first natural frequencies of the cup was considerably higher than the bone itself or of the bone-implant assembly. Figure 4 shows the values obtained for the different DIs in function of the parametrized angle and Youngs modulus. With a reduction of 5% of the Youngs modulus, only the DI-A identified the correct angles where the Youngs modulus was reduced. When the reduction of the Youngs modulus was higher, the amplitudes of DI generally increased except for DI-B, where the results do not show a general trend. Table 2 shows the noise sensitivity of every DI with damage at an angle of 0° and with all ten mode shapes included. The results for other damage locations were similar. The results listed in the table clearly indicated that DI-A performed well and could detect the smallest change in Youngs modulus with noise around 1%

Table 1: Natural frequencies of acetabular cup without Young's modulus reduction

mode #	freq. [kHz]	mode #	freq. [kHz]
1	4.9	6	21.8
2	4.9	7	26.5
3	12.4	8	27.1
4	12.4	9	27.1
5	21.8	10	31.8

Figure 4: A typical evaluation of DI magnitude for three reduction levels of Young's modulus (5, 25 and 50 %) and for six locations of damage segment(black triangles: $\phi=\{0, 60, 120, 180, 240, 300\}[\circ]$). The models A, B, C refer to a different DI formulation.

while the highest change in Young's modulus could be detected with 5% noise at the 95% confidence level. The DI-B can detect only changes with noise at the 1% level and fails to detect damage with noise at the 5% level.

3.2. Pelvic Bone-implant Model

The changes at the interface are described by the evaluation of the SCA at different cavity depths. The evolution of the SCA is presented in Figure 5. The evolution of the contact area is split into two bone regions. When inserted in the cavity with the lowest polar gap, the implant is mainly in contact with the trabecular bone. From the first stages of insertion, corresponding to a higher polar gap, to the final stages of insertion, the cortical area increases until it reaches the maximal value (1.98 mm), after the maximum of cortical area is reached, the cortical area decreases while the trabecular contact increases. The difference of the natural frequencies between

Table 2: Sensitivity of DIs to a noise (a success detection of damage > 95%)

Damage [%]	maximum noise [%]		
	DI-A	DI-B	DI-C
5	1	-	0.5
25	3	0.5	1
50	5	0.5	3

the first and final insertion step is shown in Table 3. The Table contains the maximal differ-

Figure 5: Evaluation of SCA during different insertion levels

190

191 ences only for the most sensitive natural frequencies. All differences exceeded 35%. According
192 to Figure 6, the relative sensitivity of frequencies decreased with higher modes. All first twenty
193 flexible modes however have sensitivities higher than 20%. The shapes associated with the five
194 most sensitive frequencies are shown in Figure 7 from high to low sensitivity. The first mode
195 has mainly a torsional character; however the rest of the modes significantly deforms around the
196 acetabular rim and behaves as a combination of bending and torsional modes. The sensitiv-
197 ity of the shapes is shown in Figure 8. The first and second chart present the global change of
198 minimum or maximum of MAC values due to changes of the SCA. When the modal complexity
199 increased, the difference in minimal MAC values between successive insertion steps increased
200 and the maximal MAC values decreased. The last chart shows the MAC evolution between each

Figure 6: Frequency sensitivity

Table 3: Selected sensitivity of natural frequencies

mode #	9	10	3	8	1
rel. difference [%]	43.3	42.1	41.9	37.7	37.1

201 insertion step for the mode shapes associated with the most sensitive frequencies. Although the
 202 most sensitive frequencies varied more than 35%, only shapes 8, 9 and 10 varied significantly
 203 (with a minimal MAC of 0.731 , 0.712 and 0.662 respectively).

Figure 7: Associated shapes with the most sensitive frequencies, top line: -2.3 mm, bottom-line: 4.2 mm. Grey and blue colours represent the undeformed and deformed geometry respectively.

Figure 8: MAC evaluation over the range of modes

204 In Figure A.10 (appendix) the DI (in polar coordinates) is shown with basic anatomical mark-
 205 ers on the pelvic bone indicated for clarity. Different colours correspond to different SCA levels.
 206 Figure A.10 represents an evaluation of the DI for the sum of the first ten modal shapes. The
 207 changes in stiffness due to changes in the SCA as the implant is inserted into the cavity are not as
 208 clear-cut as the local changes in stiffness simulated in section 3.1. The DIs based on the sum of
 209 the first ten shapes contribution (Figure A.10 (appendix): A, B and C) show some peaks around
 210 the whole peripheral rim. Rather than evaluation of individual peaks, the whole shape of the
 211 closed curve must be evaluated. With a closer look, the DI does not always increase monotonic
 212 across the insertion levels. For example, the DI-A at 270-300 degrees significantly increases
 213 up to insertion level 2.18 mm and then starts decreasing up to the end. In order to get clearer
 214 information about implant fixation, the averaged value of DI over the whole curve is shown in 9.
 The curves show that increasing the polar gap generally increases the damage indices too.

Figure 9: Average value of non-normalized DI over whole peripheral rim.

215

216 4. Discussion

217 The presented study investigated the dynamic behaviour of the bone-implant system for dif-
218 ferent contact conditions between bone and acetabular implant. These contact conditions cor-
219 relate well to the initial stability of the implant. The study of interface evaluation revealed that
220 during the insertion phase, different amounts of cortical and trabecular bone came into contact
221 with the implant. The evolution of the SCA indicated that the ratio of cortical and trabecular
222 bone in contact with the implant evolved non-linearly. In addition possible large variation due to
223 other factors (cup position and orientation, acetabular shape, bone quality-mechanical properties)
224 may additionally influence this distribution. Evaluating the dependency of modal parameters on
225 the SCA, significant sensitivity of the resonance frequencies was found. The natural frequen-
226 cies changes ranged from 0 to 44%. This seems to point to the frequencies to be a promising
227 predictor of the implant stability; however some important remarks must be made. To the best
228 of the authors knowledge, there is no experimental study that indicated such high sensitivity.
229 Typically, the frequency shift of some modes is around 10-20%, albeit with no guarantee that
230 the modes were properly tracked. Tracking resonance frequencies by correlating mode shapes
231 between successive steps is important to understand possible mode switching and veering of the
232 system [25]. Tracking correlated mode shapes as the structure changes has not been performed
233 in biomechanical studies, although it can completely distort the obtained frequency shifts. In this
234 work, the modal frequencies were carefully tracked by comparing the MAC criteria of each mode
235 in order to exclude possible mismatches. Mode switching and veering can occur if the changes
236 in structure are too large. Moreover, the modal density is quite high in the selected frequency
237 range and the modes could easily switch or veer. Roughly speaking, to use shifts in natural fre-
238 quencies in order to assess the implant fixation, experimental modal analysis with more than one
239 measuring point is likely advised for systems with a high modal density. The first two natural
240 frequencies of the acetabular cup/bone system are in range 4–5kHz, which is in accordance with
241 numerical study at the time domain [35]. Although the natural frequencies of the pelvic bone
242 itself are lower ([17, 36]), due to significantly higher stiffness of the cup implant and its smaller
243 dimension than pelvic bone, the bone-implant assembly (with fully evolved contact interface)
244 becomes stiffer and thus natural frequencies shift to higher frequency band.

245 The mode shapes in this study were assessed as potential predictor of implant fixation. How-
246 ever, the shapes need to be measured carefully, especially as the more complex modes are hard

247 to acquire and would indeed benefit from sophisticated measuring equipment such as a laser
248 vibrometry scanner. The sensitivity of thirty-five modal shapes was investigated in this study.
249 It was shown that the accuracy of more complex mode shapes decreased. From a practical
250 point of view, only the first ten modes can likely be taken as a predictor. The higher modes are
251 more sensitive to noise and small, rather insignificant, changes in the model as illustrated by the
252 MAC evaluation of the mode shapes (Figure 8). The higher spread of MAC values with higher
253 modes was due to the computational shape accuracy and sensitivity to model parameters. From
254 a practical point of view, measurement of the mode shapes is important in order to properly pair
255 and compare shifts in natural frequencies between successive insertion steps. Once these mode
256 shapes are available, implant fixation can be evaluated in terms of changes in the peripheral stiff-
257 ness which can be deduced from changes in these mode shapes. In this study some alternative
258 indicators to determine initial fixation were introduced. Changes in interface stiffness were linked
259 to changes in mode shape curvatures. It is generally known that higher curvature values reflect
260 lower local stiffness. Three different indicators based on the curvature were numerically tested
261 using an artificial acetabular cup to assess the indicators basic performance. Results showed that
262 the most promising indicator is the ratio between two elastic energies (expressed as the square
263 of the mode shape curvature) comparing the actual and undamaged reference states (DI-A, the
264 reference state: -2.3 mm polar gap). A second indicator (DI-B) demonstrated the least interesting
265 performance and was deemed not suitable to be used as a stability predictor. The last measure
266 DI-C was able to detect changes in stiffness with a lower accuracy. The accuracy of the different
267 DIs was tested on mode shapes contaminated by different levels of uniform noise. The minimal
268 detection level was achieved only for the DI-A and DI-C with stiffness changes of at least 5%
269 with maximal noise of 1% and 0.5% respectively. The performance of the DI is sensitive to the
270 accurate computing of the second order derivation (or the first order derivative of the strain) of
271 the mode shapes. Since the computing of second order derivative of shapes are susceptible to
272 data noise, computing by wavelet or another approximation method would be an important step
273 to clinically implement the proposed DIs [11, 49].

274 Numerical experiments on simplified cup showed a nearly perfect detection ability of the
275 DIs, although frequency changes were limited to only 5%. In that case the interpretation of DI is
276 clear. Interpretation of the DI results for the bone-implant system was more difficult. Although
277 it could detect changes in the peripheral acetabular stiffness, it was not possible to determine the

278 state of initial fixation based on only one peak in the polar DI graph but rather needed to evaluate
279 the shape. Additionally the behaviour of the DIs was not monotonic which indicated that the
280 stiffness was changing in different ways during the insertion steps. Future work will focus on
281 determining how the resulting stiffness evolves during insertion using a virtual torque testing and
282 will correlate these results with the DIs obtained. In this work the insertion level with polar gap
283 -2.3 mm was taken as the undamaged reference state (the most stable implant configuration).
284 In a clinical situation, the evolution of implant fixation should be measured between steps and
285 the initial and final step could be compared. This study has several limitations. A first limita-
286 tion is the description of the bone-implant interface. The bone and implant are kinematically
287 constrained. These constraints are more representative for the fixation the implant would obtain
288 after the osseointegration process. As a numerical modal analysis expects linear behaviour of the
289 model, implementation of non-linear contact conditions would inhibit computing the modal pa-
290 rameters in an efficient way. Nevertheless, the influence of non-linear interface conditions could
291 be explored for a limited set of cases by included them in the FE model by local approximation
292 techniques (harmonic balance method [55, 47]). The resultant frequency response would likely
293 be strongly non-linear close to a nearly zero initial fixation of the implant. This method will be
294 implemented into the model in future studies. It must be pointed out that the non-linear interface
295 behaviour is usually omitted because the behaviour of the bone-implant model during the inser-
296 tion is more or less linear when measured between insertion steps except when the implant would
297 be in clinically unstable state due to insufficient fixation during insertion or bone fractures that
298 may occur during insertion [48, 6, 21, 23, 41]. A second limitation or challenge of the proposed
299 fixation monitoring approach is to develop a clinical device that would be able to measure the
300 mode shapes of acetabular cup experimentally during surgery. From a technical point of view,
301 the experimental device should be able to measure the mode shapes by applying a controlled
302 excitation to the acetabular cup and measure its response at a certain number of locations, prefer-
303 ably on the peripheral rim of the cup. Possible measurement solutions could include the use of
304 some compact laser vibrometry or a high speed camera which provide a field of harmonic dis-
305 placements. Repeated measurements for every insertion step will likely be necessary to average
306 out the noise in the mode shape extraction process (identification and computation of the shape
307 curvature). [Another limitation of the computational model can be seen in the material model](#)
308 [used. The bone visco-elasticity affects the implant in two folds: it adds a viscous damping into](#)

309 modal dynamics and thus complicates the evaluation of the natural frequencies and shapes; due
310 to stress relaxation of bone, the initial implant stability might vary in the time.

311 From a clinical point of view, the measuring of the initial fixation during the implantation
312 step can help to objectively achieve an optimal press-fit fixation without exceeding the biome-
313 chanical limits of the bone and implant. A vibration based approach, such as the one presented
314 in this study, furthermore has the advantage that the low level of vibrations needed to perform
315 the measurement will not damage or change the interface while measuring.

316 **5. Conclusion**

317 In this study the authors presented a new method to measure implant fixation by modal shape
318 based damage indicator and assessed its potential by computer experiments on a simplified ac-
319 etabular cup model and on a validated finite element model. The next step would be the exper-
320 imental application of the proposed method on composite and cadaver bones. Simultaneously,
321 advanced computational model of bone-implant construct with complex interface properties will
322 be developed in order to gain deeper knowledge how to describe the implant fixation by modal
323 properties.

324 **Competing Interests**

325 The authors have no conflict of interest in relation to the present study.

326 **Funding**

327 None

328 **Ethical Approval**

329 Not required

Figure A.10: A typical evaluation of damage index during insertion steps. Coloured curves express the value of DI at actual insertion step and at position on implant rim. The curves contains a radial offset of value 10 in order to get figure readable. The polar graph axes correspond to acetabular rim and its centre.

References

- [1] Allemang, R. J., 2003. The modal assurance criterion—twenty years of use and abuse. *Sound and vibration* 37 (8), 14–23.
- [2] Amestoy, P. R., Duff, I. S., Koster, J., L'Excellent, J.-Y., 2001. A fully asynchronous multifrontal solver using distributed dynamic scheduling. *SIAM Journal on Matrix Analysis and Applications* 23 (1), 15–41.
- [3] Balay, S., Abhyankar, S., Adams, M. F., Brown, J., Brune, P., Buschelman, K., Dalcin, L., Eijkhout, V., Gropp, W. D., Kaushik, D., Knepley, M. G., McInnes, L. C., Rupp, K., Smith, B. F., Zampini, S., Zhang, H., 2015. PETSc Web page.
- [4] Bosc, R., Tijou, A., Rosi, G., Nguyen, V.-H., Meningaud, J.-P., Hernigou, P., Flouzat-Lachaniette, C.-H., Haiat, G., 2018. Influence of soft tissue in the assessment of the primary fixation of acetabular cup implants using impact analyses. *Clinical Biomechanics* 55, 7–13.
- [5] Chandrashekar, M., Ganguli, R., 2009. Damage assessment of structures with uncertainty by using mode-shape curvatures and fuzzy logic. *Journal of Sound and Vibration* 326 (3), 939–957.
- [6] Cristofolini, L., Varini, E., Pelgreffi, I., Cappello, A., Toni, A., 2006. Device to measure intra-operatively the primary stability of cementless hip stems. *Medical engineering & physics* 28 (5), 475–482.
- [7] Curtis, M., Jinnah, R., Wilson, V., Hungerford, D., 1992. The initial stability of uncemented acetabular components. *Journal of Bone & Joint Surgery, British Volume* 74 (3), 372–376.
- [8] Dawari, V., Vismawala, G., 2013. Structural damage identification using modal curvature differences. *Journal of Mechanical and Civil Engineering*, 2278–1684.
- [9] Fan, W., Qiao, P., 2009. A 2-d continuous wavelet transform of mode shape data for damage detection of plate structures. *International Journal of Solids and Structures* 46 (25), 4379–4395.
- [10] Fehring, K. A., Owen, J. R., Kurdin, A. A., Wayne, J. S., Jiranek, W. A., 2014. Initial stability of press-fit acetabular components under rotational forces. *The Journal of arthroplasty* 29 (5), 1038–1042.
- [11] Fu, C.-L., Feng, X.-L., Qian, Z., 2010. Wavelets and high order numerical differentiation. *Applied Mathematical Modelling* 34 (10), 3008–3021.
- [12] Georgiou, A., Cunningham, J., 2001. Accurate diagnosis of hip prosthesis loosening using a vibrational technique. *Clinical Biomechanics* 16 (4), 315–323.
- [13] Grooteman, F., 2012. Damage detection and probability of detection for a shm system based on optical fibres applied to a stiffened composite panel. *ISMA2012, Leuven*, 17–19.
- [14] Hamilton, W. G., Calendine, C. L., Beykirch, S. E., Hopper, R. H., Engh, C. A., 2007. Acetabular fixation options first-generation modular cup curtain calls and caveats. *The Journal of arthroplasty* 22 (4), 75–81.
- [15] Healy, W. L., Casey, D. J., Iorio, R., Appleby, D., 2002. Evaluation of the porous-coated anatomic hip at 12 years. *The Journal of arthroplasty* 17 (7), 856–863.
- [16] Hecht, S., Adams, W. H., Narak, J., Thomas, W. B., 2011. Magnetic resonance imaging susceptibility artifacts due to metallic foreign bodies. *Veterinary Radiology & Ultrasound* 52 (4), 409–414.
- [17] Henyš, P., Čapek, L., Feb 2017. Material model of pelvic bone based on modal analysis: a study on the composite bone. *Biomechanics and Modeling in Mechanobiology* 16 (1), 363–373.
- [18] Henyš, P., Čapek, L., Apr 2018. Impact force, polar gap and modal parameters predict acetabular cup fixation: A study on a composite bone. *Annals of Biomedical Engineering* 46 (4), 590–604.

- [19] Hernandez, V., Roman, J., Tomas, A., Vidal, V., 2007. Krylov-schur methods in slepc. Universitat Politecnica de Valencia, Tech. Rep. STR-7.
- [20] Hothan, A., Huber, G., Weiss, C., Hoffmann, N., Morlock, M., 2011. Deformation characteristics and eigenfrequencies of press-fit acetabular cups. *Clinical Biomechanics* 26 (1), 46–51.
- [21] Jaecques, S., Pastrav, C., Zahariuc, A., Van der Perre, G., 2004. Analysis of the fixation quality of cementless hip prostheses using a vibrational technique. In: *Proceedings of ISMA 2004 International Conference on Noise and vibration engineering: 20–22 September 2004; Leuven*. pp. 443–456.
- [22] Köster, G., Munz, D., Köhler, H.-P., 1993. Clinical value of combined contrast and radionuclide arthrography in suspected loosening of hip prostheses. *Archives of orthopaedic and trauma surgery* 112 (5), 247–254.
- [23] Lannocca, M., Varini, E., Cappello, A., Cristofolini, L., Bialoblocka, E., 2007. Intra-operative evaluation of cementless hip implant stability: A prototype device based on vibration analysis. *Medical engineering & physics* 29 (8), 886–894.
- [24] Lieberman, J., Huo, M., Schneider, R., Salvati, E., Rodi, S., 1993. Evaluation of painful hip arthroplasties. are technetium bone scans necessary? *The Journal of bone and joint surgery. British volume* 75 (3), 475–478.
- [25] Mace, B. R., Manconi, E., 2011. Dispersion phenomena in coupled waveguides: veering, locking and strong coupling effects.
- [26] Mathieu, V., Michel, A., Lachaniette, C.-H. F., Poignard, A., Hernigou, P., Allain, J., Haiat, G., 2013. Variation of the impact duration during the in vitro insertion of acetabular cup implants. *Medical Engineering & Physics* 35 (11), 1558–1563.
- [27] Meneghini, R. M., Meyer, C., Buckley, C. A., Hanssen, A. D., Lewallen, D. G., 2010. Mechanical stability of novel highly porous metal acetabular components in revision total hip arthroplasty. *The Journal of arthroplasty* 25 (3), 337–341.
- [28] Meredith, N., Alleyne, D., Cawley, P., 1996. Quantitative determination of the stability of the implant-tissue interface using resonance frequency analysis. *Clinical oral implants research* 7 (3), 261–267.
- [29] Meredith, N., Books, K., Friberg, B., Jemt, T., Sennerby, L., 1997. Resonance frequency measurements of implant stability in vivo. a cross-sectional and longitudinal study of resonance frequency measurements on implants in the edentulous and partially dentate maxilla. *clinical oral implants research* 8 (3), 226–233.
- [30] Meredith, N., Shagaldi, F., Alleyne, D., Sennerby, L., Cawley, P., 1997. The application of resonance frequency measurements to study the stability of titanium implants during healing in the rabbit tibia. *Clinical oral implants research* 8 (3), 234–243.
- [31] Michel, A., Bosc, R., Mathieu, V., Hernigou, P., Haiat, G., 2014. Monitoring the press-fit insertion of an acetabular cup by impact measurements: Influence of bone abrasion. *Proceedings of the Institution of Mechanical Engineers, Part H: Journal of Engineering in Medicine* 228 (10), 1027–1034.
- [32] Michel, A., Bosc, R., Meningaud, J.-P., Hernigou, P., Haiat, G., 11 2016. Assessing the acetabular cup implant primary stability by impact analyses: A cadaveric study. *PLOS ONE* 11 (11), 1–14.
- [33] Michel, A., Bosc, R., Sailhan, F., Vayron, R., Haiat, G., 2016. Ex vivo estimation of cementless acetabular cup stability using an impact hammer. *Medical Engineering & Physics* 38 (2), 80–86.
- [34] Michel, A., Bosc, R., Vayron, R., Haiat, G., 2015. In vitro evaluation of the acetabular cup primary stability by impact analysis. *Journal of Biomechanical Engineering* 137 (3), 031011.

- [35] Michel, A., Nguyen, V.-H., Bosc, R., Vayron, R., Hernigou, P., Naili, S., Haiat, G., May 2017. Finite element model of the impaction of a press-fitted acetabular cup. *Medical & Biological Engineering & Computing* 55 (5), 781–791.
- [36] Neugebauer, R., Werner, M., Voigt, C., Steinke, H., Scholz, R., Scherer, S., Quickert, M., 2011. Experimental modal analysis on fresh-frozen human hemipelvic bones employing a 3d laser vibrometer for the purpose of modal parameter identification. *Journal of Biomechanics* 44 (8), 1610–1613.
- [37] Olory, B., Havet, E., Gabrion, A., Vernois, J., Mertl, P., 2004. Comparative in vitro assessment of the primary stability of cementless press-fit acetabular cups. *Acta Orthopædica Belgica* 70 (1), 31–37.
- [38] Ovesen, O., Riegels-Nielsen, P., Lindequist, S., Jensen, I., Munkner, T., Torfing, T., Marving, J., 2003. The diagnostic value of digital subtraction arthrography and radionuclide bone scan in revision hip arthroplasty. *The Journal of arthroplasty* 18 (6), 735–740.
- [39] Pandey, A., Biswas, M., Samman, M., 1991. Damage detection from changes in curvature mode shapes. *Journal of sound and vibration* 145 (2), 321–332.
- [40] Pastrav, L., Devos, J., Van der Perre, G., Jaecques, S., 2009. A finite element analysis of the vibrational behaviour of the intra-operatively manufactured prosthesis–femur system. *Medical engineering & physics* 31 (4), 489–494.
- [41] Pastrav, L., Jaecques, S., Mulier, M., Der Perre, G. V., 2008. Detection of the insertion end point of cementless hip prostheses using the comparison between successive frequency response functions. *Journal of Applied Biomaterials & Biomechanics* 6 (1).
- [42] Pastrav, L. C., Jaecques, S. V., Jonkers, I., Van der Perre, G., Mulier, M., 2009. In vivo evaluation of a vibration analysis technique for the per-operative monitoring of the fixation of hip prostheses. *Journal of orthopaedic surgery and research* 4 (1), 10.
- [43] Pilliar, R., Lee, J., Maniopoulos, C., 1986. Observations on the effect of movement on bone ingrowth into porous-surfaced implants. *Clinical orthopaedics and related research* 208, 108–113.
- [44] Qi, G., Mouchon, W. P., Tan, T. E., 2003. How much can a vibrational diagnostic tool reveal in total hip arthroplasty loosening? *Clinical Biomechanics* 18 (5), 444–458.
- [45] Rucevskis, S., Janeliukstis, R., Akishin, P., Chate, A., 2016. Mode shape-based damage detection in plate structure without baseline data. *Structural Control and Health Monitoring*.
- [46] Rucevskis, S., Sumbatyan, M. A., Akishin, P., Chate, A., 2015. Tikhonov’s regularization approach in mode shape curvature analysis applied to damage detection. *Mechanics Research Communications* 65, 9–16.
- [47] Sanliturk, K., Ewins, D., 1996. Modelling two-dimensional friction contact and its application using harmonic balance method. *Journal of sound and vibration* 193 (2), 511–523.
- [48] Scholz, R., Hoffmann, F., von Sachsen, S., Drossel, W.-G., Klöhn, C., Voigt, C., 2013. Validation of density–elasticity relationships for finite element modeling of human pelvic bone by modal analysis. *Journal of biomechanics* 46 (15), 2667–2673.
- [49] Shao, X., Ma, C., 2003. A general approach to derivative calculation using wavelet transform. *Chemometrics and Intelligent Laboratory Systems* 69 (1-2), 157–165.
- [50] Siebel, T., Friedman, A., Koch, M., Mayer, D., 2012. Assessment of mode shape-based damage detection methods under real operational conditions. In: *6th European Workshop on Structural Health Monitoring*. Dresden, Germany. pp. 1–8.

- 448 [51] Spears, I. R., Pfeleiderer, M., Schneider, E., Hille, E., Bergmann, G., Morlock, M. M., 2000. Interfacial conditions
449 between a press-fit acetabular cup and bone during daily activities: implications for achieving bone in-growth.
450 Journal of biomechanics 33 (11), 1471–1477.
- 451 [52] Stubbs, N., Kim, J., Topole, K., 1992. An efficient and robust algorithm for damage localization in offshore plat-
452 forms. In: Proceedings of the ASCE tenth structures congress. pp. 543–546.
- 453 [53] The CGAL Project, 2015. CGAL User and Reference Manual, 4.7 Edition. CGAL Editorial Board.
454 URL <http://doc.cgal.org/4.7/Manual/packages.html>
- 455 [54] Tomaszewska, A., 2010. Influence of statistical errors on damage detection based on structural flexibility and mode
456 shape curvature. Computers & structures 88 (3), 154–164.
- 457 [55] Von Groll, G., Ewins, D. J., 2001. The harmonic balance method with arc-length continuation in rotor/stator contact
458 problems. Journal of sound and vibration 241 (2), 223–233.
- 459 [56] Wahab, M. A., De Roeck, G., 1999. Damage detection in bridges using modal curvatures: application to a real
460 damage scenario. Journal of Sound and Vibration 226 (2), 217–235.
- 461 [57] Walt, S. v. d., Colbert, S. C., Varoquaux, G., 2011. The numpy array: A structure for efficient numerical computa-
462 tion. Computing in Science & Engineering 13 (2), 22–30.