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Characterization of bone-implant fixation using modal analysis: Application to a press-fit implant model

P. Swider^{a,*}, G. Guérin^a, Joergen Baas^b, Kjeld Søballe^b, and Joan E. Bechtold^c

^a Biomechanics Group, IMFT UMR CNRS 5502, University of Toulouse, France

^b Department of Orthopaedic Surgery, University of Aarhus, Aarhus Amtssygehus, Aarhus, Denmark

^c Biomechanics Laboratory, Midwest Orthopaedic Research Foundation and Minneapolis Medical Research Foundation, Minneapolis, MN, USA

Abstract

Orthopaedic implant fixation is strongly dependant upon the effective mechanical properties of newly formed tissue. In this study, we evaluated the potential of modal analysis to derive viscoelastic properties of periprosthetic tissue. We hypothesized that Young's modulus and loss factor could be obtained by a combined theoretical, computational and experimental modal analysis approach. This procedure was applied to ex vivo specimens from a cylindrical experimental implant placed in cancellous bone in an unloaded press-fit configuration, obtained after a four week observation period. Four sections each from seven textured titanium implants were investigated. The first resonant frequency and loss factor were measured. Average experimentally determined loss factor was 2% (SD 0.4%) and average first resonant frequency was 2.1 KHz (SD: 50). A 2D axisymmetric finite element (FE) model identified effective Young's modulus of tissue using experimental resonant frequencies as input. Average value was 42 MPa (SD: 2.4) and no significant difference between specimens was observed.

In this pilot study, the non-destructive method allowed accurate measure of dynamic loss factor and resonant frequency and derivation of effective Young's modulus. Prior to implementing this dynamic protocol for broader mechanical evaluation of experimental implant fixation, further work is needed to determine if this affects results from subsequent destructive shear push-out tests.

Keywords

Implant fixation; Dynamics; Modal analysis; Finite Element Method; Bone viscoelasticity

1. Introduction

Initial mechanical stability of an implant is a key for the long-term survival of the arthroplasty. Anchorage behavior is strongly dependant on the macroscopic properties of

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*Corresponding author at: University of Toulouse, Biomechanics Group, IMFT UMR CNRS 5502, Toulouse 3, CHU Purpan, Amphithéâtre Laporte, Place Dr. Baylac, 31056 Toulouse cedex, France. Tel.: +33 5 6149 79 44; fax: +33 5 6149 6745. swider@cict.fr (P. Swider)..

Conflict of interest

The authors hereby declare to have no conflict of interest.

Appendix 1. Supporting Information

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newly formed tissue and on the time distribution of *in situ* loading. Activities by the patient can be categorized as quasi-static, cyclic and vibratory (Usui et al., 1989; Pattijn et al., 2007). The identification of mechanical parameters describing quality of implant fixation is generally determined by static shear push-out test (Søballe et al., 1992; Vestermark et al., 2004; Helgason et al., 2008). Dynamic vibration tests have also been performed in maxillo-facial surgery for dental and cochlear implants (Williams et al., 1995; Meredith 1998; Rasmusson et al., 1999; Gan et al., 2002; Bell, 2004; Atsumi et al., 2007; Elliott et al., 2007). For general engineering structures, robust techniques have been developed to characterize structures using non-destructive dynamic approaches with both numerical and experimental modal analysis (Adams & Bacon, 1973; Nashif et al., 1985; Zinoviev and Ekmakov, 1994; Swider et al., 1996; Gibson, 2000). It was our intent to apply these techniques to an experimental implant with the aim of obtaining additional information on which to characterize implant fixation.

In this study, we specifically evaluated the potential of modal analysis to derive viscoelastic properties of periprosthetic tissue. We implemented a combined theoretical, computational and experimental approach with modal analysis to obtain loss factor, resonant frequency, and to derive dynamic effective Young's modulus of interfacial tissue. We used an experimental canine implant that had been in place for four weeks in an unloaded press-fit configuration (Daugaard et al., 2008).

2. Materials and methods

A brief summary of the theory guiding numerical analysis and application to the experimental modal analysis is provided here. Further detail is found in Refs. Meirovitch (1975) and Nashif et al. (1985).

2.1. Theoretical background

2.1.1. Material viscoelasticity, resonance frequency and hysteretic damping of a single degree of freedom (SDOF) system—A viscoelastic model includes time-dependent properties of tissue (Fung 2004). In forced harmonic motion, material viscoelasticity may be modelled by coupling structural hysteretic damping and complex stiffness model (Nashif et al., 1985). This approach simplifies the investigation of dynamic behavior directly in the frequency domain.

Fig. 1a shows a single degree of freedom system (SDOF: k^* , m) submitted to a forced harmonic motion of pulsation Ω of its base. Complex stiffness $k^* = k(1+j\eta)$, accounts for both structural stiffness k and loss factor or damping factor η (complex notation: $j^2 = -1$). The steady state motion of mass m is expressed in Eq. (1). For low damping ($\eta \ll 1$), maximum magnitude of x and of the relative displacement $x_r = x - x_0$, occur when the angular forcing frequency Ω matches the resonant frequency of the undamped system $\omega = (k/m)^{1/2}$. This defines the amplitude resonance. Since η is low, the phase resonance appears for the same forcing frequency and the phase angle φ measured between the imposed displacement x_0 of the base and the response x of the mass, is $\pi/2$. With hysteretic damping model, resonance is independent from the dissipative properties of materials.

$$x = x_0 \left[\frac{1 + \eta^2}{(1 - (\Omega/\omega)^2)^2 + \eta^2} \right]^{1/2} \cos(\Omega t - \varphi) \quad (1)$$

with $\varphi = \alpha + \beta$, $\tan \alpha = \eta$, $\tan \beta = \eta/[1 - (\Omega/\omega)^2]$ and

$$x_r = \frac{(\Omega/\omega)^2}{\left[\left(1 - (\Omega/\omega)^2 \right)^2 + \eta^2 \right]^{1/2}} x_0$$

Energy dissipated per cycle is proportional to the structure stiffness k and to the square of base displacement x_0 , as shown in Eq. (2a). At resonance, it is also inversely proportional to loss factor, η , Eq. (2b). The loss factor, η , accounts for the source of energy dissipation and it is obtained using the half-power bandwidth (−3db), Eq. (2c). The limits of the bandwidth (Ω_1 , Ω_2) is corresponding to one-half of the energy dissipated at resonance ω .

$$D = \frac{\pi \eta k (\Omega/\omega)^4}{\left(1 - (\Omega/\omega)^2 \right)^2 + \eta^2} x_0^2 \quad (2a)$$

$$D = \frac{\pi k x_0^2}{\eta} \quad (2b)$$

with

$$\eta = (\Omega_2 - \Omega_1) / \omega \quad (2c)$$

2.1.2. Numerical modal analysis of a continuous structure—When the hysteretic model is used, complex elastic coefficients (E^* , C^* , ν^*) are the counterparts of the classical elastic modulus (Nashif et al., 1985). These coefficients are involved into the complex stiffness matrix $K^* = K(1+j\eta)$. For low damping ($\eta \ll 1$), it is assumed that dissipative effects do not influence the modal behavior of the structure: the real parts of elastic coefficients condition the undamped stiffness matrix K and the associated modal base (ω_i , ϕ_i).

This modal base satisfies the eigenvalue problem (3a) with K being the structural stiffness matrix and M being the structural mass matrix. As shown in Eq. (3b), resonant frequencies can also be expressed as the ratio of modal stiffness k_i related to the modal strain energy, over modal mass m_i related to the modal kinetic energy. This means that for each resonant frequency ω_i , the dynamic behavior of the continuous structure is described by an equivalent SDOF system (k_i , m_i) with k_i and m_i the modal stiffness and the modal mass, respectively. Results and conclusions of §2.1.1 keep all their relevance.

$$(\omega_i^{-1} I - K^{-1} M) \phi_i = 0 \quad (3a)$$

$$\omega_i^2 = \frac{k_i}{m_i} = \frac{\phi_i^T K \phi_i}{\phi_i^T M \phi_i} \quad (3b)$$

As shown previously, imaginary parts of complex elastic coefficients contribute to the dissipative energy, which contributes to the structural dynamic response. The theoretical prediction of complex material dissipative properties has not been solved, so in practice, modal damping factor η_i of mode shape ϕ_i is still experimentally based.

2.2. Application to a canine implant model

2.2.1. Specimen preparation and experimental modal analysis—Specimens were obtained from a previous study investigating osseointegration of press-fit unloaded canine implants (Daugaard et al., 2008). Implants were cylinders of Ti6AlV4 alloy with a nominal diameter of 6 mm and a length of 12 mm (Biomet® Warsaw, USA). They were inserted press-fit in the extra-articular cancellous bone site perpendicular to the medial plane of the proximal tibia 17 mm below the joint line (Fig. 2a). After an observation period of four weeks, implants and surrounding bone were removed and stored at -20°C . Section as shown in Fig. 2b was cut perpendicular to the implant axis using a water-cooled diamond band saw and they were nominally $h = 1.6$ mm thick. For our study, we used chemically milled micro-texture surface implants located in right tibia (R) and left tibia (L) of seven dogs. Two sections from each implant were harvested: implant outer section (o) and implant inner section (i). Finally, 28 specimens were investigated: $(L_i, R_i; L_o, R_o) \times 7$.

The use of the SDOF system described in §2.1a for the theoretical analysis dictated the design of the experimental approach. As shown in Fig. 1b and c, the bone section was clamped using two rigid blocks of polyethylene so that the specimen resembles a clamped cylindrical structure with a point mass in the center. The clamping was at $r_I = 8 \text{ mm} \pm 1\%$ and the radius of the implant was $r_O = 3 \text{ mm} \pm 1\%$. A hole in the implant at $r_2 = 1.5 \text{ mm} \pm 1\%$ centrally located the specimen in the fixture. Implant mass and fixture mass were $1.53 \times 10^{-4} \text{ kg} \pm 0.5\%$ and $0.4 \text{ kg} \pm 0.1\%$, respectively. Average mass of tissue surrounding the implant was of $3.32 \times 10^{-4} \text{ kg} \pm 1\%$. It was estimated by averaging the weight of specimens in the companion study (Daugaard et al., 2008) after destructive static push-out test. High ratios of fixture mass over specimen mass and of fixture stiffness over specimen stiffness were obtained. This represented the theoretical conditions in the SDOF system (§2.1a). We investigated the first natural frequency ω_I associated with a symmetric mode shape ϕ_I showing the maximum displacement of the central titanium part.

To generate the base excitation, the fixture was mounted on an electrodynamic shaker (Gearing&Watson® Hailsham, UK) using a piezoelectric force gage (8200 Brüel&Kjaer® Nærum, Denmark), Fig. 2c. An eddy current sensor (TQ102 Vibrometer® Fribourg, Switzerland) attached to the fixture allowed the relative displacement x_r of the central titanium part to be measured (Fig. 2c and d). Base displacement x_0 was monitored imposing a swept sine in the range 20 Hz–2.5 KHz. Experiments were managed at constant temperature of $22^{\circ}\text{C} \pm 2\%$. Frequency response functions $X_r/F(\Omega)$ were computed using a Dynamic Signal and System Analyzer (Siglab, Spectral Dynamics® San Jose, USA).

2.2.2. Numerical modal analysis—The specimen was assumed to be a clamped thick plate with polar symmetry in geometry, boundary conditions and material property distribution. For this last assumption, we used our previous study (Daugaard et al., 2008) showing that newly formed tissue was generally homogenous cancellous bone. Material properties were assumed to be homogenous isotropic (Table 1). We investigated the first symmetric mode shape ϕ_I of an undamped modal base derived from the matrix system derived above (3a). A 2D axisymmetric model was implemented with FE software (Marc/Mentat-MSC software®). The meshing was made of Q8 elements with two translational DOF per node, quadratic displacement field, and linear strain and stress fields. The mesh seed was 19 for the titanium part (r_2-r_O), 63 for the tissue (r_O-r_I), and 20 for the specimen thickness (h). The meshing shown in Fig. 3 was designed to provide a reliable prediction of strain gradients especially through the specimen thickness.

To quantify the influence of geometry and material properties on the first resonant frequency, we developed a sensitivity analysis based on a perturbation technique. We started from the first order Taylor expansion. The relative variation $\Delta\omega_I/\omega_I$ expressed in Eq. (4a)

was derived from the difference after and before perturbation δa of parameter a . The first derivative s_a was obtained using an order one finite difference scheme. For geometry, a was successively r_0 , r_1 , r_2 , and h . For titanium properties, a was successively E_b , ν_b , ρ_b , and for bone tissue a was successively E_b , ν_b , ρ_b . According to Eq. (4b), sensitivity vector was normalized with the maximum component scaled to unity.

$$\frac{\Delta\omega_1}{\omega_1}(a) = \sum_a s_a \delta a + O(\delta a^2) \quad (4a)$$

with

$$s_a = \frac{\omega_1(a+\delta a) - \omega_1(a)}{\delta a}$$

$$S_a = s_a \times (a/\omega_1) \left[\sum_i s_i^2 (i/\omega_1)^2 \right]^{-1/2} \quad \text{with} \quad (4b)$$

$$\max(S_a) = 1$$

We finalized the computational analysis by first expressing modulus E_b as a continuous polynomial function of clamping radius r_1 and resonance frequency ω_1 , the aim being to identify dynamic properties from measured frequencies. To proceed, successive direct finite element (FE) calculations first allowed generating the database of ω_1 using as input radius $r_1 = 8 \text{ mm} \pm 12.5\%$ and Young's Modulus of bone tissue $E_b = 40 \text{ MPa} \pm 100\%$. Second, the database was inverted to express E_b as a response surface of r_1 and ω_1 . A 5° polynomial interpolation was implemented. The relative variation $\Delta E_b/E_b$ to the variations of r_1 and ω_1 was calculated using a similar expression as Eq. (4a) and the response surface was a 4° polynomial interpolation. This post-treatment was managed using custom-made subroutines in Matlab package (Mathworks®).

3. Results

3.1. Experimental modal analysis

We firstly quantified the repeatability of measurements and results were lower than 0.3% for resonance frequency and lower than 0.5% for loss factor. The frequency response plotted in Fig. 3a clearly shows the resonance frequency $\omega_1 = 2111.5 \text{ Hz}$, the phase angle $\alpha + \beta \approx \pi/2$. The resonance is clearly shown by the representation of the frequency response as a quasi-perfect circle in the Nyquist plane (Fig. 3b).

The first natural frequencies and loss factor are listed in Table 2. Mean values obtained from three successive measurements for each specimen were in the range 2065–2129 Hz with standard deviations lower than 2%. Overall loss factor was 2.2% with standard deviation of 0.42%.

3.2. Numerical modal analysis

The meshing convergence test is plotted in Fig. 4. It was shown that the convergence of frequency relative variation was rapid and this agreed with the known result in numerical modal analysis. The final meshing involving 1640 elements, 5125 nodes, and 10,168 free nodes. With the geometrical and material properties of the FE model shown in Table 1, ω_1 was 2045 Hz and the symmetric mode shape ϕ_1 showed a lateral deflection x with no nodal circle. Fig. 5 shows the mode shape ϕ_1 with the distribution pattern of modal mass per element in greyscale.

Sensitivities to a variation of +1% for each parameter were calculated according to Eq. (4b) and are reported in Table 1. Absolute values were ranked as follows: $S_{Er} \approx S_{vt} < S_{r0} < S_{vb} < S_{vb} < S_{r2} < S_{vt} < S_{Eb} < S_h < S_{r1}$. Finally, the sensitivity of r_1 to fixture geometry r_1 , thickness h , and material property of tissue E_b were predominant.

The polynomial interpolation of E_b as a function of ω_1 and r_1 is plotted in Fig. 6a and the corresponding function is detailed in Appendix A1. The reverse polynomial interpolation of E_b as a function of ω_1 and r_1 provided mean experimental material properties of 42 MPa (SD 2.43 MPa), with no significant differences between specimens (Table 3, 3rd row). Fig. 6b shows the relative variation of tissue mechanical property to the variations resonance frequency and clamping radius. The polynomial function is detailed in Appendix A2 (see supplementary material). The sensitivity was bilinear with a sensitivity to the fixture geometry r_1 , greater than that of the frequency ω_1 . In Eq. (A2), coefficients b_1 and b_2 relative to r_1 were predominant.

3.3. Statistical analysis

Data were analyzed by group listed in Table 2: L_p , L_o , R_p , R_o . Each group involved seven frequencies, seven loss factors, and seven effective moduli. Each group was successively compared to other groups that led to six combinations as listed in Table 3. Variance was different for each group and a student's t-test with a heteroscedastic bilateral distribution was implemented with Excel package®. p -Value less than 0.05 was considered statistically significant. As summarized in Table 3, results did not show any significant difference from left to right specimens for both inside and outside zone of implantation.

4. Discussion

We implemented modal analysis to identify the viscoelastic properties of newly formed periprosthetic tissue. We focused on two complementary parameters: the dynamic stiffness involving the effective Young's modulus of tissue and the dissipative properties identified by the tissue loss factor. The use of this method has been evaluated in an experimental canine implant model.

According to §2-1b and in particular with Eq. (3b), distribution of modal mass and modal stiffness were related to modal kinetic energy and modal strain energy. As shown in Fig. 5, kinetic energy of first mode shape was concentrated into the implant and it was close to zero into the tissue because of their low mass. In contrast, strain energy was maximal into the biological tissue. Their intrinsic material properties were accurately derived by the dynamic testing method.

We evaluated only the first mode shape. The distribution of strain energy was complicated mainly because of the small slenderness ratio of specimens: $(r_1 - r_0)/h \approx 3$. The consequences were couplings between longitudinal and transverse strains whose effect was the warping of cross-section shown in Fig. 4. In this context, analytical structural models were inadequate and the identification of tissue material properties required a numerical approach by the FEM.

The sensitivity analysis of the numerically derived first frequency ω_1 showed that the main influence was the imposed specimen clamping radius r_1 . First frequency was also sensitive to specimen thickness h , and tissue effective module E_b see Table 1. Poisson's ratio and density played a limited role compared to implant geometry and elastic properties. In agreement with theory, an increase of effective modulus and specimen thickness stiffened the structure and led to a frequency increase (sign +), whereas reverse influence (sign -) was seen with an increase of the clamping radius.

The results from the experimental study were very repeatable despite the high natural frequencies measured (~2 kHz). The signals were of high quality as shown by the classic appearance of the Nyquist and Bode plots in Fig. 3, and experimental results were in complete agreement with the underlying theory. Since the fundamental properties of resonance were used, a low magnitude of base excitation x_0 , a few hundredth of millimeter, was required to obtain the first resonance frequency. No apparent alteration of biological specimen was observed.

When first frequencies of seven specimens were compared (Table 3, 1st row), we did not find any statistically significant differences ($0.06 \leq p \leq 0.72$). The implants were identical in terms of surface properties; implantation conditions were well controlled and reproducible. The mean loss factor η was 2% with no significant difference between specimens and implants (Table 3, 2nd row). In this setting, the contribution of newly formed tissue to the dynamic stiffness of implant anchorage and dissipative properties was constant.

The measurement method provided an average value of effective Young's modulus E_b of 42 MPa (SD 2.43 MPa). The ultimate uncertainty of the protocol was the results of discrepancies inherent to the experimental modal analysis and intrinsic numerical errors in FE resolution and polynomial interpolation. In fact, these parameters were involved into the relative variation $\Delta E_b/E_b$ expressed in Eq. (A2 see Appendix). Measurement uncertainties of frequencies $\Delta\omega_1/\omega_1$ were lower than 2%. Uncertainty on the clamping radius of the specimen $\Delta r_I/r_I$ was estimated lower than 1%. We obtained an overall uncertainty of 8% for the effective Young's modulus.

Numerous studies dealing with bone mechanical properties and elasticity–density relationships for cancellous bone have been published (Augat et al., 1998; Kang and Friedman, 1998; Morgan et al., 2003; Helgason et al., 2008). Results show large spread in predicted values depending upon species, anatomical sites and measurement method and interpretation. Despite this, we observed that Young's modulus was in the lower range of values reported in the literature. While these values are admittedly from very few canine studies, the four week observation period and potential for immature new bone or existing bone resorbing at the drill hole are plausible. The low apparent bone density observed in our tibia specimens could also explain this result.

Finally, our macroscopic approach required us to assume that the tissues were of the same biological nature and were homogeneously distributed. This was correlated by histomorphometric investigations previously developed (Daugaard et al., 2008). In addition, even in this well-fixed press-fit implant model, the significant loss factors of ~2% confirmed a viscoelastic behavior of the periprosthetic tissues.

Further work will investigate the correlation between macroscopic viscoelastic properties, tissue type and distribution pattern by histomorphometry and micro-CT. We will evaluate implant settings with different interface conditions expected to produce more fibrous tissue (e.g., with an initial gap surrounding the implant or insecure fixation). We also will specifically evaluate whether this low force/low displacement modal analysis is non-destructive with regards to alterations in subsequent static shear tests.

Supplementary Material

Refer to Web version on PubMed Central for supplementary material.

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Fig. 1. (a) Single degree-of-freedom system excited by its base. (b) Diagram of the experimental device. The eddy current sensor measured the transverse displacement ($x_r = x - x_0$) of the central titanium part (implant).

Fig. 2.

Experimental set-up. (a) Implant location (Ti6Al4V 6×12 mm) inserted press-fit in cancellous bone of the canine proximal left tibia. (b) Specimen with r_0 : implant radius, r_1 : clamping radius of the specimen, r_2 : whole radius inside the implant. (c) Assembling of the fixture on the electrodynamic shaker. (d) Location of the eddy current sensor.

Fig. 3. Experimental frequency response $X_r/R\Omega$ of specimen #1: (a) bode plane: ω_1 (2110 Hz) corresponded with the maximum magnitude of $X_r/R\Omega$ and phase angle ϕ of -90° ; (b) Nyquist plane: ω_1 corresponded with the maximum magnitude of vector $X_r/R\Omega$ located by ϕ (-90°). Loss factor η (1.9%) was calculated using the half power frequency bandwidth.

Fig. 4. Convergence of the finite element model. Geometrical data and material properties are listed in Table 1. The meshing increased from 50 to 71176 DOF and $\delta(\%)$ was the relative variation of the current frequency with the convergence frequency $\omega_1 = 2045$ Hz.

Fig. 5. Finite Element model of a specimen: meshing, boundary conditions and first mode shape ϕ_1 ($\omega_1 = 2045$ Hz). The distribution of modal mass was plotted in greyscale: the modal mass and modal kinetic energy were maximal in the implant and minimal in the clamping zone where the surrounding bone was secured. In contrast, the modal stiffness and modal strain energy were maximal in bone tissue. Warping of cross-sections as shown by deformed mesh, revealed the influence of shear.

Fig. 6.

(a) Five degree polynomial interpolation of effective Young's modulus E_b using clamping radius r_1 and resonance frequency ω_1 obtained from the FE model. (b) Sensitivity of interpolated effective Young E_b to ω_1 ($\pm 10\%$) and r_1 ($\pm 10\%$). $E_0 = 40$ MPa.

Table 1

Geometrical and material properties of the FE model.

	<i>a</i>	Sensitivity S_a
h (mm)	1.6	0.38
r_0 (mm)	3	-0.01
r_1 (mm)	8	-1
r_2 (mm)	1.5	0.11
E_t (MPa)	116	0.0
ν_t	0.36	0.0
ρ_t (kg/m ³)	4507	-0.17
E_b (MPa)	40	0.26
ν_b	0.3	0.03
ρ_b (kg/m ³)	570	-0.09

S_a was the sensitivity of first frequency ω_1 to a variation of $+1\% \times a$.

Table 2

Experimental results: resonance frequency ω_l , loss factor η and Young's modulus E_b

#	L_i			L_o			R_i			R_o		
	ω_l (Hz)	η (%)	E_b (MPa)	ω_l (Hz)	η (%)	E_b (MPa)	ω_l (Hz)	η (%)	E_b (MPa)	ω_l (Hz)	η (%)	E_b (MPa)
1	2083.2	1.5	41.5	2153.8	2.9	44.4	2111.5	1.9	42.7	2071.4	2.1	44.3
2	2117.3	2.6	42.9	2171.2	2.4	45.7	2161.9	2.1	44.7	2055.3	1.5	40.4
3	2082.0	2.1	41.4	2116.6	2.1	42.3	2114.9	2.9	42.8	2081.3	3.2	41.4
4	2114.8	2.5	42.8	1899.0	1.8	34.5	2113.3	2.1	42.7	2040.3	2.3	39.8
5	2136.4	2.4	43.7	2036.9	2.5	39.7	2119.4	2.2	42.9	2084.6	2.2	41.6
6	2126.2	1.6	43.2	2002.6	2.5	38.3	2130.7	1.6	43.4	2123.9	2.9	43.2
7	2199.0	2.0	46.3	2010.6	2.1	38.7	2150.5	2.1	44.3	2137.5	2.5	43.7
Mean	2122.7	1.9	43.1	2055.8	2.3	40.5	2128.9	2.1	43.4	2087.1	2.3	42.1
SD	39.4	0.4	1.6	96.9	0.3	3.8	19.9	0.4	8.2	37.8	0.5	1.7
SD%	1.9	22.7	3.7	4.7	14.5	9.4	0.9	18.4	1.9	1.8	21.8	4.0

Locations of specimens are L_i left inner tibia, L_o left outer tibia, R_i right inner tibia, R_o right outer tibia.

Table 3

Statistical analysis: student test p -values of resonance frequency ω_1 , loss factor η and effective Young's modulus E_b .

	p -Values		
	ω_1	η	E_b
L_i-L_o	0.13	0.12	0.25
L_i-R_i	0.72	0.34	0.73
L_i-R_o	0.12	0.20	0.25
L_o-R_i	0.09	0.32	0.10
L_o-R_o	0.45	0.45	0.35
R_i-R_o	0.06	0.58	0.10

Locations of specimens are L_i left inner tibia, L_o left outer tibia, R_i right inner tibia, R_o right outer tibia.