In silico dynamic characterization of the femur: physiological versus mechanical boundary conditions

E. Reina-Romo, J. Rodríguez-Vallés, J.A. Sanz-Herrera

School of Engineering, University of Seville, 41092 Seville (Spain)

Abstract

It is established that bone tissue adapts and responds to mechanical loading. Several studies have suggested an existence of positive influence of vibration on the bone mass maintenance. Thus, some bone regeneration therapies are based on vibration of bone tissue under circumstances of disease to stimulate its formation. Frequency of loading should be properly selected and therefore a correct characterization of the dynamic properties of this tissue may be critical for the success of such orthopedic techniques. On the other hand, many studies implement vibration techniques with *in silico* models. Numerical results are exclusively dependent on properties of bone tissue, i.e. geometry, density distribution and stiffness, as well as boundary conditions. In the present study, the influence of boundary conditions and material properties on the dynamic characteristics of bone tissue was explored in a human femur. Bone shape and density were directly reconstructed from computer tomographies, whereas natural frequencies and modes of vibration were obtained for different boundary conditions including physiological and mechanical ones. Results of this study show the moderate effect of material properties compared to the much substantial effect of boundary conditions. A factor of 2 in the natural frequency was obtained depending on imposed boundary conditions, highlighting the importance in the selection of appropriate conditions in the analysis of the bone organ.

Keywords: Bone Mechanics, Finite Element Method, Natural Frequency, Modal Analysis.

1. Introduction

It is established that bone tissue adapts and responds to mechanical loading. The characteristics of the mechanical loads, such as peak magnitude of ground reaction force, peak rate of force production, and repetition rate of these loads, are known to be important in regulating bone regeneration [1]. According to experimental studies, maximum bone formation directly depends on loading frequencies and targeted locations of bones [2-6]. Zhao et al. [7] have demonstrated in an *in vivo* study performed on mice that the application of loads with frequencies near the natural frequency of bone favors bone formation. Rubin et al. [8] have shown that very-low-magnitude, frequency vibration of 30Hz, increases significantly (by 34.2 %) the density of the trabecular bone in the proximal human femur compared to controls. Therefore, a possible application to increase bone mass under circumstances of disease would be to apply vibration loading stimulation: whole-body vibration, bone bending, axial loading, and joint loading [10-13]. Whole-body vibration seems to prevent/reverse sarcopenia and possibly osteoporosis [9]. Vibration analysis techniques have been used successfully to determine mechanical properties of human bone [14], to monitor fractures [15-16], to quantify the stability of dental implants [17-18] and to detect forms of femoral prosthesis loosening [19]. In the orthopaedic field, this technique may have potential due to apparent benefits it offers.

The vibration characteristics of bone have been analyzed both experimentally [10,20-21] and numerically [7,14,19,22-23]. Although Weiss et al. [24] have shown the importance of the boundary conditions in vibration studies in a hip endoprothesis system, the most commonly used boundary conditions are the free ones [14,22], which are known to be far from the *in vivo* environment. For example, Campoli et al. [22] already analyzed by means of finite element analysis how the natural frequencies changed by varying the density and shape of the human femur in free boundary conditions. The behavior of the femoral head prosthesis has also been analyzed *in silico* with modal analyses assuming free boundary conditions and constant mechanical properties [23]. Pérez & Seral-García [19] used other boundary conditions to simulate numerically the change in the resonance frequency during the osseointegration process of a cementless human hip system. In that study mechanical properties were assigned based on the level of Hounsfield Unit (HU).

A number of studies have already shown that geometry and material properties at boundaries and spatiotemporal distributions within bone have a profound effect on bone mechanobiology variables such as fluid velocities and pore pressures [25-26]. However, as far as the authors know, there is no vibration study that has numerically analyzed the influence of the properties of bone tissue, i.e. geometry, density distribution and stiffness, as well as boundary conditions on the dynamic characteristics of bone. Therefore, the aim of this study is to analyze *in silico* the dynamic characteristics of a human femur under different boundary conditions and assumptions found in the literature. Different numerical set-ups of the femur, which differed in the material properties assignment and boundary conditions, have been analyzed. On the one hand, either constant material properties or a technique based on HU was used. On the other hand, four different boundary conditions are implemented and analyzed: free, diaphysis, condyle and physiological boundary conditions with the purpose of determining the dynamic characteristics of the femur as closely as possible as they can be found *in vivo*. Using the information provided in this study, a proper determination of bone resonance frequencies could be conducted, which in turn may be of special relevance to calibrate existing vibration therapies performed clinically and to help to improve the treatment of some diseases such as the osteoporosis.

2. Materials and Methods

2.1. Geometry

A set of 664 computer axial tomographies (CATs) of a cadaveric adult male human femur were selected. CAT resolution includes a pixel size of 0.78x0.78 mm and a slice thickness of 0.8 mm. Geometry was built by means of the software $\mathrm{MIMICS\ 10.0}^{®}$ as follows: outer (cortical) and inner (marrow) regions were selected manually at diaphyseal region attending to the grey scale level of pictures, as well as spongy bone at proximal areas. Other regions of non interest in the study, such as hip, knee or tibia were not included in the geometry. Then, 3D geometry of the right femur was reconstructed by using the above mentioned software.

2.2 Boundary conditions

Four different boundary conditions are implemented and analyzed in the paper, namely, free, diaphysis (fig. 1a), condyle (fig. 1b) and physiological boundary conditions (fig. 1c) [27]. They are explained next.

Free boundary conditions

In this case, femur was free of loading and displacement restrictions. This is the most suitable boundary conditions used both in experimental modal analysis [14,20] as well as numerical modal analysis of the bone tissue [14,22-23], due to both reliability and simplicity at the laboratory level and finite element analysis.

Diaphysis boundary conditions

Displacements are prescribed at nodes in the mid-diaphysis where the cut was performed (see fig. 1a) [27,28]. Both nodes at exterior (cortical) and interior (marrow) regions were considered.

Condyle boundary conditions

All the displacements are prescribed at the three nodes shown in fig. 1b in the distal condyles.

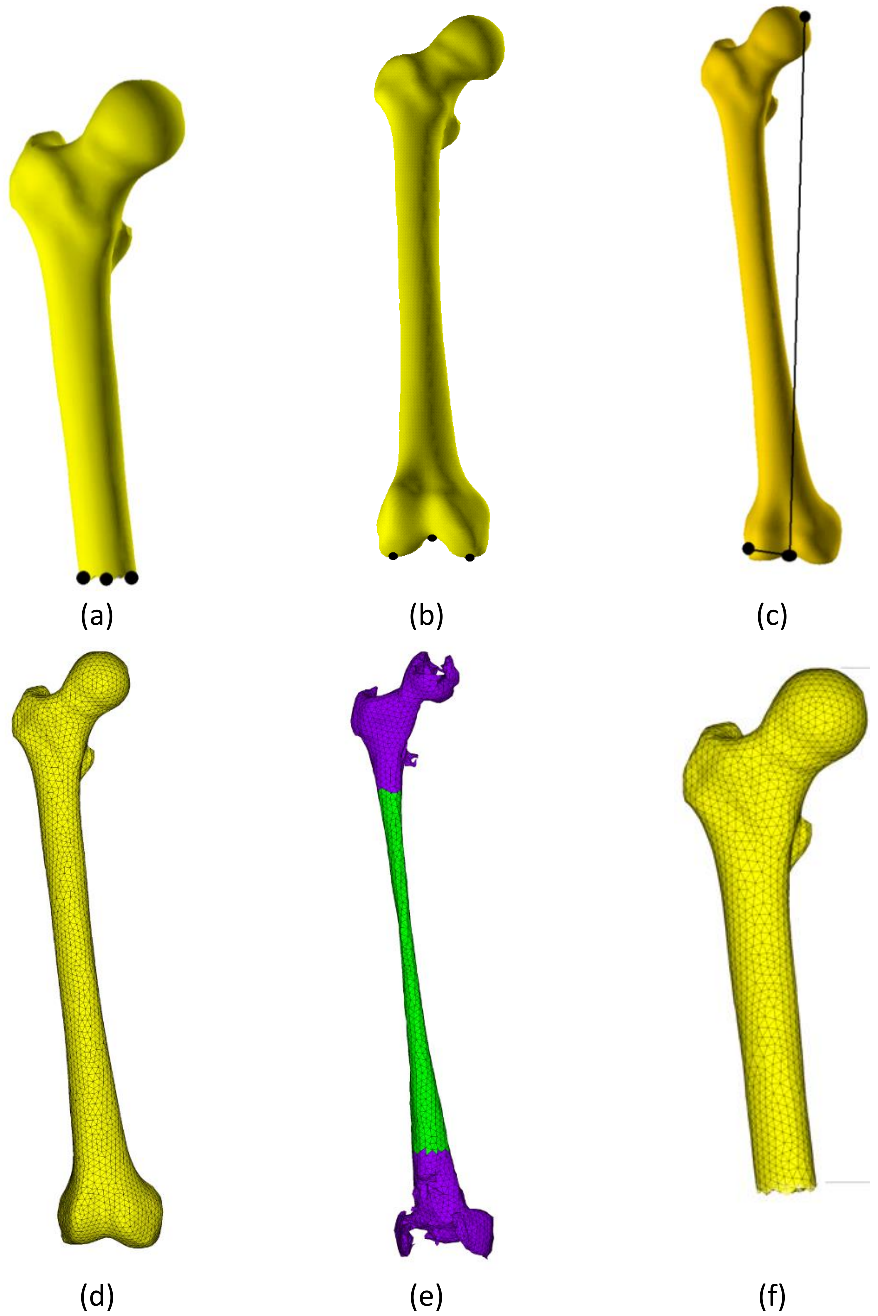


Figure 1. Finite element model of different implemented boundary conditions: (a) diaphysis, (b) condyle and (c) physiological. Tetrahedral finite element mesh at (d) the exterior (cortical), (e) the interior (marrow and spongy bone) regions; (f) finite element mesh for the diaphysis boundary conditions analysis.

Physiological boundary conditions

The so-called physiological conditions are the ones proposed by Speirs et al. [27]. In that work, authors conclude that this set of boundary conditions is the one that best represents the femur deformations at real conditions. They are implemented as follows (fig. 1c). First, the three translational degrees of freedom at a node placed in the knee centre are prescribed, specifically at the joint of the tibia with the femur.

Then, two displacements at a node in the femur head are also prescribed. At this node, only the displacement in the direction along an axis towards the knee center is allowed. Finally, the sixth degree of freedom was constrained at a node placed on the distal lateral epicondyle in order to avoid rigid body motions (see fig. 1c).

2.3 Finite element mesh

Femur geometry was exported into the finite element software ANSYS 14.5®. Tetrahedral 8 node 3D quadratic elements (SOLID 185) were used including 60227 elements and 12020 nodes in the femur mesh (see figs. 1d and e).

For diaphysis boundary conditions, a different finite element model was used (see fig. 1f). In this case a cut was performed at the mid-diaphysis [27] (see fig. 1f).

2.4 Mechanical properties of bone tissue

Linear, elastic and isotropic mechanical properties were considered in this study [14,22]. Material properties were characterized by Young's modulus and Poisson's ratio. Nonetheless, femur domain was considered to be heterogeneous, such that each finite element in the model was featured by different mechanical parameters. Mechanical properties have been assigned with two different approaches: depending on the level of HU, regardless its location and, manually, considering three different material sets.

Firstly, the technique based on HU was implemented. Bone density was linearly correlated with HU from 0,5 $\mathrm{g/cm^3}$ of the spongy bone to 1,952 $\mathrm{g/cm^3}$ of the cortical bone [22]. Mathematically, this relation attends to the following curve:

$$\rho\ (\mathrm{g/cm^3}) = 0{,}000968 \times \mathrm{HU} + 0{,}5 \tag{1}$$

where ρ is the bone tissue density and *HU* is the Hounsfield Unit, which varied from 0 to 1500. Bone density is related to the mechanical properties following Beaupre et al. [29] as:

$$\text{If } \rho \leq 1{,}2\ \mathrm{g/cm^3}\ ;\ \mathrm{E} = 2014\rho^{2{,}5}\ \text{(MPa)},\ \nu = 0{,}2 \tag{2}$$

$$\text{If } \rho > 1{,}2\ \mathrm{g/cm^3}\ ;\ \mathrm{E} = 1763\rho^{3{,}2}\ \text{(MPa)},\ \nu = 0{,}32$$

where E and ν are the Young's modulus and Poisson's ratio, respectively. A set of 10 values of density, uniformly distributed from lower (0,5 $\mathrm{g/cm^3}$) to upper (1,952 $\mathrm{g/cm^3}$) bounds, respectively, were considered. Consequently, a number of 10 different pairs (E, ν) were estimated for the mechanical properties along the bone tissue through Eq. (2) according to its estimated density. This was included as different materials in the finite element mesh by using utilities of MIMICS 10.0® and ANSYS ICEM 14.5® softwares.

Secondly, the finite element model was tested assuming constant mechanical properties. Three different material sets were distinguished: cortical bone, bone marrow and spongy bone (fig. 1e). In this case, mechanical properties are given in table 1.

Material	Density $(\mathrm{kg/m^3})$	Young's modulus (GPa)	Poisson's ratio (-)
Cortical bone [22]	1800	13	0,3
Bone marrow [29]	1060	0,001	0.5
Spongy bone [22]	500	0,6	0,12

Table 1. Mechanical properties of finite element model of the femur when considered as constant.

2.5 Analysis type

A linear modal testing analysis was used on the FE model to assess the vibration characteristics of the human femur. The modal analysis package of ANSYS software was used. To implement the numerical procedure, first the mode-extraction method to be used for the modal analysis is chosen. In this study, the Block Lanczos method is selected since it is a fast, efficient and robust algorithm to perform modal analysis. Next the frequency range has to be defined. To include the first five modes of vibrations, a range from 0 to 1000 Hz is taken.

A harmonic response analysis was also performed to obtain the vibration amplitude of the femur at a specific frequency range under a vertical load applied at the femoral head for the three different cases analyzed (condyle, diaphysis and physiological boundary conditions). This linear analysis has also been performed using the commercial software ANSYS. The harmonic response analysis method chosen is the Full method within a frequency range from 0 to 2000 Hz for all the boundary conditions analyzed.

The numerical analyses were run in a laptop PC Intel 1.8 GHz (1 core) with 8GB RAM. CPU time of the analyses was estimated in 8 minutes.

3 Results

Table 2 summarizes the dynamic characteristics of the femur analyzed. The first five natural frequencies as well as their corresponding modes of vibration are detailed for the different boundary conditions (BC) analyzed (free BC; diaphysis BC; condyle BC; physiological BC) and the two different material properties set-ups performed (constant mechanical properties and Hounsfield Units). The vibration plane of the modes of vibration is specified: frontal bending, sagittal bending, combined bending (both frontal and sagittal bending) and torsion (T).

Natural frequencies range from 245 to 814 Hz for the free boundary conditions, 108 to 887 Hz for the diaphysis boundary conditions, 222 to 803 Hz for the condyle boundary conditions and 107 to 782 Hz for the physiological boundary conditions. In order to compare qualitatively the four analyzed cases, the normalized natural frequencies for the different set-ups analyzed have been represented (fig. 2). Normalization has been performed with respect to the lowest frequency. It can be observed the moderate effect of material properties compared to the much substantial effect of boundary conditions. In particular, a high difference can be found from case to case of boundary conditions, being a factor higher than 2 in some frequencies. The biggest differences are found between free and condyle boundary conditions versus physiological ones. On the other hand, differences due to the fact of considering constant mechanical properties along the bone tissue are not so acute although important: a difference in the range 5-25% can be found for different boundary conditions across natural frequencies. Higher variations are found for the case of the condyle boundary conditions, whereas considering constant mechanical properties for the remaining cases is an assumable hypothesis (fig. 2).

To illustrate the modes of vibration for one of the set-ups analyzed, fig. 3 shows the deformed shape of the bone for the first five natural frequencies in the case of the physiological boundary conditions and constant mechanical properties assignment. It may be observed that sagittal bending modes are given at the first and fifth natural frequencies (fig. 3 and Table 2), frontal bending modes at the second and fourth natural frequencies and torsional mode at the third natural frequency. In the case of free boundary conditions, frontal bending modes are shown at the fourth natural frequency, sagittal bending modes at the fifth frequency, torsional mode at the third natural frequency and combined bending modes for the first and second natural frequencies (Table 2). For the diaphysis BC, frontal bending is shown for modes 1 and 4, sagittal bending for modes 2 and 5 and torsional bending for mode 3. On the other hand, in the case of the condyle boundary conditions, Table 2 shows front sagittal modes at natural frequencies 1 and 4, frontal bending modes at 2 and 5 and torsional mode at the third natural frequency.

The results of the harmonic analysis are shown in fig. 4. It shows the dimensionless dynamic amplification factor, i.e. the vertical displacement vector, versus its corresponding value in a static analysis with the same loading conditions, in a spectral plot, for the different analyzed boundary conditions in the case in which material properties are assigned based on the HU method. As can be observed, the natural frequencies differed in the three cases as well as the amplitude of vibration pointing out the importance (quantitatively and qualitatively) of the boundary conditions chosen on the dynamical response obtained (fig. 4).

		Free BC		**Diaphysis BC**		**Condyle BC**		**Physiological BC**	
		Const	**HU**	**Const**	**HU**	**Const**	**HU**	**Const**	**HU**
1	**ω_1 (Hz)**	271	245	111	108	276	222	146	107
	ϕ_1	CB	CB	FB	FB	SB	SB	SB	SB
2	**ω_2 (Hz)**	310	274	122	119	398	283	269	165
	ϕ_2	CB	CB	SB	SB	FB	FB	FB	FB
3	**ω_3 (Hz)**	660	567	592	531	468	378	330	240
	ϕ_3	T	T	T	T	T	T	T	T
4	**ω_4 (Hz)**	790	703	826	738	775	643	702	483
	ϕ_4	FB	FB	FB	FB	SB	SB	FB	FB
5	**ω_5 (Hz)**	814	760	887	861	803	716	782	619
	ϕ_5	SB	SB	SB	SB	FB	FB	SB	SB

Table 2. Summary of the dynamic characteristics of the femur: natural frequencies ω (Hz) and associated vibration modes ϕ. Results are shown for the different analyzed boundary conditions (BC): free BC; diaphysis BC; condyle BC; physiological BC and for the two different material properties set-ups performed: constant mechanical properties (Const) and Hounsfield units (HU). For the modes of vibration, the corresponding vibration plane is specified: frontal bending (FB), sagittal bending (SB), combined bending (CB) and torsion (T).

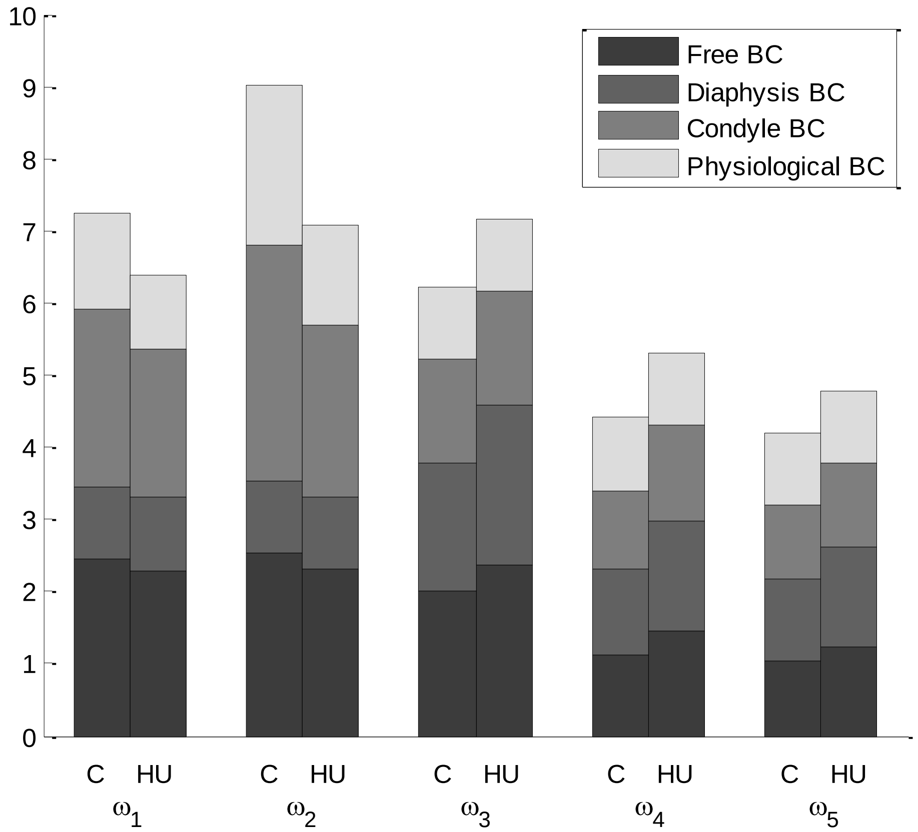
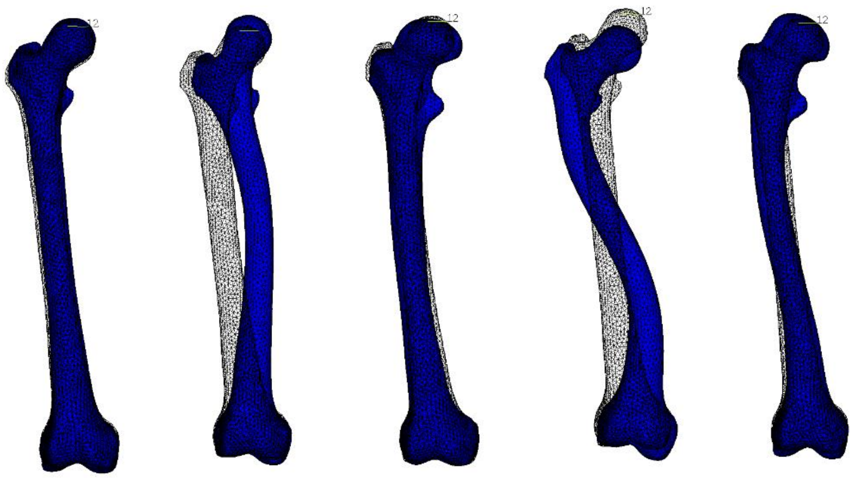


Figure 2. First 5 normalized natural frequencies (ω) of the femur under different analyzed boundary conditions (BC): free, diaphysis, condyle and physiological for the two mechanical properties assignment set-ups: constant mechanical properties (C) and Hounsfield Unit (HU). Normalization has been performed with respect to the lowest frequency for each set of analyzed boundary conditions and each natural frequency.



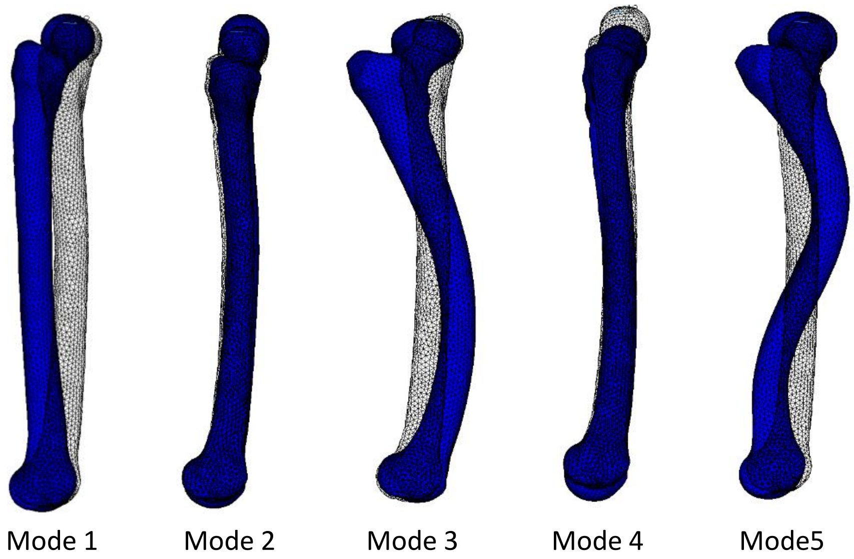


Figure 3. Vibration modes (associated to natural frequencies presented in table 2) of femur under physiological boundary conditions in the constant mechanical properties set-up. Upper part – frontal plane, bottom part – sagittal plane.

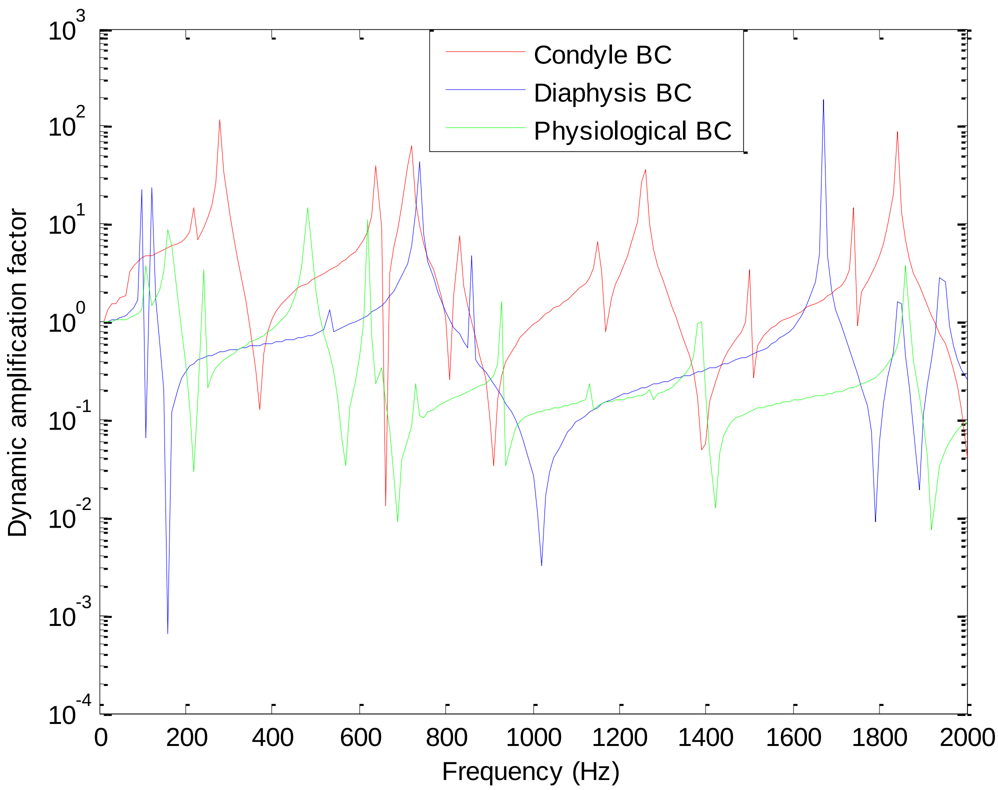


Figure 4. Dynamic amplification factor obtained for the condyle, diaphysis and physiological boundary conditions (BC) analyzed.

4 Discussion

The aim of the study presented in this paper was to highlight the great differences in the dynamic behavior (i.e. natural frequencies and vibration modes) of bone tissue when subjected to different boundary conditions. Differences are given not only referred as the value of natural frequency, but also on the shape of associated vibration modes as well as spectral dynamical behavior. As it can be seen in the results section, a proper selection of boundary conditions in modal analysis of femur is critical in order to establish right conclusions from such a study. On the one hand, free boundary conditions do not really represent actual boundary conditions of bone organ at physiological conditions. Nonetheless, they are commonly used as a benchmark and calibration of modal analysis of bone tissue, or even to establish analogies between bone tissue characteristics and (generic) dynamic behavior, both experimentally and numerically [14,20,22-23]. On the other hand, diaphysis boundary conditions at the bone organ are often used in different biomechanical analysis [19]. They are of application when some information of the geometry is missing, or even to alleviate computer resources of finite element analysis. However, as these conditions do not represent physiology of bone organ, they are not of application in a dynamic analysis nor can be of application as a model simplification hypothesis given the differences versus other implemented boundary conditions. Condyle boundary conditions are typically prescribed at surrounding locations where bone organ connect to other tissues, i.e. joints. They are used as well in a number of static biomechanical analyses and based on static equilibrium by prescribing displacements at these regions. Analysis and results are then of application far from the area where boundary conditions were applied. Again, condyle boundary conditions are non-physiological ones.

Conclusions taken of diaphysis and condyle boundary conditions for the dynamic case may not be of application at static conditions: in studies which consider static cases, boundary conditions are independently applied to find an equilibrated stress state, and hence equivalent, in the bone organ. Therefore, good fittings are found elsewhere in the literature when results are compared at these situations [31-32]. On the other hand, when the analysis is conducted in a dynamic scenario, results are strongly dependent on boundary conditions, as exposed in the present work, which have a minor importance in static problems.

The estimation of the dynamic characteristics of the human femur may contribute to improve the vibration therapies in bone tissue. For that purpose, physiological boundary conditions were selected as in Speirs et al. [27]. In that study authors conclude that such restrictions allow obtaining physiological levels of femur deflection when subjected to physiological range of forces. The analysis performed by the authors showed that other boundary conditions yielded to unreliable femur head deformations. Physiological boundary conditions are followed in other studies [33-34] and are accepted as a reference for this kind of analysis. The biomechanical analysis presented in Speirs et al. [27] was static and its availability is extended here to the dynamic case. To the best of author’s knowledge there is no reliable information about dynamic characteristics of human femur experimentally in order to compare and validate the analysis here presented.

There are several limitations of the study that should be commented. First, geometry has been taken from a single femur. Results of this study could be generalized using virtual models with different geometries [35]. Secondly, 10 different materials (mechanical properties) are chosen in this study in the HU set-up. According to Pérez & Seral-García [19], no significant resonance frequency changes occurred regardless of the number of material groups chosen. In addition, the effect of other bones and soft tissues of the musculoskeletal system and body on dynamic characteristics has been considered indirectly with the boundary conditions analyzed. Of course modeling them directly will dampen the vibration in situ and *in vivo*. However, this study is only the first step in understanding and dealing with the dynamic behavior of the femur. Additional experimental *in vivo* and *in vitro* studies are required to validate and improve the numerical modeling started with this study. These tasks are planned as a future work in the context of the present study.

The importance of predicting dynamic characteristic of bone tissues and organs has implications in many clinical scenarios. As it was reviewed in the introduction, vibration technique is a trending clinical therapy for bone mass regeneration under circumstances of bone disease such as osteoporosis [36]. There is not still a consensus on the protocol. However, whole body vibration is a promising technique [37-41] in which human body is subjected to cyclic vibration at different amplitudes in a range varying from 10-90 Hz [42]. In addition, frequencies in the range of resonance (natural frequency) of bone tissue provide positive outcomes as an alternative clinical therapy [43]. For this application, *a-priori* knowledge of bone tissue natural frequency is of critical importance in order to calibrate the setup of the clinical protocol. In a different context, knowledge of dynamic characteristics of bone tissues and organs has a great importance in the analysis of dynamic fracture behavior of bone [44] as well as long-term fatigue response of bone tissue to loads [45].

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Conflict of interest statement

No conflict of interest to disclose.

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