MECHANICAL BEHAVIOUR OF A PROSTHESIZED HUMAN FEMUR: A COMPARATIVE ANALYSIS BETWEEN WALKING AND STAIR CLIMBING BY USING THE FINITE ELEMENT METHOD

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ABSTRACT

The aim of this study is to perform finite element investigation of the mechanical behaviour of a prosthesized human femur during walking and stair climbing. Such activities are in fact encountered with the highest frequencies during daily living. In order to numerically analyze the stress shielding of the femoral bone with an artificial hip replacement, the strain and stress distributions both in the femur and in the stem were evaluated by using the finite element method. From a set of CT images the geometry of the femur was recovered and meshed. An operation of virtual surgery allowed to insert the metal stem (constructed by a CAD code) in the medullary canal. Numerical simulations showed evidence that, when comparing walking with stair climbing for stresses and strain energy density, maxima and minima values change dramatically, even if the localizations remain the same ones. Furthermore, for each type of daily activity, the unloading of bone tissue is confirmed in the case of prosthesized femur with respect to the physiological one, and hence the well-know phenomenon of stress shielding is exhibited. The development of a computational model allowed to deal with the complexity of the biomechanical problem and to describe quantitatively the mechanical behaviour of bone tissue in contact with the metal stem. The calculated results motivate the performed analyses because they are sufficient to activate the mechanisms of deposition and resorption in the bone tissue at contact with the artificial biomaterial. Thus, stair climbing confirms to be a critical task for primary stability of the prosthesized femur with respect to the activity of walking.

Key words: *Finite Element Analysis, Implant Stability, Stress Analysis, Stress Shielding, Uncemented Total Hip Prosthesis.*

INTRODUCTION

Total hip replacement is a permanent implant in the host bone with the aim to replace partly o totally the damaged hip joint. The prosthesis allows the patient to return to normal daily activities and often provides relief from pain. Recent studies concentrated on activities with the highest frequencies during daily living [1] such as walking and stair climbing, which have been shown to be important activities for assessing hip implant performance due to the prevalence of the former [1] and the high loading demands of the latter [2,3] . In fact, walking represents the most frequent daily activity while stair climbing exhibits the highest twisting loads [1,2].The differences from level walking are reflected by changes in the ranges of motion of the different joints during gait, and changes in the phasic muscle activities and in the maximum joint forces and moments. The hip joint is composed of the head of the femur, the acetabulum of the pelvis and the trochanteric sector, and it is one of the most stable joints in the body. The stability is provided by the rigid ball-and-socket configuration [4]. The prosthesis for total hip replacement consists of a femoral and an acetabular components. The femoral stem is made up of Ti-alloy, is divided into head, neck, and shaft, and is fixed into a reamed medullary canal by cementation or press fitting. The femoral head is made up of CoCr alloy, alumina, or zirconia. Although Ti-alloy heads work well under clean articulating conditions, they have fallen into disuse because of their low wear resistance to third bodies, e.g., bone or cement particles. The acetabular component is generally made up of ultra-high molecular weight polyethylene (UHMWPE) [5]. It is essential that design criteria of an implant for joint replacement be based on both the kinematics and the dynamic load transfer characteristics of the joint [6]. It is reasonable to presume that the magnitude of the forces acting on the proximal femur in the daily life depend on the specific movement the patient is performing, as well as the strategy adopted to perform it. For istance, climbing up stairs, the differences are reflected by changes in the contractions of the soleus, quadriceps femoris, hamstrings, and gluteus maximus during the support phase; going down stairs, the differences are reflected by changes in the contractions of the soleus and quadriceps femoris muscles [7].

In a review article [8] on the interaction between rehabilitative procedures and the peri-prosthetic bone remodelling processes, it is suggested that the transition from the three-points walking to the two-points walking technique depends particularly on the conditions of the muscles stabilising the hip joint. The material properties, shape, and methods used for fixation of the implant to the patient determines the load transfer characteristics. This is one of the most important elements that determines long-term survival of the implant, since bone responds to changes in load transfer with a remodeling process [9]. Bone is able to undergo spontaneous regeneration and to remodel its micro- and macrostructure. This is accomplished through a delicate balance between an *osteogenic* (bone forming) and *osteoclastic* (bone removing) process. Bone can adapt to a new mechanical environment by changing the equilibrium between *osteogenesis* and *osteoclasis*. These processes will respond to changes in the static and dynamic stress history applied to bone; that is, if larger stress than the physiological one is applied, the equilibrium tilts toward more osteogenic activity. Conversely, if smaller stress is applied the equilibrium tilts toward osteoclastic activity [10]. Many specific stimuli for bone remodelling have been recently proposed as important regulators of functional adaptation [11], namely, strain which is a primary and directly measurable physical quantity representing local deformation, and strain energy density, which is proportional to the square of the strain tensor [12]. In addition to these strain based measures, the von Mises equivalent stress has been used as a mechanical stimulus for some bone adaptation models. It is a scalar measure of the stress at a point and, for isotropic materials, is proportional to the portion of the strain energy density associated with distortion.

The principal goal of uncemented prosthesis is the osteointegration [13], that is direct contact between prosthesis and bone without fibrous tissue interposed. Minimizing contact joint surface increases local stresses and damages bone. Poor initial stability causes ingrowth of fibrous tissue and consequent resorption of bone at the bone–implant interface. Besides, stress shielding causes resorption of the proximal bone, which is not loaded anymore. Consequently it occurs loss of mechanical stability [14]. The stress distribution of femur must be assessed to predict stress shielding. A problem with currently used implant metals is that they are much stiffer than bone, so they "shield" the nearby bone from mechanical stress. The artificial hip stem shields the stress on the femoral bone since the relatively stiff stem shares the load that is usually carried only by the bone. Consequently, the femoral bone tries to adopt this new environment of altered stress patterns by reducing its weight thinning or increasing its porosity [15,16].

Three dimensional finite element models of the prosthesis and bone have be built from the CT data. Two different prosthesis designs (distal filling stem and the distal short stem) have been evaluated by numerical simulation considering single leg stance and stair climbing as loading conditions [17]. Spears *et al*. [18] concluded in their study on the basis of the average data that stair climbing produced the worst conditions for bone in-growth around the hip socket. Pancanti *et al*. [14] reach similar conclusions for the femoral component, when average results are considered.

Experimental design process of an artificial hip joint requires time consuming and difficult analysis and evaluation [17]. However, precise three dimensional finite element models of prosthesis and bone could be built from the CT (Computed Tomography) data, and many different prosthetic designs could be evaluated by numerical simulation considering various clinical issues [19,17]. Besides, the results of experimental tests using the femurs from cadavers can be affected by the disease, race, and age of the specimens. A finite element model, however, which can represent general geometry, can be used to assess the prosthesis under specific loading conditions of actual daily activities [20]. However, in order to achieve the physiological relevance of such simulations, it is essential to utilize realistic stress distributions which correlate to those reported from *in vivo* measurements [21÷24]. Unfortunately, even the most extensive experimental study can only explore a small sub-set of all possible combinations found *in vivo*. To overcome this limitation, a combination of experimental and numerical methods has been proposed [25]. The primary stability of a cementless anatomical stem was assessed *in vitro*, and a finite element model was developed to accurately replicate the same experiment.

The determination of the mechanical stresses that physiological activities induce in human bones is of great importance in both research and clinical practice. Unfortunately, the mechanical stress in bones cannot be measured in living subjects without the use of an invasive surgical procedure [26], which, in general, is not ethically permissible. A possible way to estimate bone stresses non-invasively *in vivo* is ''subject-specific'' finite element modelling [27]. This procedure allows the creation of a numerical model of the bone segment from computed tomography (CT) images of that segment [28]. Computed tomography (CT) represents the method of choice for the generation of the finite element models of bone segments. a closer correlation was found [29] between experimentally predicted and FE model calculated bone stresses when distributed material parameters were assigned to the elements on the basis of computed tomography (CT) data for bone ash density. The use of a physiological load case is recommended [30] that simulates the activity of all the major muscles acting on the femur during normal walking. Moreover, numerical simulation by finite elements, of *in vitro* mechanical behaviour of human femur and cemented hip prosthesis was carried out by Kadi *et al*. [31] in order to investigate the influence of the cement thickness on the von Mises stress distributions generated along the stem, in the cement and along the femur. Cilingir *et al*. [27] investigated the contact mechanics and stress distribution of hemi-arthroplasty hip resurfacing was investigated using a cemented metallic component]; three-dimensional FE models were analyzed in that study in order to predict von Mises stresses, and stress shielding in the bone tissue was found to occur with the hip hemi-resurfacing prosthesis considered in that study.

In previous papers the authors analyzed the stress state in the total hip replacement, considering both a bidimensional [32] and a three-dimensional model of the bone-stem implant [33]. Aim of this paper is to investigate the stress shielding of the femoral bone with an artificial hip stem for the stair climbing in comparison with the walking under the hypothesis of perfect osteointegration, focusing the attention on the mechanical response of the bone in contact with artificial material. The complexity of the problem with regard to geometry, loads, boundary conditions and material data suggested to perform a three dimensional finite element analysis. The finite element model is able to predict the load transfer towards distal zones, the risk of failure in bone and stem and the static effects of the load eccentricity. Furthermore, the stress analysis developed in this paper could be used to evaluate the risk of slippage and detachment between bone tissue and stem at the contact interface, the so-called primary or initial stability. To this end, shear stress tangent and tensile stress normal to the separation surface should not exceed the cohesion and adhesion strengths respectively, in order to avoid slipping and detachment between prosthesis and bone. If this *a posteriori* check would be satisfied, the preliminary assumption of perfect osteointegration did hold.

MATERIALS AND METHODS

To create the three-dimensional model of physiological and prosthesized human femurs the CT data sets of a male femoral bone were used [34, 35]. High-resolution images of sequential cross sections of bone region were created. These images were digitized and stored. By stacking the cross-sectional images the original structure of femoral bone was rebuilt in the computer and the bone voxels were subdivided into tethaedron element of varying sizes. The sectional distance for each slice was 3 mm and the length of the thigh was 470 mm. A solid model inspired to the F2L Multineck stem (Limagroup, Italy) [36-24] was previously constructed by a standard CAD program, autoCAD 2004, and then implanted virtually into a solid model of the femur. The system for hip prosthesis is characterised by three components: head, neck and stem. The geometry of the system has been established as follows. The femoral head has a diameter of 32 mm, and is engaged to the stem through a standard 12/14 Morse taper connection; the neck has a length of 45 mm, is straight and has a cervical-diaphyseal angle of 135°; the stem has a length of 150 mm and is symmetrical with respect to the frontal plane. The three-dimensional geometric models and the finite element models of both the physiological (Fig. 1a) and prosthesized (Fig. 1b) femurs imported into a special-purpose solver, COMSOL 3.3 [37]. The operation of virtual surgery consists with the head femur resection and inserting the metal stem in the medullary canal, Fig. 2. The prosthesis and the bone tissue were characterized by using a 10 – node quadratic Lagrangian tetrahedral element [38]. The physiological model consisted of 25637 elements and 5679 nodes, while the prosthetic model was composed of 77219 elements and 14403 nodes. As a simplifying assumption, both cortical bone and trabecular bone were assumed to be homogeneous and isotropic material [39]. As stated above, perfect adhesion and cohesion were assumed at the interface between stem and bone. Material properties were chosen as follows: for the cortical bone Young's modulus E = 17.26 GPa and Poisson's ratio ν = 0.29, for the trabecular bone E = 324 MPa and ν = 0.29, for the hip stem E = 110 GPa and ν = 0.3 [40÷43]. The type of analysis is static linear.

While there is strong evidence that muscles are major contributors to femoral loading [44,45,23], the actual forces occurring *in vivo* are hardly accessible. To date, non-invasive measurement of *in vivo* muscle forces is still impossible. Ethical considerations discourage the use of invasive methods to determine muscle forces in humans. Therefore, the only opportunity to estimate the complex distribution of muscle forces is offered by computer analysis. *In vivo* hip joint loading has previously been investigated [2]. In the research of Heller *et al.* [46], four regular THA patients were implanted with instrumented femoral prostheses and contact forces at the hip joint under different activities of daily life, including walking and stair climbing, were measured; then,

generated and validated a computer model of the human lower extremity was generated and muscle forces were determined using optimization algorithms. Loadings had been simplified by grouping functionally similar hip muscles [46,47]. An instance of maximum muscular activity and high joint contact force was then selected for the activities of walking (Fig. 3a) and stair climbing (Fig. 3b). Walking represented in fact the most frequent daily activity while stair climbing exhibited the highest twisting loads [2,1]. From the complex musculoskeletal model, simplified but representative load cases of these two activities were selected and re-validated [3].
The origin of the coordinate system was located in the middle of distal condyles and uses x-axis from lateral to medial, y-axis from anterior to posterior and z-axis from distal to proximal directions. As shown in Fig. 3, the loading condition due to the walking and stair climbing were selected for a 87 Kg body weight (860 N) as reported in Table 1.
The displacement boundary conditions at the distal end of the femur are given as the element faces fixed in all directions.

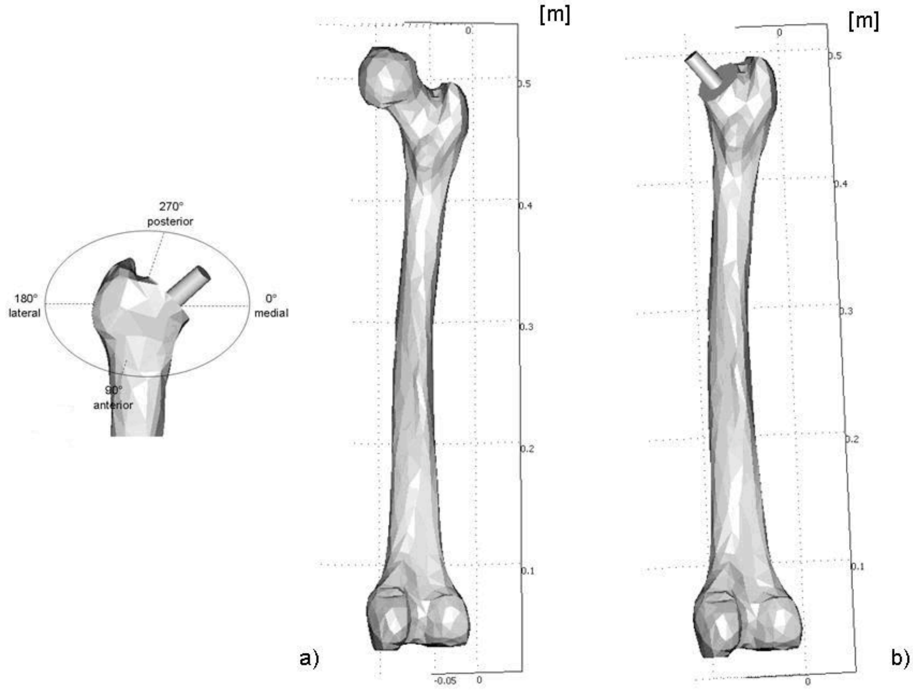

Figure 1. FEM models : physiological (a) and prosthesized (b) femurs.

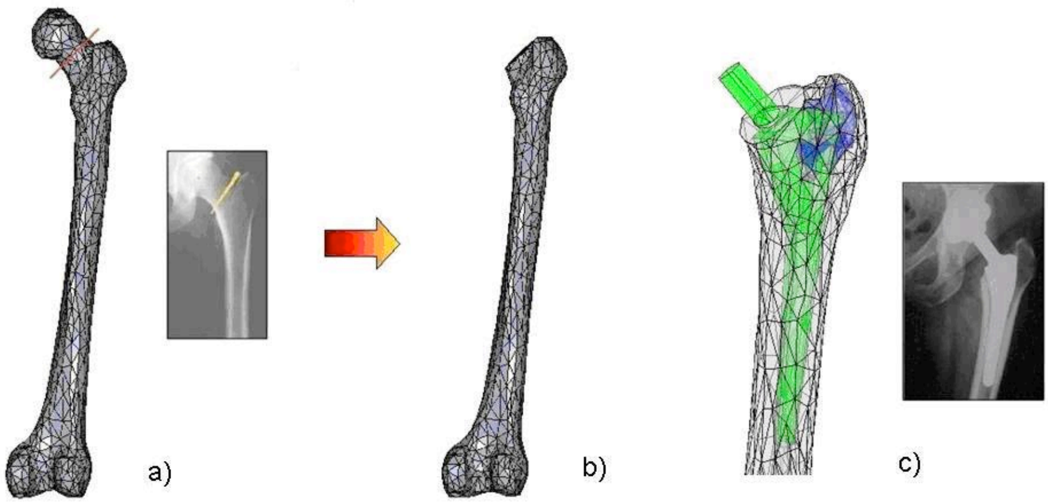

Figure 2. Virtual surgery.

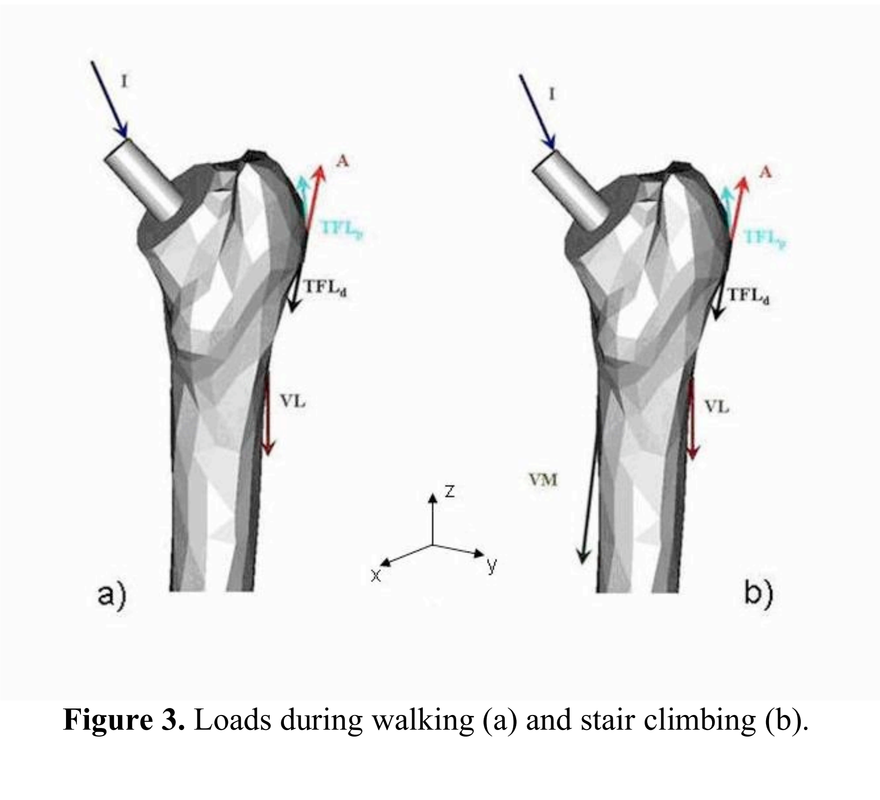

Figure 3. Loads during walking (a) and stair climbing (b).

Table 1. Mechanical loading on the prosthesized and physiological human femurs during the activities of walking and stair climbing [3].

Activity	Loads	Force	Force components [N]		
			x	y	z
Walking	Intersegmental and two-joint muscle force	I	162	17	- 1172
	Abductor	A	- 518	29	700
	Tensor fascia latae, proximal part	TFL_p	- 70	-87	113
	Tensor fascia latae, distal part	TFL_d	11	-4	- 159
	Vastus lateralis	VL	34	-199	- 766
Stair climbing	Intersegmental and two-joint muscle force	I	225	142	- 1272
	Abductor	A	- 639	-174	706
	Tensor fascia latae and ilio-tibial band, proximal part	TFL_p	- 122	-3	129
	Tensor fascia latae and ilio-tibial band, distal part	TFL_d	15	-2	- 198
	Vastus lateralis	VL	60	-257	- 1135
	Vastus medialis	VM	158	-470	- 2245

RESULTS

As far as the mechanical behaviour is concerned, a finite element analysis has been carried out for the prosthesized human femur during walking and stair climbing. The same calculations have been worked out for the corresponding physiological cases. With reference to the prosthesized state, the relevant numerical results are sketched through Figs. 4÷6, whereas, as far as the analysis of the sane state is concerned, only numerical values are reported herein for space limit and compared with those relative to the implanted femur. In the following, a detailed and quantitative comparison will be reported as far as the largest values and the relative variations of the significant parameters *are* concerned. Namely, (i) the principal stresses were considered in the femoral bone because they can be an important mechanobiological index in bone remodelling; (ii) the strain energy density

$$u_{ELASTIC} = \frac{1}{2E}\left[\left(\sigma_{I}^{2} + \sigma_{II}^{2} + \sigma_{III}^{2}\right) - 2\nu\left(\sigma_{I}\sigma_{II} + \sigma_{II}\sigma_{III} + \sigma_{III}\sigma_{I}\right)\right],$$

which is an average and integral measure of the state of stress and strain, is a favourite candidate to account for the consequence of the stiffness gradient between metallic stem and biological tissue.
In table 2 and 3, in addition to the above quoted parameters, also the values of the Mises stress, the normal stress in z-direction and the total displacement have been reported. The Mises stress defined in the following

$$\sigma_{MISES} = \frac{1}{\sqrt{2}}\sqrt{\left(\sigma_{I} - \sigma_{II}\right)^{2} + \left(\sigma_{II} - \sigma_{III}\right)^{2} + \left(\sigma_{III} - \sigma_{I}\right)^{2}}$$

is an important parameter for the failure and fatigue design of the implant [48]. The normal stress distribution in z direction due to combined pressure and bending, allows for detecting load eccentricity with respect to stem shaft's axis while the total displacement permits to check the global attitude of the patient's stance with reference to the slope of the stem head's neck. For space limit filled volume maps of Figs. 4-6 pictorially illustrate only the spatial distributions of the principal stresses and strain energy density, the influence of which has been investigated. Fill attributes are assigned so that there is a gradation change of colour from the minimum to the maximum 3-D domains. Furthermore, a colour scale shows the fill assigned to each colour on a filled volume map, and the numerical values for each level are displayed.

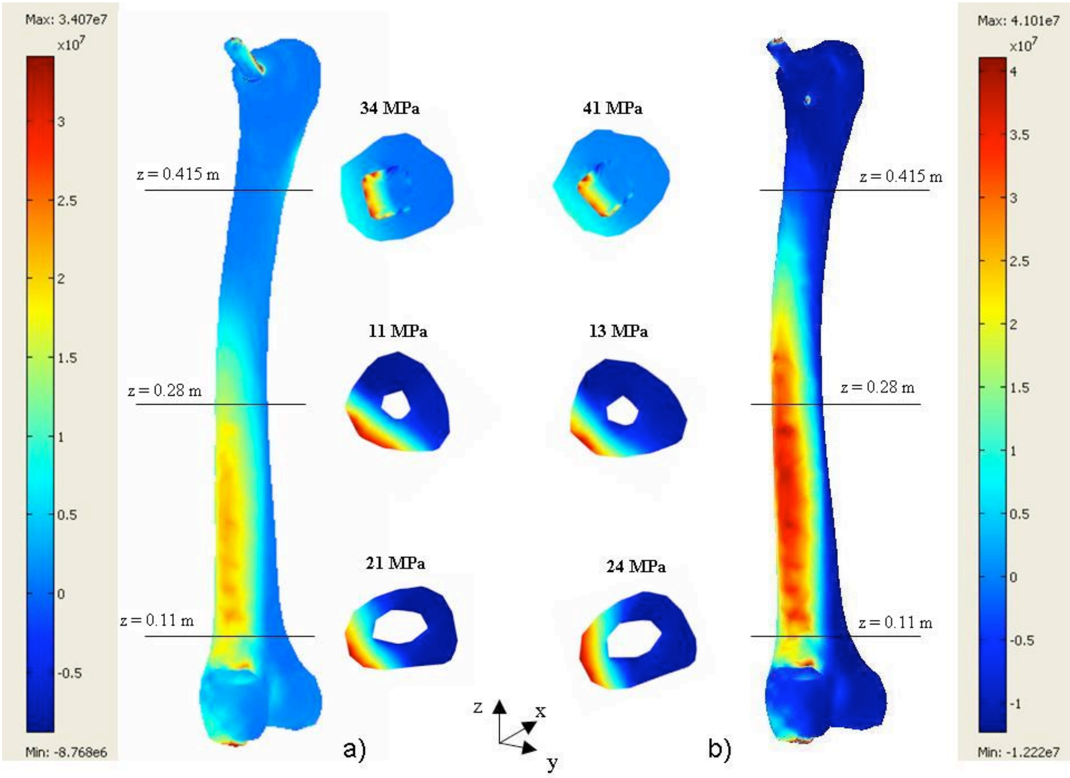

Figure 4. First principal stress (tensile) [Pa] during walking (a) and stair climbing (b).

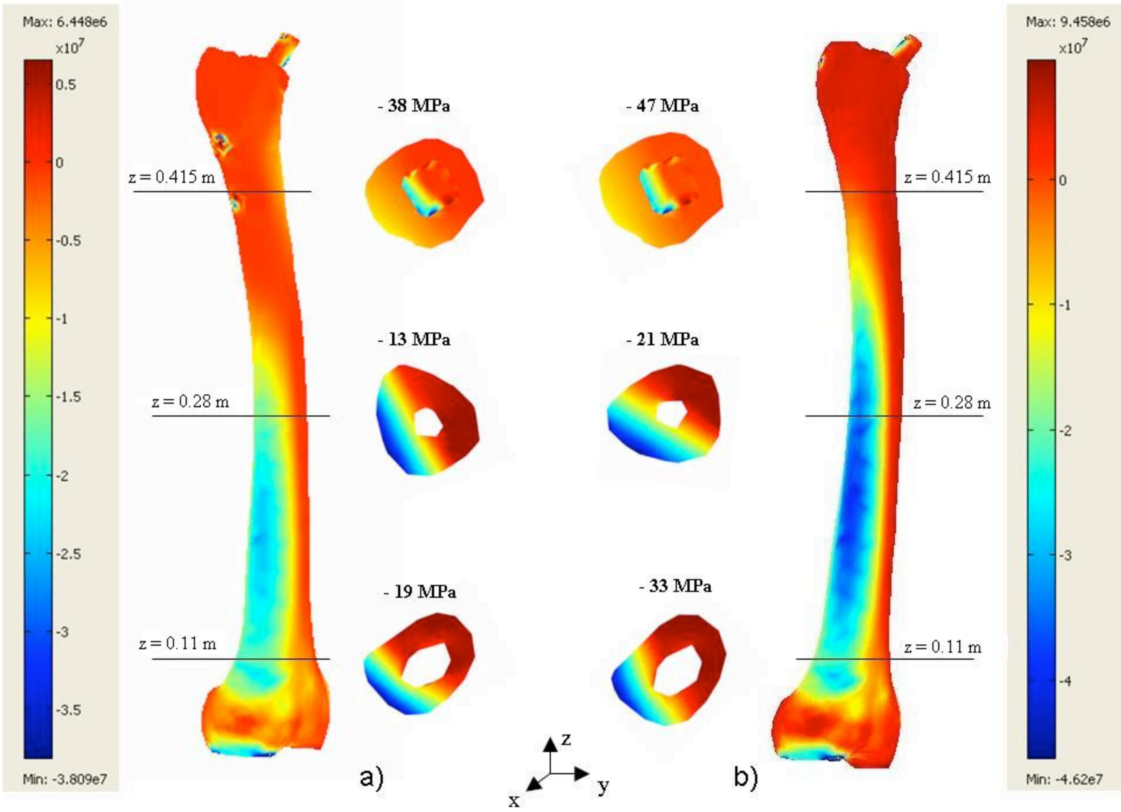

Figure 5. Third principal stress (compressive) [Pa] during walking (a) and stair climbing (b).

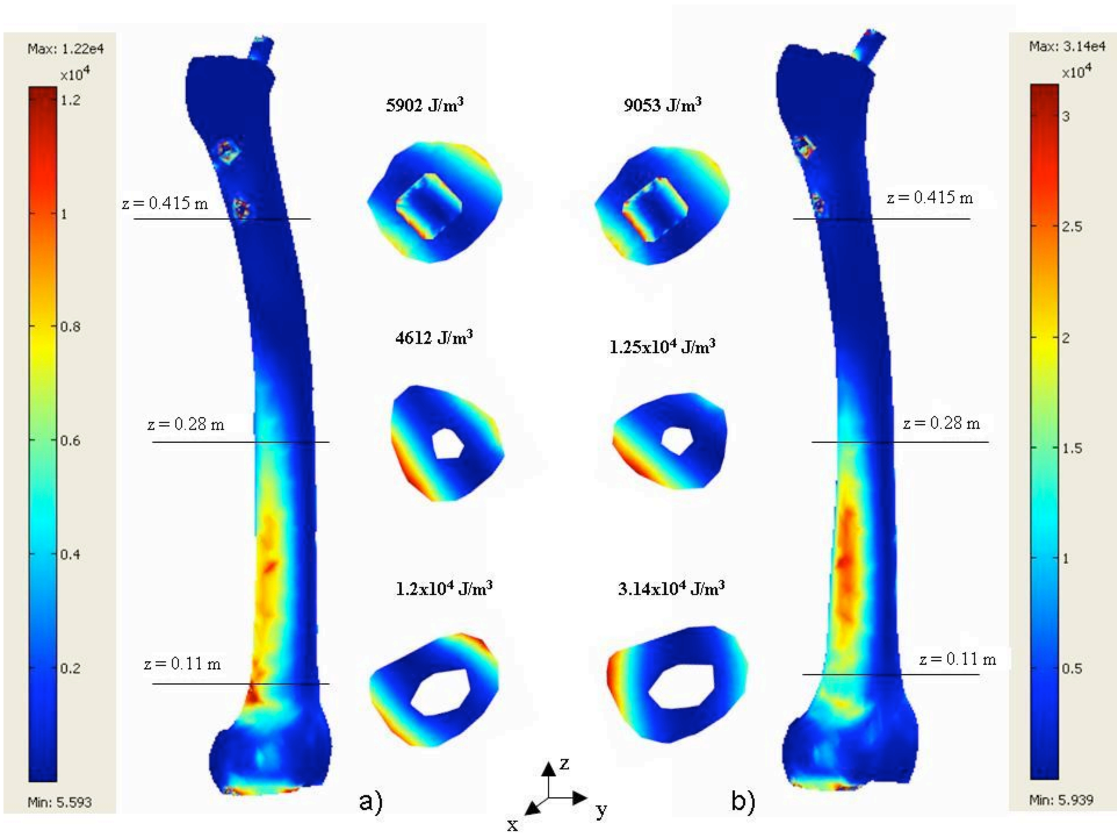

Figure 6. Strain energy density [J/m^3] during walking (a) and stair climbing (b).

Figures 4÷6 show both the overall behaviour of the three-dimensional structures and the bi-dimensional distributions of the quantities under consideration within 3 slices at proximal, medial and distal diaphyseal shaft level, respectively. In more details, during walking and stair climbing, the first principal stress attains its largest values in the distal and medial zones; the third principal stress and the strain energy density attain their largest values in distal and lateral zones. The tensile stress in z-direction attains is largest values in the distal and medial zones, while the compressive stress z-direction and the Mises stress attain their largest values in the distal and lateral zones. In both the analyzed cases, the total displacement attains its maxima in the proximal part of the femur neck and in the trochanteric sector.
Tables 2 and 3 report the maxima (first principal stress, strain energy density, Mises stress, tensile stress in z-direction, total displacement) and minima (third principal stress, compressive stress in z-direction) values of the investigated mechanical parameters and their percentual variations. The states which have been assumed as references for the calculation of the percentual variations are respectively walking and physiological cases.
In particular, table 2 emphasizes the comparison between the values of the above mentioned quantities during daily activities (namely, walking and stair climbing) for each type of femur (physiological and prosthesized). On the other hand, table 3 matches the values relevant to physiological and prosthesized femurs for each type of daily activity.

Table 2. Comparison between stair climbing and walking for physiological and prosthesized human femurs.

Physiological femur			
Mechanical parameter	*Walking*	*Stair climbing*	*Percentual variation*
First principal stress	17 MPa	25 MPa	47 %
Third principal stress	- 21 MPa	- 42 MPa	100 %
Strain energy density	1.3 x 10^4 J/m^3	3.3 x 10^4 J/m^3	153 %
Mises stress	21 MPa	33 MPa	57 %
Tensile stress in z-dir.	17 MPa	25 MPa	47 %
Compressive stress in z-dir.	- 19 MPa	- 40 MPa	110 %
Total displacement	0.0033 m	0.006 m	82 %
Prosthesized femur			
Mechanical parameter	*Walking*	*Stair climbing*	*Percentual variation*
First principal stress	34 MPa	41 MPa	21 %
Third principal stress	- 38 MPa	- 47 MPa	24 %
Strain energy density	1.2 x 10^4 J/m^3	3.14x10^4 J/m^3	162 %
Mises stress	31 MPa	44 MPa	42 %
Tensile stress in z-dir.	32 MPa	40 MPa	25 %
Compressive stress in z-dir.	- 25 MPa	- 42 MPa	68 %
Total displacement	0.0044 m	0.0057 m	30 %

Table 3. Variation of mechanical parameters for physiological and prosthesized human femurs in different daily activities.

	First principal stress (tensile) [MPa]	
	Walking	*Stair climbing*
Physiological femur	17	25
Prosthesized femur	34	41
Percentual variation	100 %	64 %
	Third principal stress (compressive) [MPa]	
	Walking	*Stair climbing*
Physiological femur	- 21	- 42
Prosthesized femur	- 38	- 47
Percentual variation	81 %	12 %
	Strain energy density [J/m^3]	
	Walking	*Stair climbing*
Physiological femur	1.3×10^4	3.3×10^4
Prosthesized femur	1.2×10^4	3.14×10^4
Percentual variation	- 8 %	- 5 %
	Mises stress (tensile) [MPa]	
	Walking	*Stair climbing*
Physiological femur	21	33
Prosthesized femur	31	44
Percentual variation	47 %	34 %
	Tensile stress in z-dir. [MPa]	
	Walking	*Stair climbing*
Physiological femur	17	25
Prosthesized femur	32	40
Percentual variation	88 %	60 %
	Compressive stress in z-dir. [MPa]	
	Walking	*Stair climbing*
Physiological femur	- 19	- 40
Prosthesized femur	- 25	- 42
Percentual variation	32 %	5 %
	Total displacement [m]	
	Walking	*Stair climbing*
Physiological femur	0.0033	0.006
Prosthesized femur	0.0044	0.0057
Percentual variation	33 %	- 5 %

DISCUSSION

Aim of this study was to establish and solve a biomechanical problem with coupled materials via the 3-D finite element method. Under the assumption of perfect osteointegration, the stress shielding of the femoral bone with an artificial hip stem was investigated for the walking and stair climbing as loading conditions.

Numerical tools have been used in order to avoid the difficulties to find solutions in closed form due to the complex behaviour of biological materials, loads and boundary conditions. Besides it is difficult to perform experimental tests on large scale.

The effect of these surgical variables on femoral loading was examined for walking and stair climbing using loads from a validated musculoskeletal model. Results of the implanted models were compared with an intact model to evaluate stress shielding.
One of the limits of the present study is its reliance on a simulation rather than on *in vivo* measurements. But it is worth to be noticed that the loads were adopted from a validated musculoskeletal model derived from direct *in vivo* measurements. However, it is not currently possible to measure directly and non-invasively the stresses induced by various activities. Thus, the modelling nature of this study is limited to the relationship between a system of external forces (measured *in vivo*) and the induced stress fields and strain energy densities. Then it is important to note that the trend of stress distributions both in stem and femoral components shown in Figs. 4÷6 are analogous to experimental findings [49,50,51], and also the results of finite element analyses [19,45]. Therefore, the results of the analysis have a fair degree of reliability.
In the present study, some mechanical stimuli have been adopted as important regulators of functional adaptation, namely principal stresses and strain energy density; in addition to these strain based measures, the von Mises equivalent stress has been used and computed, because – even being a scalar measure of the stress at a point - for isotropic materials is proportional to the portion of the strain energy density associated with distortion. It is worth to note that the strain energy density varies within a percentual range of -8 % ÷ 162 %. The quoted values sound as significant from an Engineering and physiological points of view. They motivate the performed analyses because they are sufficient to activate the mechanisms of deposition and resorption in the bone tissue at contact with the artificial biomaterial [10].
Cross-matched comparisons between the daily activities of walking and stair climbing on one hand, and the clinical states of physiological and prosthesized femurs on the other hand have been performed. These comparisons allowed to observe that the localizations of maxima and minima values of the mechanical quantities at hand remained substantially unchanged due to the loading conditions which differ only for the addition of the *Vastus medialis* muscle and the *Ilio tibial* band in the stair climbing with respect to walking. On the contrary, the relative variations of the parameters under examination exhibit significant alterations both from physiological to prosthesized states in walking (- 8 ÷ 100 %) e in stair climbing (-5 ÷ 64 %), and from walking to stair climbing in physiological (47÷ 153 %) and prosthesized femurs (21 ÷ 162 %).
Generally speaking, for each type of femur the comparison of different daily activities shows the increase of the mechanical effort required to the bone tissue to perform stair climbing with respect to walking, whereas for each type of daily activity, the unloading of bone tissue is confirmed in the case of prosthesized femur with respect to the physiological one.
On the basis of the present data, we may conclude that stair climbing is surely a critical task for primary stability of the prosthesized femur with respect to walking. The results reached by means of the numerical simulation encourage to continue in the future the structural analysis of the implanted prosthesis. An important study can examine a three-dimensional model of a complete implant made up of stem, head and cotile. The hip joint, characterized by a rigid ball-and-socket configuration and the stem-bone interface will need refined models of calculation, based on finite elements of unilateral and frictional contact. In fact, without the assumption of perfect osteointegration, it will be necessary to use finite elements able to simulate the phenomena of friction at the bone-stem interface and wear of the head sheath. At present, numerically modelling bone tissue and total hip prosthesis and simultaneously changing many variables (shapes, materials, lenghts etc.) might play an important role in improving implant design criteria, because this procedure allows to predict in real-time mechanical behaviour before the in vivo validation.

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