Original Article

Experimental strain analysis on the entire bony leg compared with FE analysis

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ABSTRACT

The present study addresses the question of evaluating, by combining both experimental and numerical approaches, the stress/strain distribution within a complete model of the entire lower bony chain. With this purpose an experimental model and a complete 3D finite element one were realised. A load of 700 N has been applied at the top of pelvis and the feet were rigidly fixed. Obtained results reveal interesting consequences deriving by taking into account the complete bony chain; it is possible to get information on load sharing between bones, location of high strain concentrations, and bone relative motion.

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1. Introduction

In general, bone is a good composite material, having an overall strength higher than either of its components, apatite or collagen. The softer collagen (low-modulus) prevents the stiff apatite (high-modulus) from undergoing brittle fracture, while apatite acts as a rigid scaffold to prevent collagen from yielding. Not surprisingly, the mechanical properties of bone are as complex and varied as the anatomy and composition. Seemingly simple properties such as bone strength, stiffness, and energy absorption to failure depend not only on material properties of bone (e.g., inherent composition, microscopic morphology of bone components, bonds between fibres and matrix and bonds at points of contact of fibres) but also on structural properties (e.g., geometry of whole bone, bone length, and bone curvature). Furthermore, it is well known that the material strength of bone varies with the age, sex, and species of animal under investigation and with the location of bone, such as femur versus humerus. In attempting to assess structural and material properties of bone using mechanical testing techniques, additional variation in bone strength may result from factors such as the orientation of load applied to the bone (since bone is anisotropic), strain rate (rate of deformation), and testing conditions, including tension versus compression, bending versus torsion, wet bone versus dry bone.

Mechanical testing may yield even wider dispersion in results when specifically applied to the material and structural properties of healing bone. Amount of callus, type of callus, and degree of callus reorganisation all affect the mechanical assessment of bone healing.

Bone, in performing its function, must withstand a complex pattern of imposed forces. In a static situation, bone acts largely to resist the forces of gravity, supporting the weight of the body and the attendant muscular activity necessary to maintain a given static posture. In a dynamic mode, however, such as during locomotion or athletic activity, these forces may be magnified many fold and may be omnidirectional.

Bone, in function, experiences two types of imposed forces. In general, intrinsic forces may be considered physiologic and are imparted to bone through articular surfaces by means of ligaments surrounding joints and at tendinous sites of muscle insertion. Under normal circumstances such forces sustain ground reaction forces during posture and gait and only under unusual circumstances do they approach the inherent breaking strength of bone. Extrinsic forces, on the other hand, originate from the environment and, unlike the intrinsic system, have no limitation on magnitude or direction of application, for example, automobile impact. Clearly it is these nonphysiologic forces that have the greatest potential to result in catastrophic bone failure (fracture) and that must be understood to evaluate the biomechanics of fracture aetiology.

Intrinsic and extrinsic forces act to cause microscopic deformations of bone. The degree of deformation is dependent on the magnitude of the imposed force, the geometry of the bone

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Fig. 1. Experimental model, finite element model with identification of reference nodes, loading and constrain conditions.

(size, shape, diameter, curvature), and the material properties of bone (cortical versus cancellous). It is intuitive that should the magnitude of imposed forces on bone exceed the ultimate strength of that bone, catastrophic failure will result. The introduction of finite element analysis (FEA) into orthopaedic biomechanics allowed continuum structural analysis of bone and bone-implant composites of complicated shapes. However, besides having complicated shapes, musculoskeletal tissues are hierarchical composites with multiple structural levels that adapt to their mechanical environment. Mechanical adaptation influences the success of many orthopaedic treatments, especially total joint replacements. Recent advances in FEA applications have begun to address questions concerning the optimality of bone structure, the processes of bone remodelling, the mechanics of soft hydrated tissues, and the mechanics of tissues down to the micro structural and cell levels, but still have deeply difficulties to analysed complied skeletal chains involving different bony parts, because of the model size, and above all the boundary conditions to impose. Many different works in literature have investigated the single bony part such as femur, knee, tibia, or feet, fixing the base and loading with more or less detailed systems of forces. In this paper this question is faced by adopting a simplified experimental and numerical models of the human leg, intended to assess how the stress shielding can influence the integrity and resistance of bones, if loaded with a vertical force of 700 N, and by comparing the obtained results with the ones presents in literature. The comparison of experimental strains measured on the model with those predicted by the finite element model revealed good agreement.

2. Materials and methods

2.1. Experimental model

Two couple of fourth-generation femur and tibia (*model 3403 and 3401*), a couple of fourth generation patella (*model*

3419), a fourth generation pelvis (*model 3415-1*), feet and a couple of fibula in a solid foam, produced by Pacific Research Labs., Vashon Island, WA, USA, were used to perform the experimental tests. These bone analogues model natural bone using a mixture of short e-glass fibres and epoxy resin for the corticalis (density 1.7 g/cm³, tensile modulus 12.4 GPa, compressive modulus 7.6 GPa, Poisson ratio 0.3, tensile strength 90 MPa, compressive strength 120 MPa) and a polyurethane foam core for the cancellous bone (density 0.27 g/cm³, compressive modulus 0.1 GPa, compressive strength 4.8 MPa). Left leg was instrumented with six resistive foil strain gauges (Hottinger Baldwin Messtechnik, Darmstadt, Germany), connected in quarter bridge to a digital data acquisition system (System 6100, Vishay Measurement Group, Raleigh, NC, USA). Unidirectional strain gauges (SGs) were positioned as shown in Fig. 1(a). Before bonding the SGsensors, the area for strain measurement was preliminary treated with sandpaper (#100), carefully cleaned and degreased first with ethanol, then with acetone and 2-propanol, and smoothed with fine sandpaper (#400). A bi-component epoxy adhesive was employed to glue the SGs onto the femur surface. Bones have been connected by elastic rods to simulate tendons by screws, while feet were rigidly constrained to a base; in order to avoid sliding between pelvis and femur, femur and tibia, and tibia and foot extremity of bones were rasped. Hence, a series of tests have been carried out on the entire model, having cared to fix the structure. Simulation of the loading conditions occurring in the stance of gait was obtained by applying a single axial load of 700 N on the top of the pelvis. Although this may appear as a rough approximation of the physiological loading condition, in which there are also several thigh muscles acting on the bones at different locations, the experimental set-up was greatly simplified, improving measurement repeatability and reproducibility. A material testing machine (Lloyd Instruments Inc., Fareham, UK); equipped with a 1000 N load cell ($\pm 0.5\%$ full scale accuracy) was used to impose the vertical load. The model was pre-loaded with 100 N, to asses equilibrium conditions, and then a

quasi-statically axial loads was imposed ranging from 100 to 700 N at 10 N/s. Test was repeated 3 times, repositioning the model in the testing machine each time, in order to determine repeatability of the experimental set-up and check for the absence of permanent deformations or damages.

2.2. Finite element model

The geometrical data of the model developed herein were obtained by a computerised tomography (CT) of artificial bones, separated at intervals of 1.5 mm in the sagittal, coronal and axial planes with the knee at 0° flexion (accuracy 0.5 mm). These lines were transferred into the commercial code Hypermesh by Altair® where the main surfaces and solid version of the model were reconstructed; in particular 6.842 elements and 2.345 nodes were used for pelvis, 21.457 elements and 1.302 nodes in femur, 2.843 elements and 564 nodes in patella, 2.945 elements and 987 nodes in fibula, 16.987 elements and 2.454 nodes in tibia and 990 elements and 456 nodes for the foot, see Fig. 1. On the upper zone, the ilio-femoral ligament, the ligament of the hip joint, which extends from the ileum to the femur in front of the joint, was modelled with 273 tetrahedral elements and 161 nodes. The knee joint constituted by the medial collateral ligament, which extends from the medial femoral epicondyle to the tibia, the lateral collateral ligament, which extends from the lateral femoral epicondyle to the head of the fibula, the anterior cruciate ligament which extends postero-laterally from the tibia and inserts on the lateral femoral condyle, and the posterior cruciate ligament which extends antero-medially from the tibia posterior to the medial femoral condyles were modelled with 601 tetrahedral elements and 306 nodes. On the lower zone, the foot joint, constituted by the plantar fascia, the medial and lateral ligaments were modelled with 386 tetrahedral elements and 199 nodes.

The six reference points, see Fig. 1(b), where SGS were positioned have been identified also on the numerical model, and verified comparing different measurements executed with a precision gauge. Non-linear finite element analysis of the models was performed with Abaqus version 5.4 (Hibbitt, Karlsson and Sorensen, Inc., Pawtucket, RI) using the geometric nonlinearity and automatic time stepping options. Material properties were defined as nonlinear elastic materials for the structures and contact interfaces were imposed at the ilio-femoral (femur-pelvis), knee (femur-patella, patella-tibia, and fibula-tibia) and foot (tibia-feet) joints; defined using a penalty-based method with a weight factor, a coefficient of friction of 0.04 was chosen to be consistent. A distributed, on 70 nodes, vertical (y axes) load of 700 N was applied on the left side of the pelvis as depicted on Fig. 1(c), while a fixed constrain, of 200 nodes, was imposed at the lower extremity of the foot.⁹

3. Results

The present study aims to investigate the stress/strain distribution within complete model of the entire lower bony chain. With this purpose an experimental model and a complete 3D finite element one were loaded with 700 N. Both experimental and numerical models were fixed at feet. Obtained results reveal interesting consequences deriving by taking into account the complete bony chain.

3.1. Validation of results, strains comparison

Two optical transducers (x and z), focused on the left upper side of pelvis, and the LVDT transducer of the testing machine (y) were used to collect experimental data about complete model

Table 1

Comparison between experimental and numerical displacements along the three axes x, y, z.

	X _{Disp.} [mm]	Y _{Disp.} [mm]	Z _{Disp.} [mm]
Experimental model	0.72 [mm]	−1.61 [mm]	−9.85 [mm]
Numerical model	0.66 [mm]	−1.87 [mm]	−10.21 [mm]
Error	(8%)	(−16%)	(−3%)
Average error		(9.37%)

Fig. 2. Comparison between experimental and numerical data collected on the different six reference nodes.

displacements along the three axes. The obtained data were compared to the corresponding numerical ones, see Table 1. Results revealed a good agreement, the relative errors ranged from −3.65% (Dz) to −16.15% (Dy). The average error was 9.37%. In Fig. 2, are reported the measured longitudinal (i.e., along the y-axis) strains at full load, and the correspondent numerical strains computed at same locations, as well as the corresponding percentage error between estimated and experimental data. The average relative error between computed and measured strains was 9.37% ($n = 6$).

In Fig. 2, a histogram of the comparison between experimental and numerical values of strain is shown. The error indicated in figure ranges between 21 and 31% (average error 25.33%) and a good agreement can be deduced between experimental and numerical approaches. By analysing results numerical model seems generally to be subjected to higher strains, only SG#3 and SG#1 exhibit lower ones. In Fig. 3, strain registered on the SG#6 positioned under the neck of the femur is depicted. Obviously in this case the main solicitations is compressive and for this reason strains are negative, value reached by the experimental test on the gauge is −13 [μm/m], while the homologous numerical point reaches −16 [μm/m]. Curves showing behaviour of SG#5 and numerical point 5, SG#4 and point 4, located at distal and condylar areas of the femur, present the same increasing trend. Values of 353 and 430 [μm/m], respectively for experimental and numerical models, are reached in SG#5 and point 5, while values of 69 and 90 [μm/m] are reached in SG#4 and point 4. Lower values of strain measured in SG#4 and point 4 compared to the ones registered in SG#5 and point 5 can be explained by the spreading of forces which other bony and linking parts play each other. The effect of contact surfaces or elastic ligaments reduces concentrations of stress around the lower part of the femur. The areas around points 4 and 3 are located closely to the knee articulation and present opposite increasing trend of curves. As it is possible to notice, SG#3 reaches a higher value than point 3, thus experimental model seems to perform a more rigid configuration than the numerical one causing

Fig. 3. Experimental and numerical comparison of strain-load curves calculated for 6, 5, 4, 3, 2, 1 areas.

higher strains. The SG#2 and point 2 register significant values of strain, respectively 389 and 470 $\mu\text{m/m}$, while in SG#1 and point 1 values are of 140 and 105 $\mu\text{m/m}$. Also in this case numerical model reaches strain values lower than the experimental one. Data confirm flexural solicitations added to the compression imposed by the load ageing in the medial area of femur and tibia.

3.2. Stress shielding, bony displacements, and equivalent total strain on the model

In order to evaluate trending of displacements and stresses FE model was divided into three parts upper, concerning the pelvis, medium with femur upper ligaments, femur, knee ligaments, and patella, and finally the lower part concerning tibia, fibula, feet, and

Table 2

Comparison in terms of displacements and Eq. Von Mises Stress obtained data by different authors.

Femur and hip joint				
	Femur disp. at 350 N [Eq. (1)]	Error	Femur stress at 350 N [Eq. (2)]	Error
Rapperport ¹	–	–	8.75 [MPa]	(–28%)
Montanini (LC-2) ²	7.50 [mm]	(–7%)	17.51 [MPa]	(–43%)
Filardi ⁴	9.29 [mm]	(+15%)	13.38 [MPa]	(+9%)
Current	8.11 [mm]		12.22 [MPa]	
Meniscus				
	Meniscus disp. at 350 N [Eq. (1)]	Error	Meniscus stress at 350 N [Eq. (2)]	Error
Filardi ⁵	2.75 [mm]	(+2%)	1.38 [MPa]	(+5%)
Bendjaballah ⁶	–	–	1.16 [MPa]	(–12%)
Current	2.69 [mm]		1.32 [MPa]	
Tibial plateau				
	Tibial plateau disp. at 350 N [Eq. (1)]	Error	Tibial plateau stress at 350 N [Eq. (2)]	Error
Li ⁷	–	–	1.93 [MPa]	(–27%)
Vrahas ⁸	–	–	3.08 [MPa]	(–18%)
Current	2.07 [mm]		2.61 [MPa]	
Foot				
	Foot stress at 350 N [Eq. (1)]			Error
Filardi ⁴	4.12 [MPa]			(+2%)
Cheung ⁹	5.00 [MPa]			(18%)
Current	4.21 [MPa]

Fig. 4. Contour maps and quantitative representation of global displacements and Eq. Von Mises Stress versus Position (0–140) calculated for the upper part of the FE model (pelvis).

ligaments. Each part was sectioned by a crossing plane, as depicted in Figs. 4–6 and values of displacements and equivalent von mises stresses were calculated at regular intervals on the frontal surface. Fig. 4 shows a pick of about 10 mm on the top of pelvis (quote 0 mm) and a decreasing trend for the remaining part (0–140) till a value of about 7 mm. Different behaviour can be identified at the corresponding equivalent von mises stress which has its peak on the acetabulum with a value of about 11 MPa. Fig. 4, concerning the medium part, depicts the same trend of displacements with a maximum value of about 8 mm on the head of femur (quote 140 mm) till a value of about 3 mm (quote 540 mm) localised on the condilar zone. The maximum stress is located on the medial zone (quote 340 mm) and has a value of about 13 MPa. Finally the lower part reports separately data of displacements and stresses obtained for tibia and fibula which are shown in Fig. 5. As it is possible to notice fibula undergoes to sustain higher displacements on the lower part (quote from 640 to 890 mm) also if values are not worrying. Displacements ranging respectively from about 2.5 mm in tibia and 2.2 mm in fibula, at quote 540–590 mm. At quote 890–940 mm displacements values are 0.3 mm in tibia and 0.5 mm in fibula. Different conclusion can be deduced by comparing stresses, as depicted in Fig. 4, fibula appears much more unloaded than tibia with a maximum value of 2.3 MPa reached at quote 740 mm. On the corresponding zone also tibia reaches its maximum stress of about 13 MPa.

4. Discussion

4.1. Experimental and numerical strain field

Results confirm a bending effect, opposite to the compression load applied, present under the neck of the femur which determines a negative value of strain. Medial portions of femur and tibia manifested higher values of deformations, while lower and upper extremity result almost unloaded, because of tensions are probably concentrated on the contact surfaces. Although the validation was

restricted to only few sparse points on the femur surface and to a single load case, the obtained results provide evidence that it is possible to successfully predict the structural response.

4.2. Numerical stress and displacements

Comparison among results permits to evaluate if the multi body investigation approach, followed in this paper, obtains results too different from the traditional ones largely used in literature. The problem consists in comparing the different setup configurations adopted by authors. The obtained results reveal interesting consequences deriving by taking into account the complete bony chain. The stress shielding effects, displacements, and equivalent total strain induced in each part of the model are summarised in Table 2. As it is possible to notice the maximum total displacement is localised on the upper zone of the leg, concerning the pelvis and the femur, with values ranging from 10 to 8 mm. The lower part undergoes a minor effect producing displacements of the order of 2–3 mm localised in knee, tibia and fibula. Different consideration must be done for the Equivalent Von Mises Stress which does not follow a continuum trend like displacements. Maximum values of about 19 MPa are reached on the knee ligaments, while values of about 16 MPa were found in pelvis and fibula. Femur, femoral ligaments, and tibia register values of about 13 MPa, while patella, feet ligaments and feet are solicited with less than 8 MPa.

4.3. Comparison among different results

A comparison among the literature data and our model, carried on the works studying femur, knee, and tibia show generally less stressing conditions ageing on the structure, and moreover can furnish a complete vision of the global spreading of stresses along the various bony components. In order to carry on a useful comparison same simplification should be due. The elastic work expressed by Eq. (1), can be used in order to compare displacements by assuming a perfectly elastic behaviour of simulations,

Fig. 5. Contour maps and quantitative representation of global displacements and Eq. Von Mises Stress versus Position (140–540) calculated for the medium part of the FE model (femur, patella, ligaments).

thus a linear dependence is defined.

$$\mathbf{W}(J) = \mathbf{F}(N) * \mathbf{d} [\text{mm}] \quad (1)$$

Similarly by assuming a similar area where forces can spread within an analogue geometry, stresses can be compared by Eq. (2):

$$\mathbf{F}(N) = \frac{\sigma [\text{MPa}]}{\mathbf{A} [\text{mm}]} \quad (2)$$

Results, in terms of displacements and equivalent von mises stress, obtained by different authors at different loading conditions, and calculated by using Eqs. (1) and (2) are reported in Table 2.

Rapperport et al.¹ by finite element analysis for the human upright stance position evaluated pressure distributions across the articular surface for a resultant femoral load magnitude of 1.000 N and 40° angled on the medial of vertical. Results obtained by applying Eq. (2) compared with those ones presented in this paper show an error of –28% concerning the peak of Eq. Von Mises Stress. Montanini et al.² considered the effect of muscle action proposed by Bergmann et al.³ This load profile, which basically consists of four distinct muscle groups in addition to the hip joint force, is based on a validated musculoskeletal analysis and provides consistent torsion loading of the femur in addition to bending,

loaded on the head with a force of 980 N. Results obtained by applying Eq. (1) revealed an error of –7% in displacements and 43% for Eq. Von Mises Stress.

Filardi,⁴ in a precedent paper found a value of 9.29 mm for displacement in the femur and a Von Mises stress of about 14 MPa, by using different material properties.

Other results, Filardi,⁵ confirm a good agreement in displacements calculated in meniscus, error range around 2% for displacements and 5% for stresses.

Bendjaballah et al.⁶ considered a compression load of 1.300 N over the femur to measure compression stress on the meniscus in the external and internal periphery, respectively, for the healthy joint. Results, in this case, evidenced an error of –12% referred to the maximum Eq. Von Mises Stress.

Li et al.⁷ obtained a peak stress at the tibial plafond producing an error of –27%, calculated by imposing a load estimated in about 3% of the body weight. The peak stress, measured by Vrahas et al.,⁸ in cadaveric tibia statically loaded with 1360 N, produced an error of 18%.

Filardi⁴ found stresses ageing on the foot of 4.12 MPa with a good agreement with results herein presented. Cheung et al.⁹ developed a geometrical accurate 3D FE model of the human foot and ankle, and for a subject with body mass of 70 kg, a vertical

Fig. 6. Contour maps and quantitative representation of global displacements and Eq. Von Mises Stress versus Position (540–940) calculated for the lower part of the FE model (tibia, fibula, foot).

force of approximately 350 N is applied on each foot during balanced standing, producing in this case an error 18% in the peak of stress.

5. Conclusions

The aim of this study was to validate a complex FE model of the lower part of the skeletal: pelvis, femur, patella, tibia, fibula, feet, and ligaments through identical experimental models subjected to the same loading and constrain conditions. In a first phase numerical model was optimised comparing results, in terms of displacements and strains, obtained by the experimental one. In a second phase data obtained by FE analysis were collected to furnish a complete map of displacements, stresses and strains ageing on the bony elements. Results confirm a regular trend in displacements which decrease from the top to the bottom, maximum strains and Eq. von mises stresses are localised on the hinge, knee ligaments. Also femur and tibia appear quite solicited on the medial sections. The reason to realise this kind of investigation lies on the observation that not similar studies have been found in literature. Many authors focused their attention on the singular components of the skeletal: femur, knee, tibia, etc., and the goal of researches is often to assess how bony part can resist to a determinate load. For this reason often the standard testing configuration used consists of a fixed constrain at

the base and loading on the top. No interaction between bony parts is possible. The present research does not pretend to solve this problem but represent a first step for considering more than one or two parts in the researches in order to address test conditions towards physiological loads and constrain and to investigate their reciprocal interactions. Several limitations of this study have to be considered. First of all, the results were obtained without considering muscles contributions. Results refer to a static model of the full extension position of the knee joint, without considering how loads modify during walking, running or jumping. In spite these limitations, the obtained results do not deviate too much from the literature ones, demonstrating that subject specific FE models can predict the complex, non uniform stress and deformations fields that occur in biological bony complex chains.

Conflicts of interest

The authors have none to declare.

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