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Abstract-**The static position of single leg stance is a generally accepted worst case scenario for hip joint loading. However, it is essential in particular for investigations including temporal effects such as studies on fracture healing, fatigue, micromotion and remodeling to examine the dynamic loading situation not only at a single moment in time, but during the complete range of motion. In this study, a three-dimensional dynamic finite element model of the human femur during the gait cycle is developed. A temporally varying hip joint reaction force distribution during walking and jogging is employed and a temporally varying abductor muscles force is included while the distal end of the femur is constrained in translation only. The distribution of the displacements and stresses throughout the femur during different instants of the gait cycle is obtained in both loading conditions and compared. Also, an intramedullary prosthesis is nailed in the previous model and the femur-prosthesis is subjected to a similar type of loading. The results can be used to visualise the mechanical environment in the intact femur during dynamic loading and compare it to that after total hip arthroplasty. Such knowledge is vital for surgical procedures, healing processes as well as therapeutic regimes.**
Keywords – **Human femur modeling; Intramedullary prosthesis; Gait**

I. INTRODUCTION

The load-transfer mechanism from prosthesis to bone is one of the most important issues in a total hip arthroplasty (THA). Excessive stresses are responsible for fatigue failure of components or disruption of the implant/bone fixation [1]. Lower than normal stresses due to "stress-shielding" and adaptive bone remodeling may cause bone resorption around femoral hip stems with subsequent loosening, threatening the long-term integrity of the implant [2,3,4]. The introduction of wear debris into the bone-implant interface and the relative motion across the interface are other causes of loosening [5]. The cementless fixation of the implant employing direct bone-prosthesis contact offers an alternate technique in the attempt to overcome the problems associated with the cement. In all cases, the quality of the initial fit is critical since fixation requires intimate contact along the interface [6]. The study of stress patterns generated by the load-transfer mechanism and the understanding of their relationships with loading characteristics, prosthetic design, materials, and fixation characteristics, is critical in facing the issue of implant stability.

The effect of various types of realistic loading to the load transferred to the femur across a fixed bone-prosthesis interface has been the motivating reason for this study. A solution to the shielding problem would be a prosthesis which loads the femur in a manner as similar as possible to the natural state. Before validating a certain prosthesis design, a careful analysis of the stresses in an intact and an implanted bone should be made under a variety of as many realistic-simulating boundary conditions as possible.

In this paper a dynamic model of the human femur during the gait cycle was developed. A three-dimensional solid finite element model of an intact human femur was constructed assuming linear elastic, isotropic and homogeneous material properties. The geometry of the model was obtained from the three-dimensional reconstruction of the periosteal contours of transverse CT images of a human cadaver femur. The femoral head was then removed from the above model and a designed prosthesis was nailed inside the model, simulating a cementless case. An attempt was made to simulate a realistic distribution of the force exerted by the hip joint on the femoral head or the superior end of the prosthesis, respectively, and to include the force applied by the abductor muscles group on the femur. The distal end of the femur was constrained only in translation. A distinction of the bone material properties in the diaphysis and epiphyses was accounted for. Two loading conditions were considered -one for walking and one for jogging- based on published recorded kinematic data of the hip joint reaction force [7]. The stress and displacement distribution throughout the femur and the femur-implant system was obtained and compared in both loading cases. The results are indicative of the fact that three-dimensional finite element models employing realistic loading conditions could be potentially useful in the mapping of the mechanical environment in bones, with subsequent practical applications in the improvement of prosthesis design and implementation of therapeutic methods.

II. METHODOLOGY

A. *Gait analysis*

The gait cycle is defined as the time interval between two successive occurrences of one of the repetitive events of walking and is the basic measurement unit in gait analysis. The gait cycle begins when one foot comes in contact with the ground and ends when the same foot contacts the ground again [8]. During gait, important movements occur in all three planes – sagittal, frontal and transverse. The largest movements occur in the saggital plane. Saggital plane hip motion is essentially sinusoidal in walking. The loads on the hip joint during dynamic activities have been studied by several investigators. The force vector acting on the femoral head depends on the external forces acting on the limb and the internal forces primarily generated by muscle contraction. In general, the resultant force at the hip joint during walking reaches an initial peak in the early stance phase and a second peak in the late stance phase, as described in [9,10]. Its magnitude ranges from 0.5 to 4 times the body weight depending on the instant of the cycle period. In jogging, only the first peak is prominent with a higher magnitude than the corresponding peak in walking. As far as the muscles activity during gait is concerned, during stance the primary muscles controlling the hip are the extensors and abductors, in swing it is the flexors while the adductors tend to participate during the intervals of exchange between swing and stance. In the present model, the variation of the hip joint reaction force acting on the femoral head throughout the walking and jogging cycle has been employed. Also, the variation of the abductor muscles force acting on the major trochanter during the gait cycle has been used.

B. Finite element model

The mesh generation relied upon three-dimensional (3-D) surfaces resembling the periosteal geometry of a femur. These surfaces were obtained by 3-D reconstruction of 2mm thickness transverse CT slices of a human femoral bone. The length of the femur was about 35 cm and the diameter of the femoral head around 4 cm. A 3-D finite element model was generated using the Patran 8.5 Software (MSC Software Corporation). A solid mesh of the intact femur was constructed consisting of 5584 tetrahedral elements and 1393 nodes. The global element edge length was 7.5 mm. In the model of the nailed femur were employed 531 tetrahedral elements for the nail and 5124 tetrahedral elements for the bone, resulting in a total of 5124 elements and 1099 nodes, after equivalencing.

Along the femoral shaft, the contact between bone and the prosthesis was modelled by means of surface elements. The inner surface of the femoral shaft was meshed with 3101 triangular elements of global edge length 2 mm and 1946 nodes. Friction between bone and the prosthesis was neglected.

C. Material properties

The femur was assumed to behave as a linear elastic, isotropic and homogeneous material. The material properties were distinguished between two principal regions, namely, trabecular bone at the epiphyses and cortical bone in the diaphysis [11] (Table I). The values of the assigned cortical bone properties are lower than their average values in human

TABLE I

MATERIAL PROPERTIES FOR BONE AND PROSTHESIS

Property	Cortical Bone[11]	Trabecular Bone[11]	Titanium Alloy [12]
Young's modulus	15 GPa	0.7 GPa	110 GPa
Poisson's ratio	0.33	0.2	0.3
Density	1650 kg/m^3		620 kg/m^3

cortical femoral bone. This was done in order to compensate for the lack of existence of the trabecular bone portion present in the diaphysis and of the medullary canal. The titanium alloy used for the prosthesis material was assumed to be linear elastic, isotropic and homogeneous (Table I).

D. Boundary conditions

Both hip-joint contact and muscle forces were applied quasi-statically to represent the gait cycle. Force and muscle data were derived from the literature [7, 13, 14, 15]. The magnitude and orientation of the force and muscle components were defined with respect to a right-handed cartesian coordinate system for the right femur with its origin at the center of the femoral head. The +x-axis pointed medially, the +y-axis posteriorly and the +z-axis was directed superiorly. The resultant hip joint reaction force was applied on the near-hemispherical surface of the acetabulum-femoral head contact area of the intact femur. In the implanted femur the resultant force was applied similarly on the top of the prosthesis neck. The body weight was taken as 75 kg.

The temporal variation of the resultant force throughout the gait cycle in walking and jogging has been obtained from [7]. The muscle force attachment data were taken from [13, 14] and the temporal variation of the muscles involved throughout gait was based on data presented in [15]. The same set of muscle data was used in the jogging condition as well. The employed force and muscle loading data at two different instants of the walking and jogging cycle appear in Table II. The distal end of the femur, at the location of the knee joint was constrained in translation.

TABLE II

LOADING CONDITIONS IN WALKING AND JOGGING

Heel strike	Walking	Jogging
Hip joint contact force	[-0.36, -0.10, -0.93] [1]	[-0.2, -0.10, -2]
Abductor muscle force	[0.66, 0.45, 0.61]	
Midstance		
Hip joint contact force	[-0.35, 0.05, -0.93]	
Abductor muscle force	[0.67, 0.18, 0.72]	
Toe-off		
Hip joint contact force	[-0.31, -0.10, -0.95]	[-0.8, -0.25, -0.13]
Abductor muscle force	[0.71, 0.00, 0.70]	

[1] The numbers in the brackets are multipliers of the body weight (BW)

E. Calculation

The analysis was done using Nastran (Nastran 70.5 MSC Software Corporation).

III. Results

The Von Mises stress distribution in the intact femur during walking and jogging is shown in Figs. 1 and 2. Specifically, Figs. 1(a) and 1(b) show the distribution of the Von Mises stress during walking, at heel strike and midstance respectively. Figs. 2(a) and 2(b) show the corresponding distribution at toe-off in walking and jogging respectively. The Von Mises stress distribution in the nailed femur at toe-off during walking and jogging is shown in Figs. 3(a) and 3(b) respectively. The range bars are numbered in Patran units. Fig. 4 shows the variation of the Von Mises stress at heel-strike in walking along a line in the lateral surface of the nailed femoral diaphysis which is in contact with the prosthesis. The line is defined by the position of certain nodes and the numbering has an inferior-superior direction with respect to the long axis of the femur. The Von Mises stress at the corresponding locations in the intact femur appears in the same graph.

IV. Discussion

In walking, the maximum higher stress value appears at heel-strike in the intact femur, followed by the toe-off and midstance phases as seen in Figs. 1 and 2(a). The lower part of the diaphysis exhibits the higher stresses with the largest variation between the different instants of the cycle.

In jogging, the higher stresses in the diaphysis are found in the anterior side of the femur, at toe-off (Fig. 2(b)), whereas in walking the medial and lateral sides of the diaphysis are more heavily loaded. The nailed femur at toe-off seems to be overloaded in jogging, compared to the intact femur (Fig. 3(b)). The lower part of the diaphysis is more heavily stressed in jogging than in walking (Fig. 3(a)).

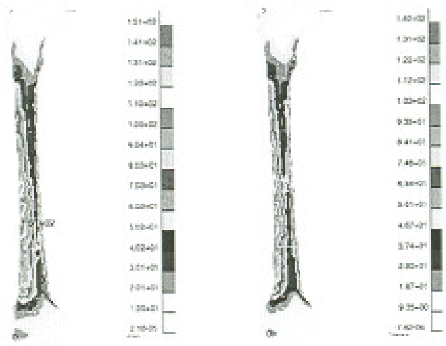
(a) (b)

Fig. 2. Von Mises stress distribution in the intact femur at toe-off in (a) walking, and (b) in jogging

The lack of existence of a medullary canal makes the femur appearing stiffer than it actually is, as depicted by the excessively "hot" areas in the diaphysis in comparison to the upper epiphysis.

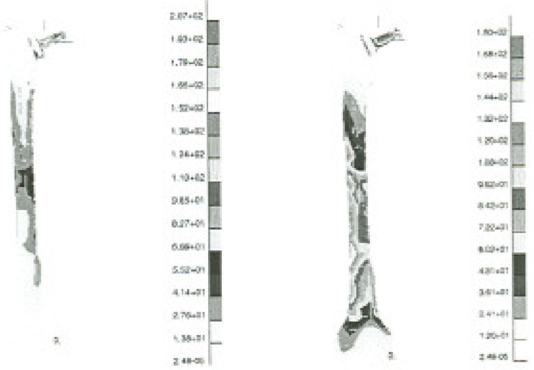
The stress situation in the bone-prosthesis interface appears in Fig. 4. The stress magnitude generally drops when moving along the inferior-superior direction. It is interesting to note that the nailed femur seems to be stress-shielded initially-along the lower part of the stem- and overloaded after a certain position along the z-axis in comparison to the intact one.

It should be stressed out that the results presented here are indicative of the method followed and should be looked at from a qualitative point of view. Further improvement of the present model would potentially contribute to the quantification of the mechanical environment under various loading conditions in the intact and implanted femur models.



(a) (b)

Fig. 1. Von Mises stress distribution in the intact femur at (a) heel-strike, and (b) mid-stance in walking



(a) (b)

Fig. 3. Von Mises stress distribution in the nailed femur at toe-off in (a) walking, and (b) in jogging

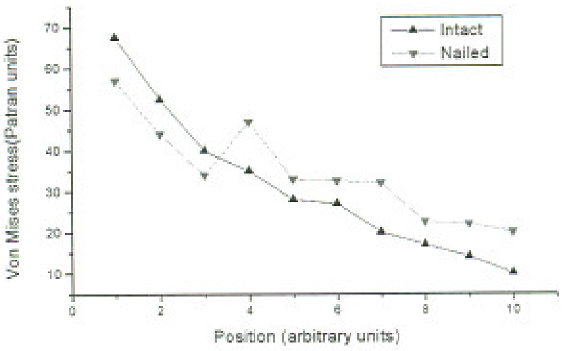


Fig. 4. Von Mises stress at heel-strike in walking

V. Conclusion

The complexity of human locomotion makes it difficult to develop a realistic and reliable dynamic model of the human femur during the gait cycle. Certain difficulties are encountered when trying to model femoral movement. First, the three-dimensional movement of the hip during gait requires the knowledge of three components of the hip joint reaction force and their temporal variation throughout the gait cycle: this is translated as the need for measured experimental data in a given system of coordinates during the cycle period. Second, the actual contact area of the acetabulum and the femoral head and the way the joint reaction force is distributed on that area must be known in order to have a realistic description. Third, there is a variety of different muscle groups attached to the femur at various locations and acting at different periods during the gait cycle. This imposes the need for experimental data and knowledge of the specific attachment sites. Finally, the simulation of the exact movement of the knee joint during gait is necessary before a realistic dynamic model can be constructed.

The present model should be looked at as a preliminary attempt to develop a dynamic finite element model of the femur during gait. Our results should be viewed in a comparative sense only between the two loading conditions of walking and jogging. An extension could be made to include other types of loading as well. The development of a reliable dynamic model of the femur, which is carefully tested and validated, would lead to a proper understanding of the biomechanical situation of the lower extremity. Once such a model is obtained and the mechanical environment around a certain type of prosthesis is correctly quantified for a variety of physical activities, it would become meaningful to look at factors such as the prosthetic design, materials, and fixation characteristics.

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