# A DYNAMIC FINITE ELEMENT MODEL OF THE HUMAN FEMUR DURING WALKING AND JOGGING

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## 1. SUMMARY

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 at 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 presented. A temporally varying hip joint reaction force distribution during walking and jogging is employed and a constant point abductor muscles force is included while the distal end of the femur is fully constrained. 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. The temporal variation of the Von Mises stress in the diaphysis during the walking and jogging cycle period is obtained. The results can be used to visualise the mechanical environment in the femur during dynamic loading, which is vital for surgical procedures, healing processes as well as therapeutic regimes.

## 2. INTRODUCTION

The finite element method (FEM) was introduced in orthopaedic biomechanics in 1972 [1] to evaluate stresses in human bones. The early FEM was adopted for stress analysis of the femur by Brekelmans *et al.* [2] and Rybicki *et al.* [3]. Both of them used 2-D plane-stress elements of uniform thickness, although Rybicki *et al.* analyzing the proximal part of the bone only, accounted for non-uniform thickness by varying the Young's modulus of the elements. A comparable model was used by Wood *et al.* [4] who used 2-D elements of variable thickness. Early 3-D FEM models of the femur were those of Scholten [5] and Olofsson [6] followed by Valliappan *et al.* [7], Harris *et al.* [8] and Rohlmann *et al.* [9]. These publications were all method-oriented, demonstrating the problems associated with data manipulation, interpretation and representation. Some models contained a high degree of sophistication, including realistic non-homogeneity of trabecular bone based on experimental data [5, 9].

Since then, FEM has been applied with increasing frequency on problem-oriented studies pertaining to bone growth and remodeling, problems of fracture and fracture fixation, stress analyses of bone-prosthesis structures, artificial joint design and fixation, total hip replacement (THR), stress-shielding and others [10]. The aim of those investigations has been mainly to assess relationships between load carrying functions and morphology of the tissues, and to optimize designs and fixation techniques of implants.

Three-dimensional finite element stress analyses of the femur with endoprosthesis were performed by among others Rohrle *et al.* [11], Tarr *et al.* [12], Huiskes [13], Crowninshield *et al.* [14], Prendergast, P. J. and Taylor, D. [15] and Rohlmann A. *et al.* [16]. In most of these models the geometry and the distribution of material data of the femur were much simplified.

Few authors have correlated theoretical and experimental results in detail. Huiskes *et al.* [17], through detailed experimental and theoretical analyses on bilateral bones, showed that the femoral shaft can be modeled rather refined when the cortical bone is assumed to be transversely isotropic, although local inhomogeneities produce deviations to a certain degree in the experiment. The FE model's validity by verification with experimental work has also been investigated by Rohlmann *et al.* [16] and Keyak *et al.* [18].

The choice of forces which have been applied to the femur in most finite element studies in the literature has been somewhat arbitrary, with a large variation between different authors. The majority of the studies consider the femur to be simplistically loaded with usually only a point hip joint reaction force applied to the femoral head. Some experimental and analytical investigations consider only one or two abductors [19, 3, 20]. Only a few finite element analyses have developed more sophisticated load configurations, with the attachment of other muscle groups [21, 22]. A major limitation of most studies is that the stress distribution has only been evaluated at one instant of time, during the stance phase of gait.

The present paper aims in developing a dynamic model of the human femur during the gait cycle. A three-dimensional solid model of the human femur has been constructed with assumed homogeneous, isotropic elastic material properties. The reference geometry of the model has been obtained from the standardized femur program proposal by Viceconti *et al.* [23]. An attempt has been made to simulate a realistic distribution of the forces exerted by the hip joint on the femoral head and the inclusion of the abductor muscles force acting on the major trochanter. The distal end of the femur is fully constrained. Two loading conditions are considered: one for walking and one for jogging based on published recorded kinematic data of the hip joint reaction force [24]. The stress and displacement distribution throughout the femur has been obtained and compared in both loading cases. Also, the temporal variation of stress in the diaphysis is obtained. The results are indicative of the fact that three dimensional finite element models employing realistic loading conditions are potentially useful in the mapping of the mechanical environment in bones, with subsequent practical applications in the improvement of orthopaedic design and implementation of therapeutic methods.

## 3. MATERIALS AND METHODS

### 3.1 Gait Analysis

Walking uses a repetitious sequence of limb motion to move the body forward while simultaneously maintaining stance stability [25]. 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 unit of measurement 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 (Fig. 1).
These moments in time are referred to as initial contact. The stance phase, which is also called the support phase or contact phase, lasts from initial contact to toe off, that is when the foot is no longer in contact with the ground. Toe off marks the beginning of the swing phase of the gait cycle, which lasts from toe off to the next initial contact of the same limb. Each of

these phases for walking can be further subdivided [26]. In each gait cycle there are two periods of double support when both feet are on the ground and two periods of single support. The stance phase usually lasts about 60% of the cycle, the swing phase about 40%, and each period of double support about 10%, but this may vary with the walking speed. The gait cycle period is called the cycle time or stride time and for normal subjects ranges from 0.54 to 1.25 sec depending on the age and sex [26]. In this study, the cycle time for walking has been taken as 1 sec and that for jogging as 0.652 sec [24].

During gait, important movements occur in all three planes – sagittal, frontal and transverse. The largest movements occur in the sagittal plane. Sagittal plane hip motion is essentially sinusoidal in walking. The loads on the hip joint during dynamic activities have been studied by several investigators [24, 27-30]. 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. Hip joint forces have been estimated indirectly using inverse dynamics and analytical methods or directly measured with implanted transducers.

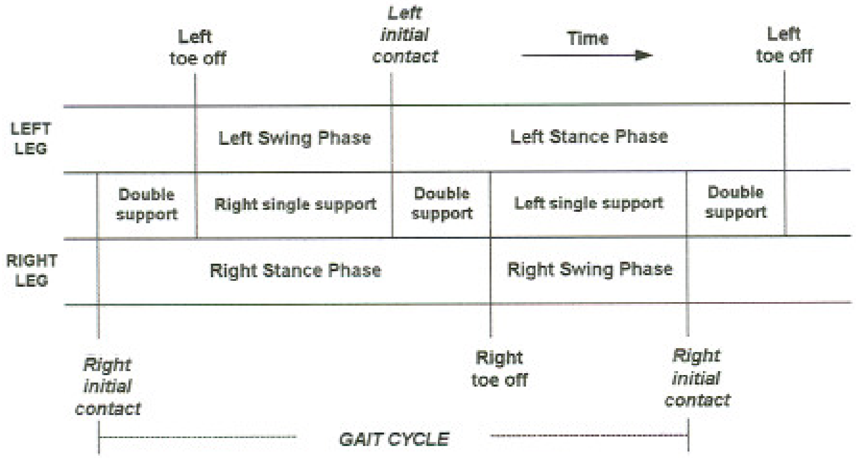
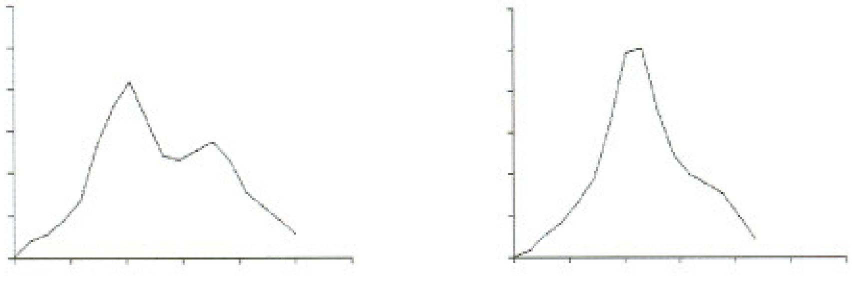



**Figure 1:** Description of the human gait cycle

In general, the resultant force at the hip joint during walking reaches an initial peak in early stance phase and a second peak in late stance phase (Fig. 2a), as described in the initial work of Paul [27]. Its magnitude ranges from 0.5 to 4 times the body weight depending on the instant of the cycle period. In jogging (Fig. 2b), only the first peak is prominent with a higher magnitude than the corresponding peak in walking.



(a) (b)

**Figure 2:** Resultant reaction force versus time at the hip joint during (a) walking and (b) jogging.Thegraphs are based on measured experimental data by Bergmann *et al.* [24].

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. Action of the hip extensor muscles occur from late mid swing through the loading response. The hip abductors function during the initial half of stance. Flexor muscle action begins in late terminal stance and continues through initial swing into early mid-swing. The actions of the major adductor muscles are either as a hip flexor or as an extensor. In the present model only the abductor muscles action has been considered.

### 3.2 FE model

Measures of the Von Mises stress and translational diplacement distributions all over the femur and the temporal variation of the Von Mises stress tensor in a node in the diaphysis during walking and jogging were determined using three-dimensional FEM computations.

The bone model geometry was based on the periosteal contours of the Standardized Femur developed by Viceconti *et al.* [23]. A solid mesh of tetrahedral elements was constructed consisting of 4998 elements and 2258 nodes (Patran 8.5, MSC.Software Corporation). To keep the overall number of elements low, the element edge length was selected to be 10 mm. A solution was obtained using the Nastran package (Nastran 70.5 MSC.Software Corporation) on a 16 CPU Origin 2000 Silicon Graphics computer.

The bone was assumed to behave as a homogeneous, isotropic, elastic material with properties given in Table 1.

**Table 1:** Material properties for bone

| Property | Value |
|---|---|
| Young's modulus | $15\times10^9$ N/m$^2$ |
| Shear modulus | $5.639\times10^9$ N/m$^2$ |
| Poisson's ratio | 0.33 |
| Density | 1650 kg/m$^3$ |

The magnitude and orientation of the force and muscle components were defined with respect to a right-handed Cartesian coordinate system for the left femur with its origin at the center of the femoral head. The axis +z points upwards, the axis +x points medially and the axis +y is directed anteriorly. The resultant hip joint reaction force was applied as pressure distribution on the near-hemispherical surface of the acetabulum-femoral head contact area at a mean frontal angle (angle formed between the direction of the resultant force and the z-axis) of 30°. The temporal variation of the resultant force throughout the gait cycle in walking and jogging has been obtained from [24]. The body weight was taken as 75kg. The abductor muscles force was based on data used in [22] and was applied as a point force at the location of the major trochanter and kept constant throughout the gait cycle. The diaphysis was fully constrained at its distal end, that is the knee joint motion was fully prohibited.

The finite element analysis was performed for the two loading cases (Table 2): walking and jogging with inclusion of the abductor muscles. A transient response type of analysis was employed.

Table 2: Load cases

| Load case type | Hip joint reaction force | Abductor muscles force |
|---|---|---|
| 1 Walking | *Temporal variation*: walking cycle [24]<br>*Spatial distribution*: upper hemisherical surface of femoral head | *Steady:* {-430 N, 0, 1160 N} [22]<br>*Spatial location:* node position at the site of the major trochanter |
| 2. Jogging | *Temporal variation*: jogging cycle [24]<br>*Spatial distribution*: upper hemisherical surface of femoral head | *Steady:* {-430 N, 0, 1160 N} [22]<br>*Spatial location:* node position at the site of the major trochanter |

## 4. RESULTS

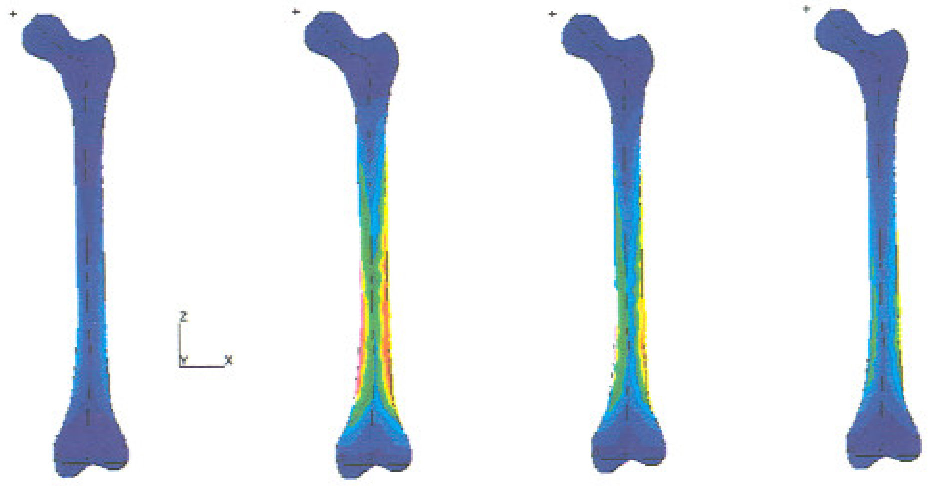
The Von Mises stress distribution during walking and jogging are shown in Figs. 3 and 4 respectively. The same range bar has been selected for Figs. 3 and 4, so that a comparison is feasible. The laboratory coordinate system is shown. Figures 3(a) to 3(d) show the distribution of the Von Mises stress during walking, at heel strike, first and second peak and toe-off, respectively. Figures 4(a) to 4(c) show the corresponding distribution for jogging, at heel strike, peak and toe-off respectively. As expected, the higher stresses in walking occur during the first peak of the applied hip joint reaction force, followed by the ones during the second peak. As can be seen the lower part of the diaphysis exhibits the higher stresses with the largest variation between different instants of the cycle.

In jogging, the greatest stresses appear during the peak of the hip joint reaction force. The higher stresses in the diaphysis in comparison to walking are obvious. At toe-off , the distribution is nearly similar to the corresponding one in walking.

Figure 5(a) shows the temporal variation of the Von Mises stress tensor at a node location in the diaphysis during walking. The pattern of the hip joint force distribution is evident. A corresponding graph at the same node location is plotted for jogging in Fig. 5(b). The higher stress during the peak in jogging is noted.

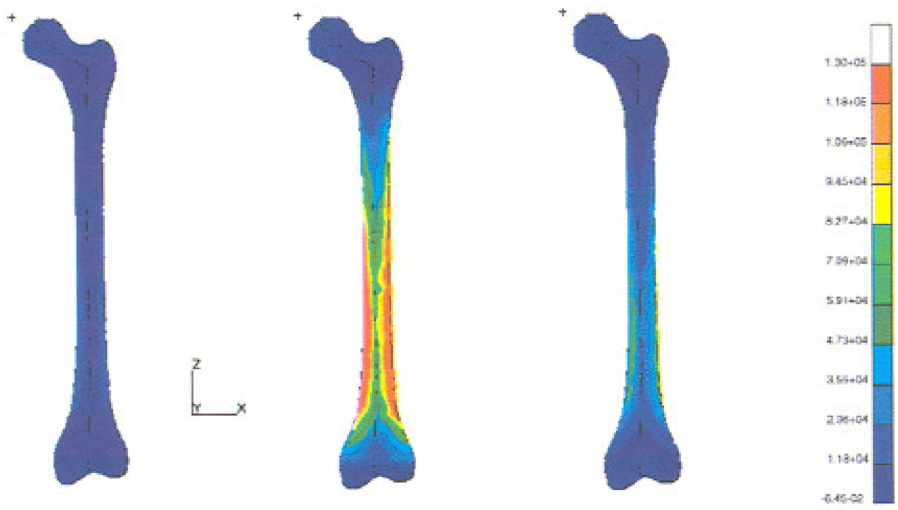
The translational displacements distribution generated by the two load cases can be seen in Figs. 6 and 7 respectively. Figures 6(a) to 6(d) show the distribution of the magnitude of translational displacements during walking, at heel strike, first and second peak and toe-off, respectively. Figures 7(a) to 7(c) show the corresponding distribution for jogging, at heel strike, peak and toe-off respectively. As can be seen the greatest displacements in walking occur during the first peak of the gait cycle in walking, followed by the ones during the second peak. The head of the femur, femoral neck and upper part of the diaphysis exhibit the higher displacements with the largest variation between different instants of the cycle. At toe-off, the diplacements on the head of the femur are 1.5 to 2 times as large as the ones at heel strike.

In jogging, the greatest displacements occur during the peak of the cycle and affect more the upper part of the diaphysis than in walking. At toe-off , the distribution is similar to the one at toe-off in walking. At heel strike, a lower distribution appears on the femoral head and neck which gradually annihilates along the diaphysis.



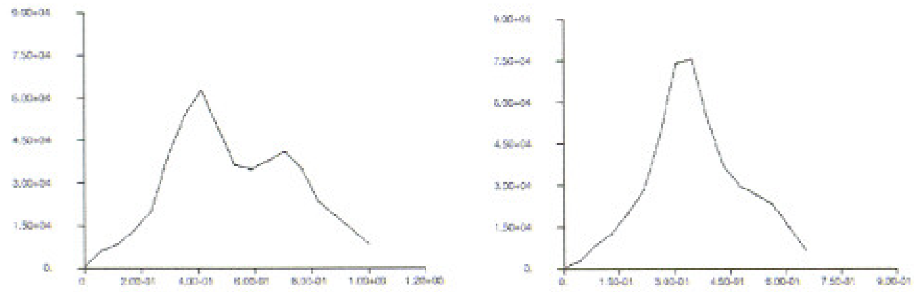

(a) (b) (c) (d)

**Figure 3:** Von Mises stress distribution during walking at (a) heel strike (t=0.17 sec), (b) first peak (t=0.41sec), (c) second peak (t=0.71 sec) and (d) toe-off (t=0.82 sec)



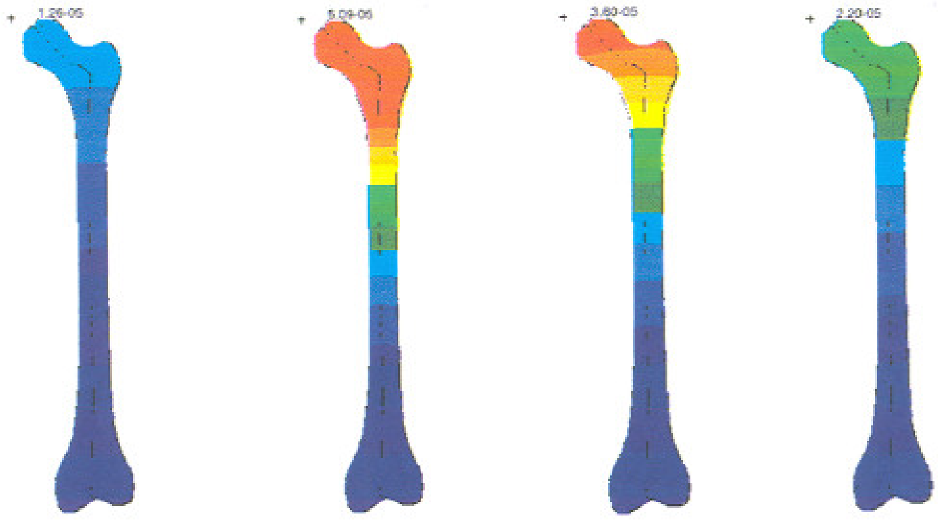

(a) (b) (c)

**Figure 4:** Von Mises stress distribution during jogging at (a) heel strike (t=0.08 sec), (b) peak (t=0.30sec) and (c) toe-off (t=0.56 sec)



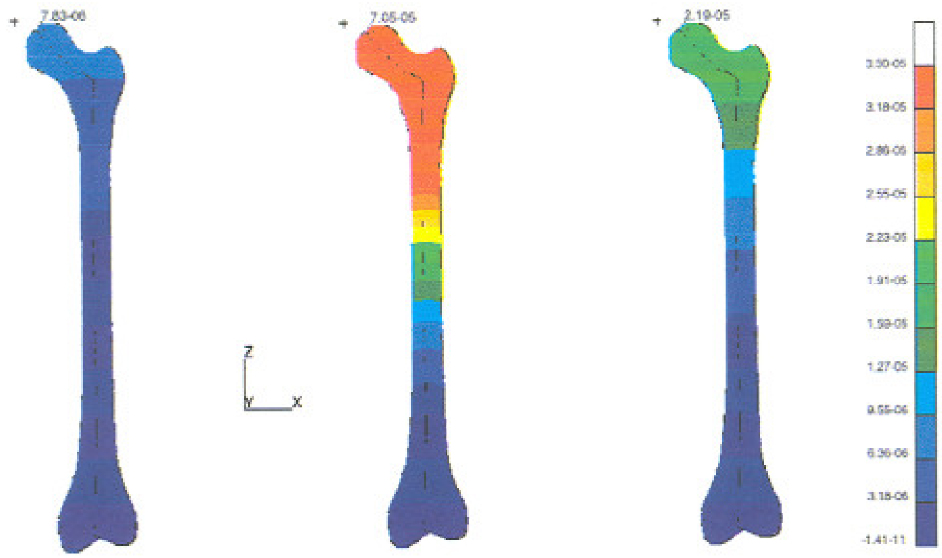
(a) (b)

**Figure 5:** Von Mises stress versus time at a point in the diaphysis during (a) walking and (b) jogging



(a) (b) (c) (d)

**Figure 6:** Distribution of the magnitude of translational diplacements during walking at: (a) heelstrike (t=0.17 sec), (b) first peak (t=0.41 sec), (c) second peak (t=0.71 sec) and (d) toe-off (t=0.82 sec).



(a) (b) (c)

**Figure 7:** Distribution of the magnitude of translational diplacements during walking at: (a) heelstrike (t=0.17 sec), (b) first peak (t=0.41 sec), (c) second peak (t=0.71 sec) and (d) toe-off (t=0.82 sec).

## 5. DISCUSSION

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:

(i) 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.

(ii) 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.
(iii) 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 poses the need for experimental data and knowledge of the specific attachment sites.
(iv) The simulation of the exact movement of the knee joint during gait is necessary before a realistic dynamic model can be constructed.

The above limitations may explain the fact that the bulk of finite element models appearing in the literature are static, usually referring to certain instants of the stance phase of gait.
In the present study an attempt has been made to develop a three-dimensional dynamic model of the human femur during the gait cycle in walking and jogging. Using the geometry of the Standardized Femur by Viceconti *et al.*, the joint reaction force data by Bergmann *et al.,* the muscle force components by Taylor *et al.*, and the simplification of a homogeneous, isotropic, elastic material description, a dynamic solid model of the human femur during walking and jogging was developed.

The temporal variation of the resultant hip joint reaction force during walking and jogging based on measured experimental data should be considered an advantage of our model. The assumption for the spatial distribution of the applied force on a hemispherical surface on the proximal-medial part of the femoral head, should be considered realistic enough. Although the direction of the applied pressure is always normal to the elements and the change in the direction of the applied resultant force is not taken into account throughout gait, this should not be considered a problem at least, as far as the frontal plane motion is concerned. The frontal angle averagely remains constant and around 30° during a gait cycle. It is the transverse angle that changes sign during the cycle period in an almost sinusoidal form. In that sense, our model could be thought of as quite reliable in a 2D aspect.

The inclusion of the abductor muscles force should be considered only for the stance phase of gait when these muscles act. The change of their direction of action, and their magnitude throughout the stance period has not been accounted for. No other muscle groups acting during the cycle period have been included in the model. A further limitation of the model lies in that the knee joint is considered fully constrained both in walking and jogging.

The assumption of average homogeneous material properties throughout the femur and the lack of existence of the medullary canal makes the femur appearing stiffer than it actually is. This can be easily realised when considering the distribution of cancellous and cortical bone in the femur. The bulk of the epiphyseal bone is cancellous and the opposite holds for the diaphysis. Therefore, it is not surprising that the stress distribution pattern appears quite different than the one expected, exhibiting "hot" areas in the diaphysis, when one would expect the proximal femur to be more heavily stressed.

Several alterations could be made to the present model to achieve a more realistic simulation, such as improvement of the geometry, assignment of different material properties to different parts of the femur, inclusion of a temporal variation of the muscle force during gait and inclusion of more muscle groups.

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 cases of walking and jogging. 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 and its use as a predictive model to investigate the effect of changed conditions.

## 6. REFERENCES


[1] Huiskes, R. and Chao, E. Y. S., A survey of finite element analysis in orthopedic biomechanics: the first decade. *J. Biomechanics*, **16(6)**, 385-409, 1983.

[2] Brekelmans, W. A. M., Poort, H. W. and Sloof, T. J. J. H., A new method to analyse the mechanical behaviour of skeletal parts. *Acta Orthop. Scand.* **43**, 301-317, 1972.

[3] Rybicki, E. F., Simonen, F. A. and Weis, E. B., On the mathematical analysis of stress in the human femur. *J. Biomechanics* **5**, 203-215, 1972.

[4] Wood, R., Valliappan, S. and Svensson, N. L., Stress analysis of human femur. *Theory and Practice in FEM Structural Analysis*, Tokyo Seminar (edited by Yamada, Y. and Gallagher, R. H.), pp. 461-478. University of Tokyo Press, Tokyo.

[5] Scholten, R., Ueber die Berechnung der mechanischen Beanspruchung in Knochenstrukturen mittels fuer den Flugzeugbau entwickelter Rechenverfahren. *Med. Orthop. Technik* **6**, 130-138, 1975.

[6] Olofsson, H., Three dimensional FEM calculation of elastic stress field in human femur.Dissertation, Institute of Technology, University of Uppsala, Sweden, 1976.

[7] Valliappan, S., Svensson, N. L. and Wood, R. D., Three dimensional stress analysis of the human femur. *Comput. Biol. Med.* **7**, 253-264, 1977.

[8] Harris, L. J., Chao, R., Bloch, R. and Weingarten, V., A three-dimensional finite element analysis of the proximal third of the femur. *Proceedings of 24$^{th}$ Annual Meeting of Orthopaedic Research Society*, p. 16. ORS. Chicago, 1978.

[9] Rohlmann, A., Bergmann, G. and Koelbel, R., The relevance of stress computation in the femur with and without endoprostheses. *Finite Elements in Biomechanics* (Edited by Gallagher, R. H., Simon, B. R., Johnson, P. C. and Gross, J. F.), pp. 361-377, JohnWiley, New York, 1982.

[10] Mackerle, J., A finite element bibliography for biomechanics (1987-1997), *Appl Mech Rev*, **51(10)**, 587-634, 1998.

[11] Rohrle, H., Scholten, R., Sollbach, W., Ritter, G. and Gruenert, A., Der Kraftfluss bei Huftendoprothesen. *Arch. Orthop. Unfallchir*, **89**, 49-60, 1977.

[12] Tarr, R., Lewis, J., Ghassemi, F., Sarmiento, A., Clarke, I. and Weingarten, V., Anatomic three-dimensional finite element model of the proximal femur with total hip prosthesis. *International Conference Proceedings, Finite Elements in Biomechanics*, (Edited by Simon, B. R.), Vol. 2, pp. 511-525. The University of Arizona, Tuscon, 1980.

[13] Huiskes, R., Some fundamental aspects of human joint replacement. *Acta orthop. Scand.* Suppl. No. 185, 109-199, 1980.

[14] Crowninshield, R. D., Brand, R. A., Johnson, R. C. and Milroy, J. C., An analysis of femoral component stem design in total hip arthroplasty. *J. Bone Jt Surg.* **62-A**, 68-78, 1980.

[15] Prendergast, P. J. and Taylor, D., Stress analysis of the proximo-medial femur after total hip replacement. *J. Biomed. Engng* **12**, 379-382, 1990.

[16] Rohlmann, A., Mössner, U. and Bergmann, G., Finite-element-analysis and experimental investigation in a femur with hip endoprosthesis, *J. Biomechanics*, **16(9)**, 727-742, 1983.

[17] Huiskes, R. *et al.*, A detailed comparison of experimental and theoretical stress analyses of a human femur. *Mechanical Properties of Bone* (edited by Cowin, S.) AMD-Vol. 45, pp. 211-234. American Society of Mechanical Engineers, New York, 1981a.

[18] Keyak, J. H. *et al.*, Validation of an automated method of three-dimensional finite element modelling of bone. *J. Biomed. Engng* **15**, 505-509, 1993.

[19] Ghista, D. N., Toridis, T. G. and Srinivasan, T. M., Human gait analysis: determination of instantaneous resultant joint forces, muscle forces and the stress distribution in bone segments. Part II. *Biomedizinische Technik (Berlin)* **21**, 66-74, 1976.

[20] Toridis, T. G., Stress analysis of the femur. *J Biomechanics* **2**, 163-174, 1969.

[21] Cristofolini, L., Viceconti, M., Toni, A. and Guinti, A., Influence of thigh muscles on the axial strain in a proximal femur during early stance in gait. *J Biomechanics* **28**, 617-624, 1995.

[22] Taylor, M. E., Tanner, K. E., Freeman, M. A. R. and Yettram, A. L., Stress and strain distribution within the intact femur: compression or bending?. *Med. Eng. Phys.* **18-2**, 122-131, 1996.

[23] Viceconti, M., Casali, M., Massari, B., Cristofolini, L., Bassini, S. and Toni, A., The 'Standardized Femur Program' proposal for a reference geometry to be used for the creation of finite element models of the femur, *J. Biomechanics*, **29(9)**, 1241, 1996.

[24] Bergmann, G. *et al.*, Hip joint loading during walking and running, measured in two patients, *J. Biomechanics* **26(8)**, 969-990, 1993.

[25] Perry J., *Gait Analysis, Normal and pathological Function*, McGraw-Hill, 1992.

[26] Whittle, M. W., *Gait Analysis, an Introduction,* Butterworth-Heinemann, Second Edition, Oxford, 1996.

[27] Paul, J. P., Comparison of EMG signals from leg muscles with the corresponding force actions calculated from walk path measurements. *Human Locomotor Eng* **48A**, 6-26, 1971.

[28] Paul, J. P., McGrouther, D. A., Forces transmitted at the hip and knee joint of normal and disabled persons during a range of activities. *Acta Orthop Belgica* **41**(Suppl.):78-88, 1975.

[29] Seireg, A., Arvikar, R. J., The prediction of muscular load sharing and joint forces in the lower extremities during walking, *J Biomechanics*, **8**, 89-102, 1975.

[30] Brand, R. A., Pederson, D. R., Davy, D. T., Kotzar, G. M., Heiple, K. G., Goldberg, V. M., Comparison of hip force calculation and measurements in the same patient, *J. Arthroplasty*, **9(1)**, 45-51, 1994.