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The possibility of failure prediction in hip and knee prosthesis by using a damage estimator

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ABSTRACT

Finite element simulatios were performed to dynamically determinate the area of the contact, the contact pressure and the strain energy density (identified as a damage function) for three different activities – normal walking, ascending stairs and, descending stairs – that could be considered to define the level of the activity of the patient. The finite element model uses a modern contact mechanism that includes friction between the metallic femoral condyles or femoral head (considered rigid) and the tibial polyethylene insert or acetabular cup respectively, considering a non-linear behaviour). For all three activities, the finite element analyses were performed, and a damage score was computed. Finally, a cumulative damage score (that accounts for all three activities) was determined and the areas where the fatigue wear is likely to occur were identified. A closer look at the distribution of the damage score reveals that the maximum damage is likely to occur not at the contact surface, but in the subsurface.

Keywords: Total hip prosthesis, acetabular cup, total knee prosthesis, tibial tray insert, wear, FEM, experimental methods.

1. INTRODUCTION

Human locomotion is a natural process that is mechanically based on the relative movement between rigid levers – the skeletal bones – as the result of the movement of passive and active effectors – the muscles. Relative motion is allowed by the joints, which also play an important role in transferring the load from one bone segment to another. The load is transferred by a contact-based mechanism that involves friction and wearing. Replacement of a natural joint with an artificial one affects the load transfer mechanism, and the components of the prosthesis need to allow the relative movements representing the entire range of human activity over a long period. If certain activities are prohibited or severely limited, the level of patient satisfaction will be low.
The field of prosthetic design has been strongly influenced by the low demand. If a patient has no complaints related to the functionality of the prosthesis after rehabilitation, the attention will be focused on problems related to slower-acting phenomena, such as wearing, that limit the durability and lifetime of the prosthesis. Previous studies have been focused on two main areas: estimation of the load transfer mechanism in relation to the frictional contact parameters, and prediction of the wear based on the influence of those parameters on the phenomenon itself.

Several loading regimes are considered, as characterised in (G. Bergmann et al., 2001, A. Rohlmann et al., [1]. For every regime, a dynamic finite element simulation of the dry-friction contact was performed to establish the contact traces and the contact pressure distribution. Those characteristics, combined with the frequencies of the activities considered in M. Morlock, et al., 2001, [2], are used as the input data for a statistical evaluation of the joint loading over some period. Previous tribological studies showed strong correlations between the contact pressure and the wear rate (L. Capitanu, et al., 2005 [3]). This means that different activities will lead to different levels of wearing. Furthermore, for the same activity, the contact pressure will vary during the different instances of the activity. Therefore, the wearing phenomenon will be not a constant rate phenomenon for a given activity or between activities.The issue is even further complicated considering that different activities have different frequencies. Further investigation is needed to understand how to quantify a priori the characteristics measures of the wearing phenomenon across activities. In this study, we sought to develop a reliable method to predict the volumetric wear rate in total hip prostheses [4, 5]. This method involves FE analyses of the contact mechanism (to estimate the contact pressure variation during activity), evaluation of specific quantities that have been introduced to allow comparison between the instances within an activity and between different activities and a special summation technique that accounts for the frequency and load magnitude of every activity.

Some clinical studies Blunn et al., 1997 [6] and Wasielevski et al., 1994 [7] have identified the considerable wear of the polyethylene tibial insert due to the mechanics of non-conforming contact as the main cause of prosthetic failure.

Knight et al., reported in [8] that from 209 cases of primary TKR, 18 prostheses failed, with all of the tibial inserts showing wear pits and delamination. The average time to failure was approximately 80 months, close to that reported by Heck et al., 1992 [9], but other studies (Engh et al., 1992 [10], Jones et al., 1992 [11], Kilgus et al., 1991 [12] and Mintz et al., 1991 [13] have shown that gross delamination could occur even earlier.

Examining an early retrieved prosthesis, Ries and his collaborators [14], noticed gross pitting and delamination of the tibial insert accompanied by an increase of percentage cristalinity at a plane coincident with the plane of delamination and the presence of subsurface oxidation peak typically observed in UHMWPE inserts, gamma-irradiated in air and aged. Although the issue remains controversial, the idea that increasing the thickness of the tibial inserts would reduce wear was shown to be false. Indeed, it is the incongruence of the contact surfaces that leads to high contact pressure localised over small areas and subsequently to the fatigue wear and delamination of a polyethylene part. In an analysis of two retrieved TKPs (after 21 and 16 years from implantation), Oonishi [15], noticed the presence of parallel scratches (oriented anteroposterior) on the femoral component and considerable delamination of the polyethylene tibial insert.

Still controversial, the idea that increasing the thickness of tibial inserts will reduce wear was rather false. Indeed, it is the incongruence of contact surfaces which conduct to high contact pressure localized on small areas and subsequently to the fatigue wear and delamination of polyethylene part. Examining two retrievals one could identify three characteristic marks of fatigue wear phenomena.

In our study, J. Onisoru et al., 2006 [16] which is based on examination of two retrieved prostheses, also identified characteristic marks of the fatigue wear phenomena. Have been observed wear pits on the contact surface, which are specific to adhesive wear, as well as the delamination of the peripheral area of the medial part, which is an effect of fatigue wear. From the qualitative evaluation of this two retrieved implants we could conclude that the fatigue wear debuted with some pitts of the contact surface and delamination (especially in the peripheral medial part of the tibial insert) but, in time, more severe damages occur (gross delaminations, massive losses of polyethylene).

2. METHODOLOGY AND RESULTS FOR ANALYSIS OF TOTAL HIP PROSTHESIS

In case of Total Hip Prosthesis (THP), Clinical experience has shown that for in cases of revision replacement of the cup due to wearing, the contact surface is no longer spherical, and some areas are more damaged than others. This phenomenon is caused by the unequal distribution of the load over the contact surface.
The distribution of the contact pressure is difficult to evaluate experimentally. Instead, by transferring the contact force from the femoral head to the acetabular cup using a numerical model of the contact mechanism, it is possible to obtain a good estimation of the distribution of the contact pressure over the entire interface. A non-linear, dynamic FE analysis of the contact couple behaviour under the loading produced by a specific activity provides a good method for such an analysis. The mechanical properties used in the analyses are listed in Table 1. Coulombic friction is assumed with a value of 0.06 for the coefficient of friction.

Table 1: The mechanical properties of the materials used in the FE analyses

Material	**Young Modulus E (GPa)**	**Poisson ratio ν**
Co - Cr alloy	200	0.3
UHMWPE	1.06	0.36

In our study, we used an FE model of a rigid-to-flexible contact between a rigid sphere (that simulates the femoral head of a total hip prosthesis) and a flexible hemispherical cup – assuming elastic behaviour of the UHMWPE – with the same geometry as a real cup (see Figure 1).

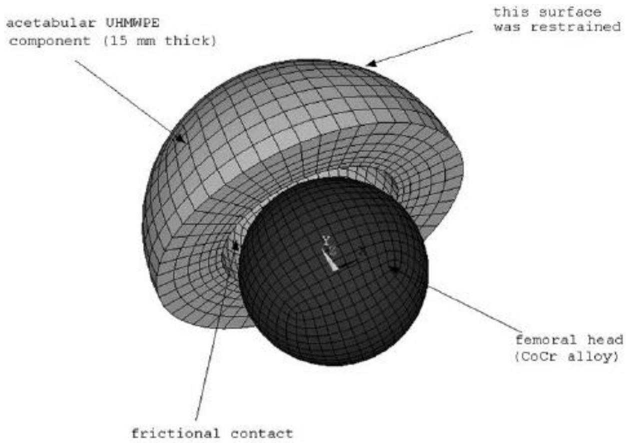


Figure 1 Finite element model for the hip artificial joint

To validate our assumption of a rigid-to-flexible contact, we estimated the contact stiffness of the two continua that are contact with each other by the following formula:

$$\gamma = \frac{E}{4\left(1-\nu^2\right)} \qquad (1)$$

Using formula (1), one could obtain the ratio of the stiffneses of the two continua, which is large enough to support our assumption:

$$\frac{\gamma_{fem.head}}{\gamma_{UHMWPE}} = 200.$$

In our analysis, we obtained the dynamic distribution of the pressure, which means that we determined the value of the contact pressure for every point on the surface and in every moment.
If we denote the contact surface with and the duration of the active cycle of a specific activity with, we can determine the contact pressure as a function of time and position:

$$p(t,x); t \in [0.T]; x \in S,$$

and the maximum pressure as:

$$p_M = \{p(t,x); x \in S, t \in [0,T]\}.$$

These two measures are specific for the level of the loading at a specific moment of the activity or for the entire period. Based on the functions defined above, we can evaluate the relative measure of the proximity to the maximum area of pressure, designated as the "instantaneous index of pressure".

$$\widetilde{p}(t,x) = \frac{p(t,x)}{p_m(t)}. \qquad (2)$$

For every moment of time, we can determine the level of loading by:

$$\lambda(t) = \frac{p_M(t)}{p_M}. \qquad (3)$$

The volume of the material removed by wear is (J.F. Archard. 1953 [17]):

$$V = kPnL = \int_0^T k(t)P(t)vdt, \qquad (4)$$

where

$$v = cont. = \frac{L}{T} \qquad (5)$$

is the velocity of the relative movement of the frictional couple (assumed constant here), and $P(t)$ is the result of contact determined by integrating the contact pressure over the entire interface surface:

$$P(t) = \int_S p(t,x)dA. \qquad (6)$$

As mentioned above, experimental tests showed some correlation between the contact pressure and the wear rate. By testing the CoCr/UHMWPE hip prostheses in a hip joint simulator, Wang [18], found that the wear factor could be expressed as:

$$k = C\sigma_0^n. \qquad (7)$$

They also established the values for constant C and the exponent n: $C = 7.99 \cdot 10^{-6}$ and $n = -0.653$. By substituting (5), (6) and (7) in (4), we can obtain the volume of the material removed during a single activity:

$$V = nC\frac{L}{T}p_M^{n+1} \cdot \int_0^T\int_S \left(\frac{p_m(t)}{p_M}\right)^{n+1}\left(\frac{p(x,t)}{p_m(t)}\right) \cdot dAdt = nC_M^{n+1}L\int_S\frac{1}{T}\int_0^T \lambda^{n+1} \cdot \tilde{p}(t,x)dtdA. \qquad (8)$$

For more than one activity, we can perform an algebraic summation as follows:

$$V_{med} = \sum_i V_i = \sum_i\left(n_i Cp_M^{n+1}L\int_S\frac{1}{T}\int_0^T \lambda^{n+1} \cdot \tilde{p}(t,x)dtdA\right) = \int_S\left(\sum_i n_i Cp_M^{n+1}L \cdot \frac{1}{T}\int_0^T \lambda^{n+1} \cdot \tilde{p}(t,x)dt\right)dA. \qquad (9)$$

One can see that the integrand

$$I = \sum_i n_i Cp_M^{n+1}L \cdot \frac{1}{T}\int_0^T \lambda^{n+1} \cdot \tilde{p}(t,x)dt, \qquad (10)$$

is a function of position only, and it may be a good measure of the tendency of some areas of the contact interface to be affected by wear.

This means that a map of the distribution of the values of this integrand could be a good predictor for the geometrical features of the wear of the joint surfaces. We have denoted this integrand (either summed over activities or specific for every activity) as being a wear geometrical descriptor.

The methodology for predicting the fatigue wear phenomena of the tibial tray combines FE analyses for the active loading cycles of the relevant routine activities with a summation technique that is based on computation of a cumulative estimator of damage. An FE model of the artificial joint contact was used (see Fig. 1) for all analyses. The model includes one femoral condyle and one half (the medial one) of the polyethylene insert, and the metallic tibial tray.

For deformable parts (the tibial parts), solid brick elements with 8 nodes and 3 DOFs per node (all three translations) are used. The femoral condyle has a toroidal shape with a radius of 22 mm in the sagital plane (the flexion plane) and a radius of 30 mm in the transversal plane and was considered rigid. The elastic linear constitutive laws are considered for the metallic and the plastic parts. The material properties are listed in Table 1.

Table 1: The mechanical properties of the materials used in the FE analyses

Material	Young Modulus *E* (GPa)	Poisson ratio *v*
Co - Cr alloy	200	0.3
UHMWPE	1.06	0.36

The routine activities considered relevant for this study are the active cycles of normal walking, stair descending and stair ascending. The kinematics of all three movements are illustrated in Figure 2 – in (a), (b), and (c), the segmental movements of the lower member are sketched [1].

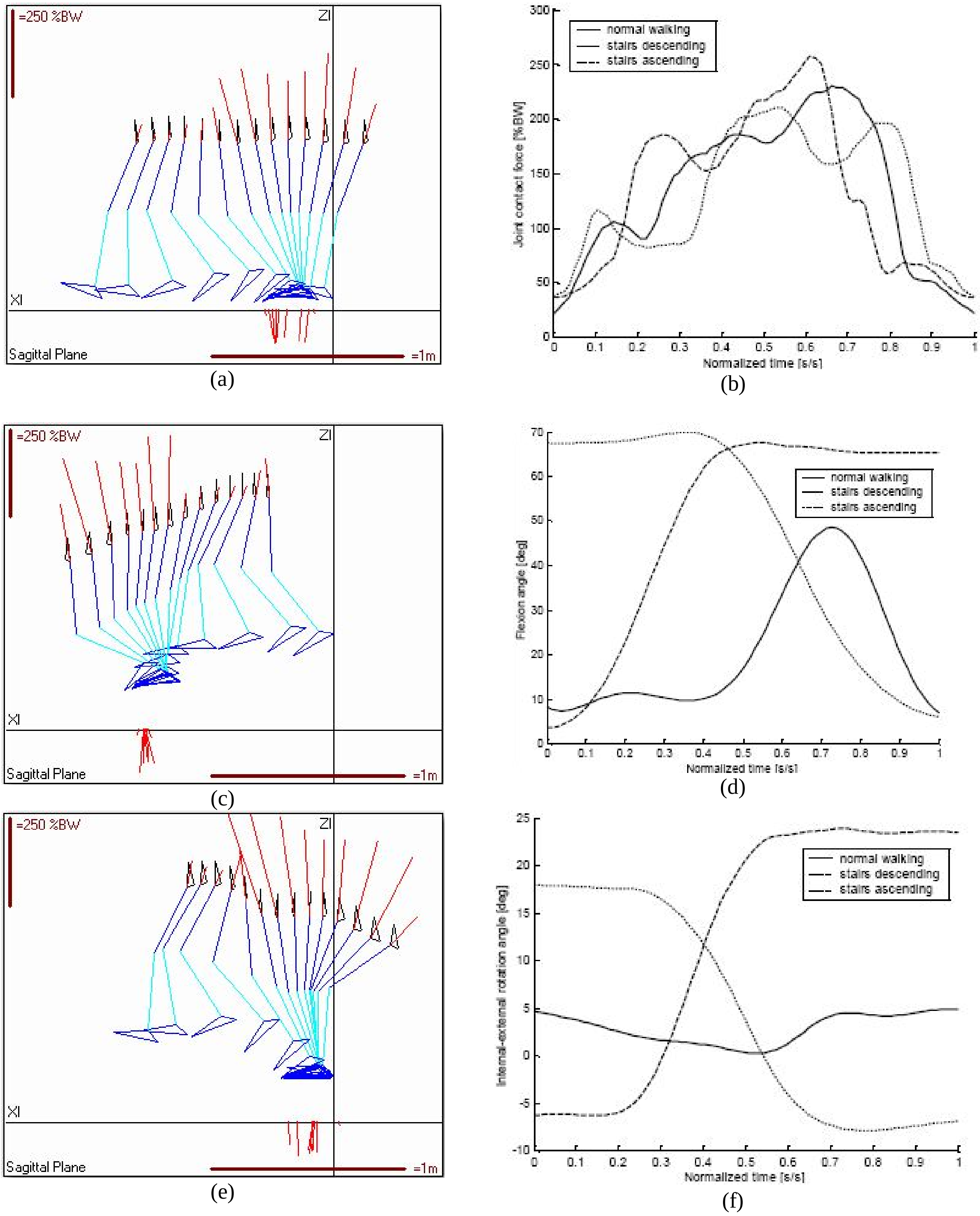
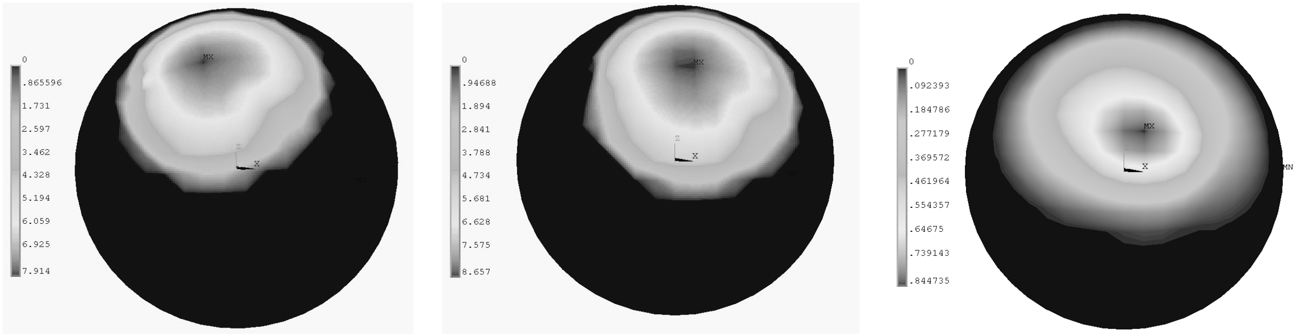


Figure 2 Dynamic conditions during all activities: kinematics of the lower member from S.J.G. Taylor and P.S. Walker, 2001 [19] (a – normal walking, b – stairs descending, c – stairs ascending) and loading from T. Villa, 2004 [20], and kinematics from S.J.G. Taylor, 2001, of the joint surfaces (d – contact force, e – flexion angle, f – internal-external rotation angle).

In (d) the compressive forces of the contact are listed and compared for all activities (Taylor and Walker, 2001) [20]. In (e) and (f), two determinants of the kinematic conditions (the flexion angle and the internal-external rotation of the tibia) are plotted (Bergmann *et al*., 2001 [1] and S.J.G. Taylor, 2001 [20]).

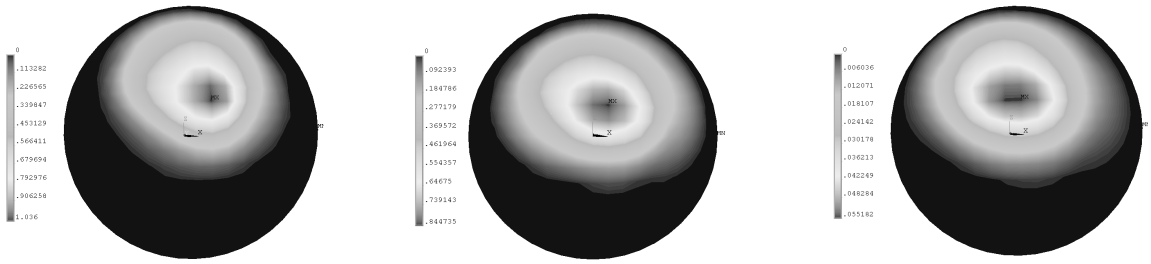
In Figure 3, one could see, the distributions of the contact pressure specific for every activity at the instance of time when the maximum pressure is achieved. One could see that even for all activities the maximum is located in the superior part of the acetabular cup, the maximum wear is likely to occur closer to the median part of the cup.



(a) (b) (c)

Figure 3 Contact pressure for (a) normal walking, (b) stairs-up, (c) stairs-down

One could see that even for all activities the maximum is located in the superior part of the cup, the maximum wear is likely to occur closer to the median part of the cup (Figure 4).



(a) (b) (c)

Figure 4 The wear geometrical descriptor for (a) normal walking, (b) stairs-up, (c) stairs-down

Using the summation technique described above, we obtained the distribution of the cumulative wear geometrical descriptor (see Figure 5) for the activities considered.

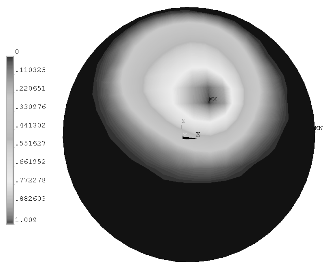


Figure 5 Cumulative geometrical descriptor for the three activities considered

In case of TKP, the compressive force is applied to the tibial tray (as the distributed pressure on the lower basis of it). The two rotations (flexion and internal-external rotation) dynamically constrain the femoral part. We considered that the actual level of flexion (not higher than 70 degrees) does not induce an antero-posterior translation (usually appearing when the ligaments are overstrained), as it does for extended flexions. The contact mechanism includes rolling and sliding of the two joint surfaces – the femoral condylar toroidal surface and the planar surface of tibial insert. The friction obeys Coulomb's law with a constant friction coefficient of 0.12 (Villa et al., 2004 [20]). For every activity, a damage estimator can be computed from the variation of the shear maximum principal stress in every element (Sathasivam and Walker, 1999 [21]) – relation 10.

3. Methodology and Results for Analysis of Total Knee Prosthesis

Performing the dynamic analyses of the contact between the femoral metallic condyles and the polyethylene tibial insert (in the conditions described above), it is possible to determine the characteristics of the contact mechanism in the artificial knee. For example, the trajectories of the contact spot for all three activities are plotted on a sketch of the medial part of the polyethylene insert (see Figure 6).

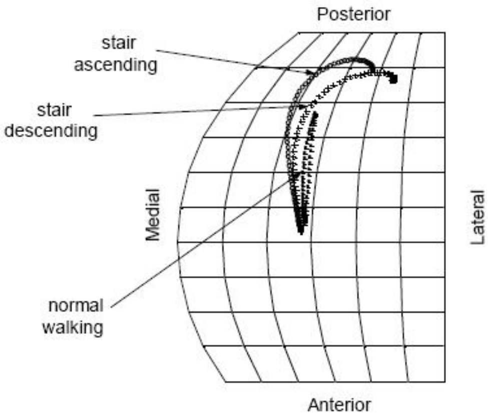


Figure 6 The contact paths for all three cases considered

By examining the sketch, one can see that not all activities involve extension of the knee – this means that only the median and posterior parts of the insert are used. For normal walking, the trajectory is a closed curve (like a hysteretic loop). The stair ascending and stair descending trajectories are quite similar (although they are opposite in direction). Both activities have a large internal-external rotation (for stair ascending, loading occurs at the beginning of the cycle; for stair descending, loading occurs at the end of it). It can be seen that the normal walking loading cycle will affect the median area of the medial part of the insert, and the stair ascending and descending activities will have their maxima located in the posterior parts of the insert (Figure 7).

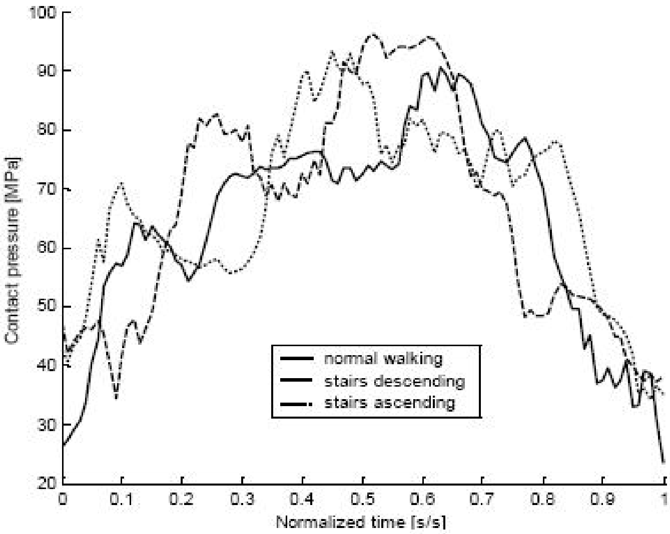
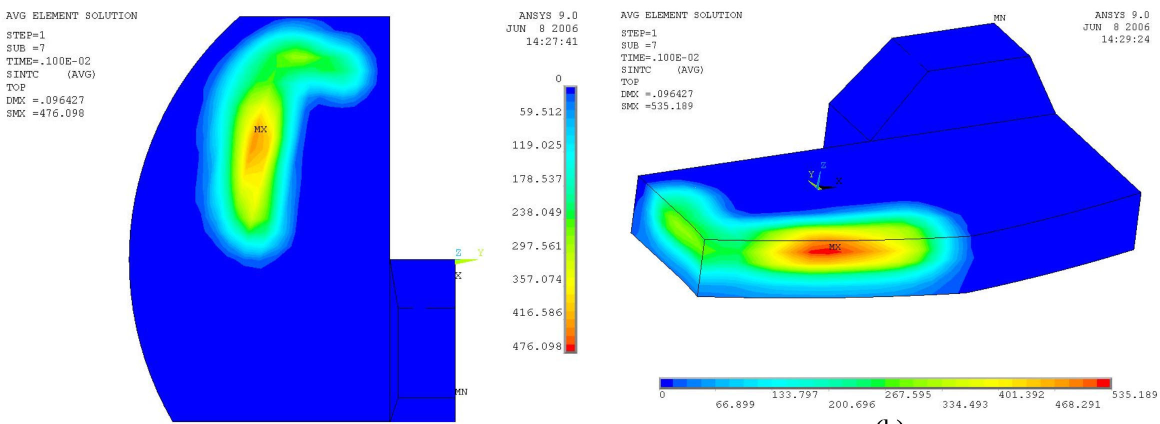


Figure 7 The contact paths for all three cases considered.

As expected, due to the geometry of the contact (the joint surfaces are a toroidal surface for the femoral condyle and a plane surface for the tibial insert), it is clear that the contact spot will have an elliptical form with the major axis oriented transversally to the sagittal plane of the femoral condyle.

The maximum contact pressure is a good estimate of the intensity of the loading. The variation in time of this contact parameter will follow the variation of the magnitude of loading. An initial step maximum can be seen for all activities, corresponding to the initiation of the movement – the contact between the foot and the floor. The contact pressure is important as an estimator of the loading magnitude, but it is even more important for estimating the fatigue wear and the maximum shear stress, as the parameter used in the damage estimator, as stated by formula (1). From the contact mechanism, which involves both rolling and frictional sliding, the maximum shear stress will be located under the contact surface at a distance dependent on the aspect ratio of the elliptical contact spot. Using the cumulative estimator defined in formula (2), the areas where the damage is likely to occur can be identified from the cumulative distributions plotted in Figure 8 (the L shaped region in a different colour). It can be seen that the most affected areas are the medial and posterior parts of the tibial insert.



(a) (b)

Figure 8 The damage score map (a) at the contact surface, (b) in the subsurface.

One could may also notice that the maximum damage will occur at the subsurface of the insert (where the shear stresses from normal walking is maximal), which is usually the starting plane of severe delamination. The magnitudes of the damage estimator (~ 4.77 MPa^2 at the surface and ~ 535 MPa^2 in the subsurface) are in concordance with the results of Sathasivam and Walker, 1999 [21], who obtained a damage score of ~ 230 MPa^2 for a constant loading of 1,000 N (approx. 120% BW), representing one half of the maximum load from normal walking or from stair ascending or stair descending activities.

The routine activity studied here is the normal walking for which relevant data are available in (C.L. Vaughan et al., 1999 [22]). The dynamics of knee joint is plotted in Figs. 9 – in (a) the compressive force acting on joint is plotted as resulting from inverse dynamics analysis, in (b) and (c) two determinants of the cinematic conditions (the flexion angle and the internal-external rotation of tibia) are plotted (as resulting from kinematics analysis of angular motion in joints).

The compressive force is applied to the tibial tray (as distributed pressure on the lower basis of it). The two rotations (flexion and internal-external rotation) are dynamically constraining the femoral part. We considered that for the actual level of flexion (not higher than 70^0) does not induce an antero-posterior translation (usually appearing when the ligaments are overstrained, as for extended flexions.

The contact mechanism includes rolling and sliding of the two joint surfaces – the femoral condyliar thoroidal surface and the planar surface of tibial insert; the friction obeys the Coulomb law with a constant friction coefficient of 0.12 (Villa et al., 2004 [20]).

For each studied case a damage estimator could be computed from the variation of the shear maximum principal stress in every element (Sathasivam and Walker, 1999 [21]):

$$I = \sum_i n_i C p_M^{n+1} L \cdot \frac{1}{T} \int_0^T \lambda^{n+1} \cdot \tilde{p}(t,x) dt, \qquad (11)$$

where: D_f^k - damage function for k activity and τ_i - shear maximum stress on elements for time t_i.

Figures 9 shows the dynamics of knee joint.

4. DISCUSSION AND CONCLUSSION

A method utilising finite element analysis for evaluating the contact mechanism in an artificial knee joint and an artificiak knee joint, with a summation technique based on a damage estimator was successfully used to predict the areas where the damage is likely to occur. In the case of total hip prostheses, previous studies reveal that there is a strong correlation between the wearing rate and the contact pressure. Therefore, the wearing phenomenon will be not a constant rate phenomenon for the same activity or across activities.

The distribution of the contact pressure was dynamically evaluated by non-linear FE analyses that simulated the active cycle of normal walking, ascending stairs and descending stairs. The maxima of the contact pressure are plotted for each activity. A summation technique was adopted to evaluate the wear rate across activities. The adopted method used a newly defined parameter, namely, the wear geometrical descriptor, which is a good estimator of the distribution of worn areas across the contact interface. By integrating this descriptor over the entire joint contact surface, the wear volumetric rate was obtained.

The resulting value (28.17 $mm^3/10^6$ cycles) is in concordance with the values determined from experimental tests performed with special hip joint simulators (Engh et. al., 1992 [10], Jones et. al., 1992 [11], Kilgus et. al [12]., 1991, Mintz et. al., 1991 [13]).

In the case of total knee prostheses, based on the study described above, one can formulate several conclusions. First, the fatigue wear was identified as the main phenomenon on responsible for the polyethylene tibial insert failure. Clinical studies show that even tibial parts that are retrieved for other reasons (implant loosening, misalignment, etc.) presented markers of fatigue wear (cracks in the subsurface, wear pits, delamination, and loss of large pieces of polyethylene).

The cumulative nature of the wear fatigue phenomenon requires a qualitative and quantitative evaluation of the transfer of loadings through the joint and a summation technique that is relevant for the variety of the human activities.

The first part was assessed by a combination of dynamic finite element analyses of the joint contact mechanism. For the second part, we assumed that normal walking, stair ascending and stair descending were the regular activities with a dominant impact on the phenomenon.

For all three activities, the finite element analyses were performed, and a damage score was computed. Finally, a cumulative damage score (that accounts for all three activities) was determined and the areas where the fatigue wear is likely to occur were identified. From the distribution of the cumulative damage score (plotted in Fig. 5), one can see that the areas that may be damaged are located medially in the median and posterior regions of the tibial insert. Although the activities of stair ascending and descending generate a higher contact pressure due to the increased frequency, normal walking is still dominant.

A closer look at the distribution of the damage score reveals that the maximum damage is likely to occur not at the contact surface, but in the subsurface.

The presence of maximum values in an area where other studies (Ries *et. al.*, 1995 [14]) have identified an increase the percentage crystallinity and the presence of a subsurface oxidation peak induced by gamma irradiation and the ageing of the prosthesis could be an explanation for the early delamination of the polyethylene.

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