Influence of the diameter and the material type on the contact pressure in the acetabular component of a total hip prosthesis.

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Abstract

The hip joint is often affected by diseases that lead to its failure and consequent need of replacement by an artificial joint. In this case, the total hip arthroplasty is the most common medical procedure to substitute the natural joint. The total hip prostheses are available on a wide range of sizes and materials. In this work, it is develop a computational model to compute the contact pressures on the joint surfaces of the prosthesis. This model is able to analyze the most commercialized diameters and the different material combinations used on the hip prosthesis. The model, based on the Finite Element Method, is developed in the finite element software ABAQUS®. Firstly, the geometric model of the three components of the hip joint, corresponding to the external cup, the liner and the femoral stem's upper part, the head, is built and discretized using a suitable finite element mesh. Then, the stress on the interface between the head and the liner are computed solving a quasi-static contact problem. The finite element mesh is composed of 8 node hexahedral elements. The load cycle used to simulate the contact pressures on the acetabular component was measured *In Vivo* by Bergmann, using instrumented hip prosthesis. The results were analyzed in a way to promote the study of the effects of the different diameters and material type combination on the contact pressure. These results are in agreement with results obtained in previous studies and can be an important factor on the decision of which type of prosthesis to implant.

Keywords: Hip joint; Contact pressure; Finite Elements; Hip Prosthesis Diameter; Hip Prosthesis Materials.

1. Introduction

Artificial hip joints are nowadays a usual procedure to replace the anatomic hip joint. The success of this surgical intervention is due to a number of scientists and physicians that have been following this line of work over the last few centuries. The prosthesis itself has been modified and initial flaws have been readjusted over the years but the main concept and design of the prosthesis maintains roughly the same. Although the hip prosthesis performance is reasonably good, there are still some complications that lead to the artificial hip failure. Pathologies such as osteoporosis, osteoarthritis, rheumatoid arthritis or acetabular protrusion are examples of those complications. In addition, artificial prosthesis produce more friction than the natural ones which consequently means an increase of the wear rate, which still is one of the major prosthesis unsolved problems.

Studies have been made as a response to such complications and as a way to improve the prosthesis' performance. Some of these studies are related to computational simulators that try to predict the hip prosthesis' functioning under specific types of load cycles, which are generally related to human movements. Some authors have already presented models to analyze the hip joint. For example, Maxian [1] presented a computational model to compute the wear on acetabular cups or Fialho [2], which not only tried to relate the contact pressures with the wear rate of the acetabular component but also addressed the heat generation problem. These models were based on the Finite Element Method and used related software in order to compute the desired results.

As a consequence of the studies and progresses made on this area, there is a wide range of different prosthesis type currently available in the market [3,4,5]. The main differences between the commercialized products are in the materials from which they are made. Other differences include the diameter of the acetabular components, the size of the femoral stem [6] and the way the materials are attached to the patients' bone mass [7].

The aim of the present work is to develop a computational simulator that estimates the differences in the contact pressures on the acetabular bearings in prosthesis with different diameters and made of different materials combinations. The model was computed with the help of the finite element software ABAQUS® in order to obtain both numerical and graphical results regarding the contact pressures. The contact problem is reproduced with significant similarity. The spatial discretization of the designed bodies was pondered with a mesh convergence study to increase the results' reliability. The chosen load cycle to obtain the results simulates a human step and was measured in vivo by Bergmann [8] using instrumented prosthesis.

The results obtained with the computational simulator are clinically meant to be an important factor to take in account in the decision of which type or size of the prosthesis to implant in a specific patient.

2. Methods

The contact pressure is measured in on the joint bearing surfaces. In order to obtain such results, three different bodies were designed per diameter. The bodies correspond to the femoral spherical head, the acetabular liner and the acetabular external cup. Therefore, three different models were reproduced due to the different diameters of the femoral head, more specifically, 28 mm, 32 mm and 36 mm. Several different material combinations were also tested for each of the considered diameters and will be analyzed in more detail later in this paper.

The contact problem is a problem of elasticity with quasi-static contact between the femoral head and the acetabular liner. It is, therefore, a non-linear contact problem. For the problem discretization, 8-node hexaedric elements were used, where the nodes are used to create a frontier in between the consequent nodes [9,10]. The computer generates the mesh although some restrictions were made. By the time the bodies were modulated, the incisions were made dividing the bodies into subparts, as shown on Figure 1. The advantage of such procedure is that the subsequent elements to the frontier between the subparts will have their nodes coincident with

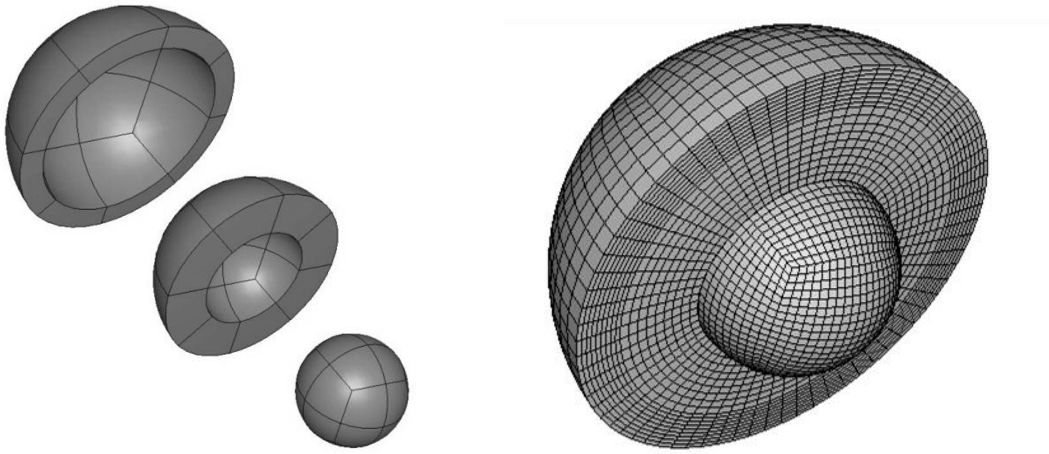

Figure 1. Computational model of the three parts and consequent discretization in finite elements.

the frontier itself. This enables the user to reshape the mesh in order to create a more homogenous mesh, which will provide more consistent results. According to the mesh convergence study made, a mesh with 812 contact nodes is enough to reproduce the computed contact pressures without compromising the computational process time.

The acetabular cup has an external diameter of 54 mm and an internal diameter of 48 mm and it is made out of titanium. These measures and material type are independent of the size of the spherical femoral head. This part is encastrated and therefore all its rotational and translational movements are null. The liner has a variable thickness, depending on the diameter of the chosen femoral head. Its thickness is always the difference between the inner diameter of the cup and the femoral head diameter. A boundary condition was defined in order to tie this part to the external cup. It has no specific orientation, i.e., it is oriented in accordance to the cup. The femoral head can have three different diameters, 28 mm, 32 mm and 36 mm, as referred above. There will be an initial distance to the liner, which is often called as clearance and has the value of 0.05 mm for all the different sized computational simulators. The head will have no spatial constrictions other than the contact established with the liner. It is on these two latest parts that the different material combinations will be tested.

The characteristics of the materials used on the current analysis were chosen according to [2]. On the other hand, the friction coefficients between the different material combinations were consulted in technical monographs of commercial prosthesis, gently assigned by Depuy Orthopaedics®.

Table 1. Mechanical properties of the used materials.

Material	Young's Modulus (MPa)	Poisson Ratio
Alumina (AlO)	375000	0.3
Polyethylene	2200	0.3
Cobalt – Chromium (CoCr)	230000	0.3

Table 2. Friction coefficients for the used material combinations.

Material Combination	Friction Coefficient
PE – CoCr	0.07
PE – AlO	0.15
CoCr – CoCr	0.05
AlO – AlO	0.05

The load cycle is manly obtained from experimental gait analysis data for the HIP 98 CD [8], where the gait cycle for different patients is described. Only 28 time instances from the patient KWR are considered in order to reduce the computing time of the method.

A study on the influence of the clearance value on the contact pressure was also made. In that case, a 14 mm PE – CoCr model was used and several clearance values were tested. Results of this analysis can be verified on the figure 2.

After a detailed analysis, we can state that, generally, the pressure contact increases as we increase the gap between the prosthesis components. For the first tested clearance values, 0.01 mm and 0.03 mm, the lowest contact pressures were noticed. The reason why these clearances are not often commercialized is because such lows values prevent the contact surfaces to be lubricated properly, which leads to an increased friction coefficient and an augmented contact area between the components. In addition, low clearance values tend to generate contact pressure peaks in the equatorial axis, which may lead to early prosthesis failure. On the other hand, high values of clearances tend to increase the contact pressure values, which may lead to and increased wear debris. Therefore, all the developed simulators have a clearance value of 0.05 mm.

3. Results

The figures 3, 4 and 5 represent the contact pressure distributions of a PE – CoCr hip prosthesis for the 28 mm, 32 mm and 36 mm respectively.

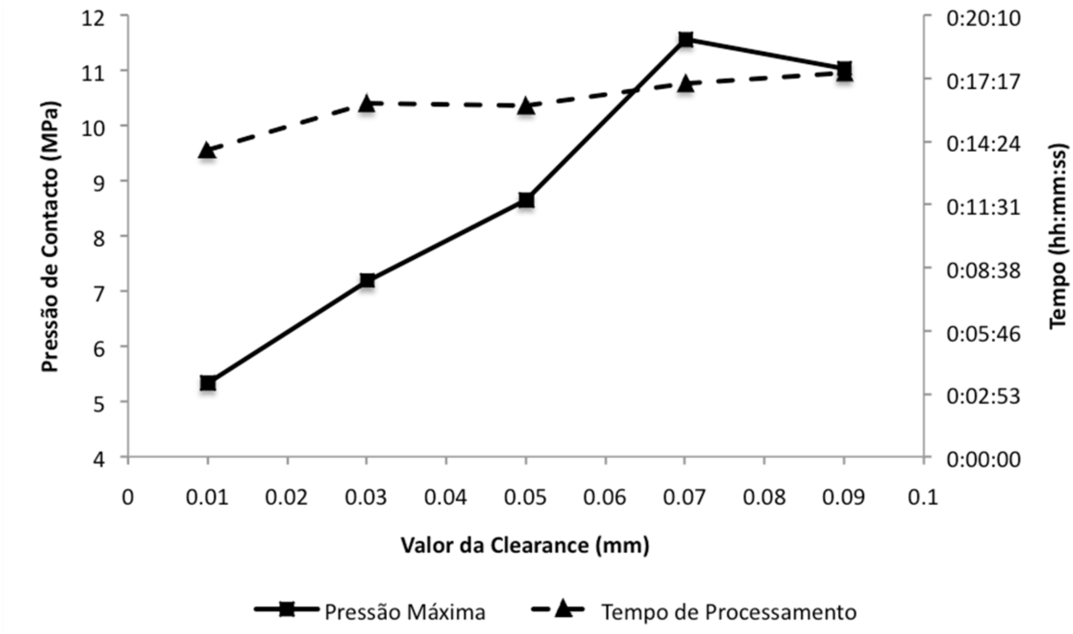

Figure 2. Influence of the clearance value on the contact pressure.

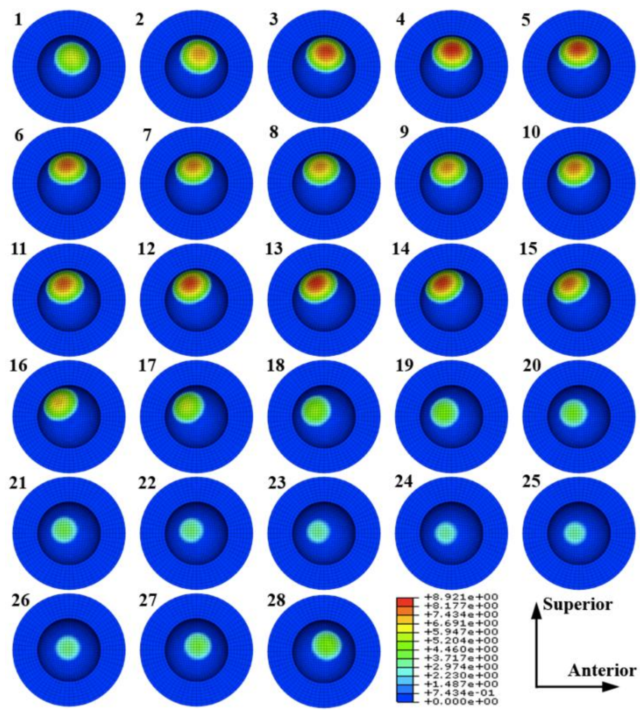

Figure 3. Contact pressure on the 28 mm model for the 28 instances of the gait cycle (CoCr – CoCr).

In an initial stage, some other load cycles were tested in order to validate the present results. The developed computational simulator had, for all these cases [1] and [2], minimal differences in terms of both maximum contact pressures per step and pressure distribution along the liner surface.

As we can see from a closer look at this set of figures, the load cycle can be divided in two different phases: a first action during which there is physical contact with the ground called Heel Strike and a second phase from the moment the patient leaves the ground, here called Toe-Off.

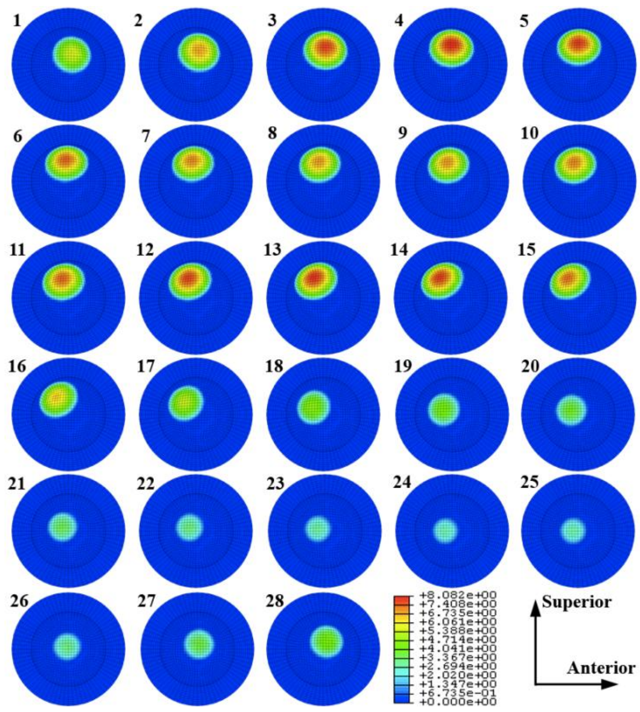

Figure 4. Contact pressure on the 32 mm model for the 28 instances of the gait cycle (PE – CoCr).

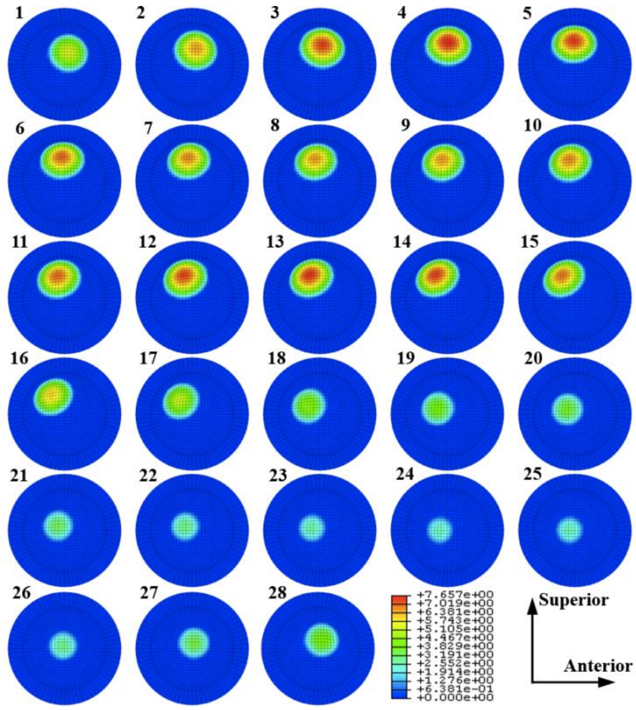

Figure 5. Contact pressure on the 36 mm model for the 28 instances of the gait cycle (PE – CoCr).

The Heel Strike phase includes instances 1 to 17 and is characterized by larger distribution areas along with higher maximum contact pressure values due to contact with the ground. On the other side, the Toe-Off phase includes the remaining instances and, because of the absence of contact, both of these parameters have lower intensity.

From the same analysis, we can conclude that instances with a higher contact resultant load will originate a higher contact pressure and vice-versa, which is an expected finding but also validates the simulator consistence.

As an evolutionary analysis along the diameter augmentation, allows noticing that the distribution area increases as well. This fact is due to the bigger polar surface contact between the head and the liner promoted by the bigger diameters. Another fact is that the value of the maximum contact pressure is decreased in every cycle for the same load as we increase the prosthesis diameter. For a 0.05 mm clearance, the contact pressure value decrease is non linear since two 4 mm increases do not lead to similar percentage increase. This fact can be checked in Table 3, where specific periodic instances have been analysed in order to quantify these contact pressure differences.

Table 3. Pressure contact (MPa) for the considered instances (PE – AlO).

Instance	28 mm	32 mm	36 mm
4	8.921	8.082	7.657
8	7.681	6.960	6.595
12	8.596	7.779	7.364
16	7.137	6.422	6.099
20	4.410	3.912	3.696
24	3.137	2.750	2.555
28	4.881	4.392	4.156

The reason why only 7 out of the 28 steps were chosen for analysis is to prevent an exhaustive and unnecessary study as the most important instances are among the chosen: instances 4 and 24, respectively the absolute higher and lower resultant load applied.

A quantitative analysis to Table 2 allows saying that the maximum contact pressures in each step decrease in average 10.3% if we switch a 28 mm prosthesis with a 32 mm. If changed by a 36 mm, the decrease is, again in average, 15.2%.

In order to consolidate the contact pressures on materials with different Young's modulus, an analysis to a PE – AlO prosthesis was made. Regarding the distribution area of the contact pressures, it is now slightly bigger and as for the maximum contact pressure it increases as well. This is due to the difference on the materials' elasticity, which causes the contact problem to be more localized for the same applied load. The obtained results are in the following table 4.

In an analogous way, a quantitative analysis was made over the decrease on the maximum contact pressure values. A change in diameter from the 28 mm to the 32 mm or the 36 mm will induce decreases of 10.3% and 15.3% respectively. Note that the values are similar to the results obtained for the PE – AlO prosthesis, which denotes that contact pressure evolutes in a similar way with the increase of the diameter on prosthesis made from materials with such difference in the elasticity modulus.

Table 4. Pressure contact (MPa) for the considered instances (PE – CoCr).

Instance	28 mm	32 mm	36 mm
4	9.039	8.181	7.735
8	7.758	7.022	6.645
12	8.672	7.844	7.413
16	7.173	6.461	6.105
20	4.412	3.914	3.697
24	3.128	2.744	2.553
28	4.906	4.419	4.179

Components with similar elasticity modulus were analyzed as well and the obtained results for a CoCr –CoCr hip prosthesis are represented in the Figures 6, 7 and 8.

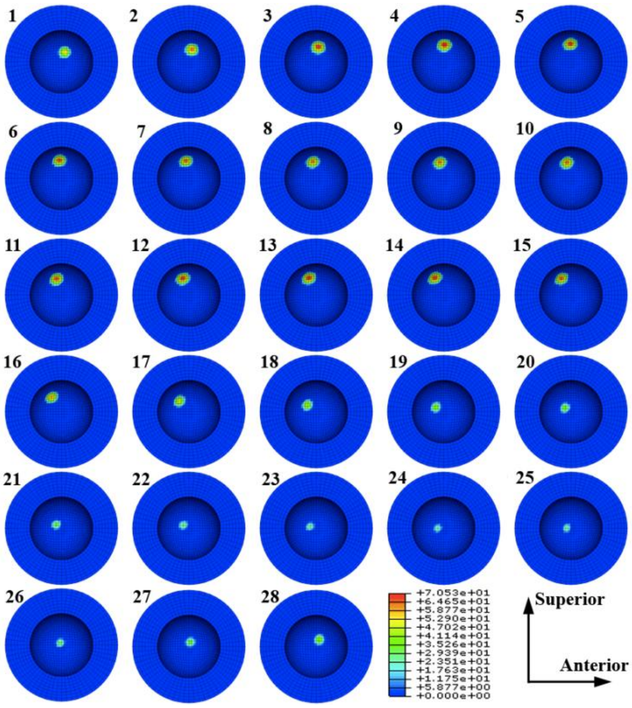

Figure 6. Contact pressure on the 28 mm model for the 28 instances of the gait cycle (CoCr – CoCr).

The contact pressure distribution area is now smaller compared to the PE – CoCr case. The significant increase in the stiffness of materials originates very small deformations, which is reflected by the smaller contact pressure distribution area. Once again, the instances where a higher load is applied are the instances where this distribution is broader.

A smaller polar contact area may reveal a decrease on the volumetric wear debris. The volumetric wear is the amounts of debris the prosthesis components that results from the friction between the contact surfaces. Now, if the polar contact region is smaller, less friction should occur in this area which proves the decrease of the volumetric wear.

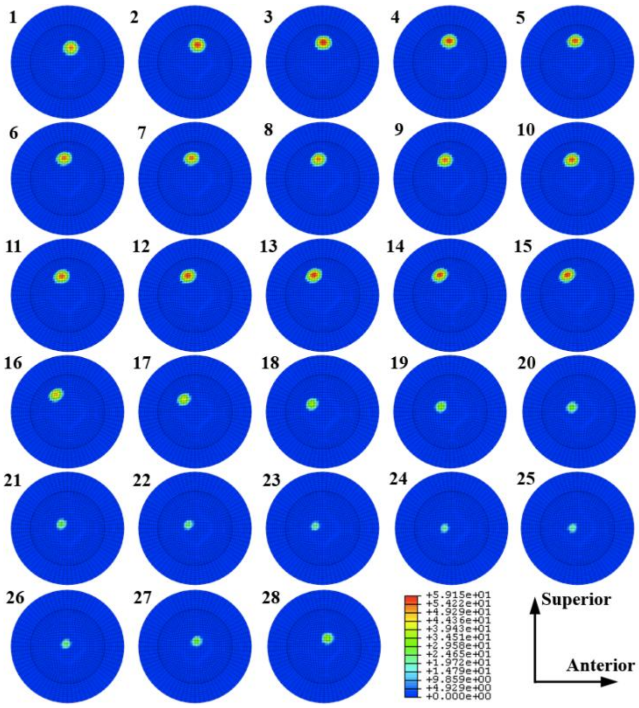

Figure 7. Contact pressure on the 32 mm model for the 28 instances of the gait cycle (CoCr – CoCr).

On the other hand, the values of the maximum contact pressure in each instance are in average ten times higher than the results obtained for the prosthesis made by materials with different stiffness. This can mean that the linear wear increases as well. The linear wear is the level of penetration of the femoral head into the acetabular liner and is usually expressed in mm. Therefore, higher pressure contact values for the same applied load may induce that there is an increase for the linear wear.

Quantitative analyses were made both for the CoCr – CoCr and the AlO – AlO prosthesis. Let’s first focus on the CoCr – CoCr hip implant. The pressure contacts for the selected instances are shown in the table 3. After a rather

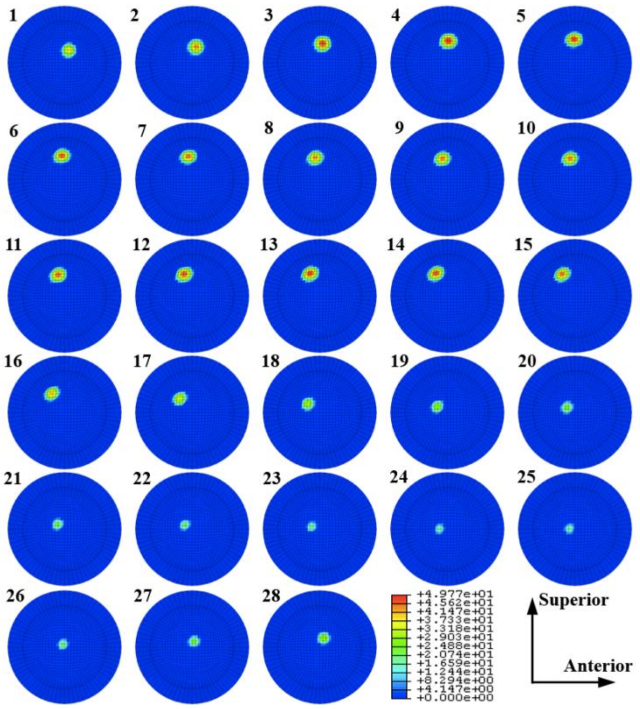

Figure 8. Contact pressure on the 36 mm model for the 28 instances of the gait cycle (CoCr – CoCr).

Table 5. Pressure contact (MPa) for the considered instances (CoCr – CoCr).

Instance	28 mm	32 mm	36 mm
4	70.53	59.15	49.77
8	63.70	52.90	44.60
12	67.85	57.40	48.25
16	60.08	49.59	41.26
20	40.39	32.23	26.69
24	29.86	24.70	20.29
28	43.90	39.50	33.04

Simple data treatment, it is possible to calculate the decrease rate in terms of percentage. As we changed the 28 mm diameter for the 32 mm, we experienced average changes of about 16.2%. Between the 28 mm diameter and the 36 mm diameter average changes were now 29.9%.

On the other hand, average changes in the AlO – AlO hip prosthesis were 17.7% and 31.1%, respectively as we can see in Table 6. In comparison to the results obtained in the CoCr – CoCr implant, we noticed that as the material's stiffness increased, the pressure contact differences along the diameter variation were

Table 6. Pressure contact (MPa) for the considered instances (AlO – AlO).

Instante	28 mm	32 mm	36 mm
4	96.15	79.52	66.90
8	87.17	72.52	61.21
12	92.19	77.83	65.54
16	81.36	67.85	56.99
20	51.65	39.79	33.03
24	38.99	29.86	24.67
28	56.77	50.64	42.39

more severe. Another important fact to take in consideration is that for the same clearance value, prosthesis made out of materials with similar stiffness are more sensible to diameter variations than materials with different stiffness, i.e., the relative decreases on the contact pressure as we increase the diameter are higher. Nevertheless, result differences on the diameter variations of the AlO – AlO prosthesis and the CoCr – CoCr prosthesis still predicts that stiffness is also an important parameter on the resulting contact pressures. As the stiffness of the materials of the bearings increases, the decrease on the contact pressure as the diameter increases is bigger.

The conducted tests allowed us to analyze the difference on the contact pressure as the material combination changed as well. The selected instance to preview such changes was instance number 4, again because it is the instance where the resultant load is the highest.

Figure 9 shows the difference on the contact

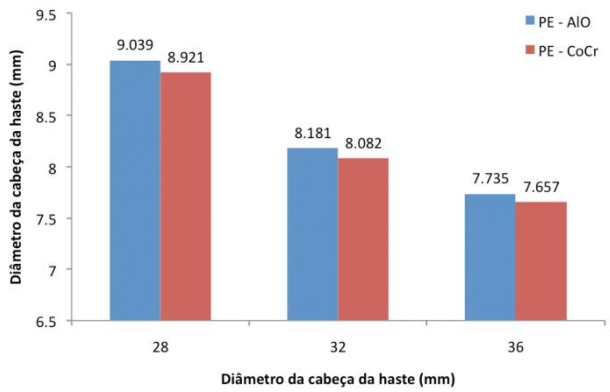

Figure 9. Contact pressure on the on the instance 4 for the different diameters.

pressure between the two material combinations with different Young's modulus (PE – CoCr and PE – AlO). As expected, higher stiffness results on higher contact pressure, although this difference is not very significant. For such different stiffness values, the contact pressures decreases, in average, only 1.17% when the Cobalt – Chromium femoral head replaces the alumina. The mean difference for every step of the gait cycle is even smaller (0.05%).

On the contrary, when components with similar rigidity where analyzed, the contact pressure differences is, in average, 26.0% for the instance 4 as we can see on Figure 10. The mean value for the decrease of the contact pressure on every single instance is 24.2%, which still is a very significant value.

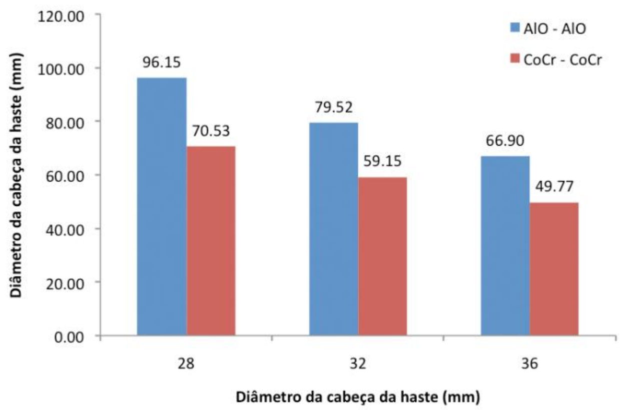

Figure 10. Contact pressure on the on the instance 4 for the different diameters.

Looking back on the analysis of the material combination on a hip prosthesis, some facts can be inferred. Firstly, the material choice for a femoral head that establishes contact with a polyethylene liner has no significant influence on the pressure contact values if the material stiffness is significantly greater than the polyethylene's stiffness. In addiction, when dealing with materials with similar stiffness, this difference assumes an important role on the contact pressure on the liner. We have to take in account the material's stiffness in order to take in account the contact pressures that will be exercised in the patient's prosthesis. This is an important factor to take in consideration when studying the linear wear in any hip prosthesis.

3. Conclusions

A computational simulator was developed that predicts the contact pressures on the acetabular component of a total hip prosthesis. This simulator has in account the contact details between the hip bearing surfaces as well as the prosthesis diameter and constituent materials.

As seen on the Results chapter, this type of analysis based on the Finite Element Method reproduces with liability a real case of a total hip arthroplasty. If, on the one hand, the mesh convergence supports the computed results, on the other hand there are no biological factors that induce great influence on the contact pressures values. The differences on the contact pressures on the different diameters tested induce that hip prosthesis of materials with similar stiffness are more sensible to diameter changes than prosthesis with major differences in their stiffness. This fact is also valid when we take in account the material type. Again, the prosthesis in which the materials' stiffness is similar are much more sensible to diameter changes than in prosthesis with differences on the materials' stiffness.

The obtained results may be an important factor in predictions of both linear and volumetric wears. These predictions are of great benefit to the evolution of the prosthesis design as the linear and volumetric wear are an important prosthesis failure cause. As referred in the previous chapter, the linear diameter depends on the maximum contact pressure on each step of the gait cycle. On the other hand, volumetric wear depends on the distribution of the contact pressures. The wear debris can get into the circulatory system and induce inflammatory reactions even in distant organs. Therefore, it is possible to develop a computation simulator of both wear coefficients in the different diameters and materials combinations, taking in account the data provided by this study.

Another improvement that is possible to add to this simulator is personalized gait cycles. In that case, one could use instrumented prosthesis that measure the load forces *In Vivo* on a post-operatory situation. Another possibility is the development of inverse dynamics computational models that simulate the load cycle and from there calculate the contact forces.

To summarize, the developed work pretends to be an added value to the decision process of which type and size of hip prosthesis to implant in a certain patient as well as to be the starting point for a major work on this topic.

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