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New Solution for Low Mass Hip Prostheses using Carbon Fiber

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Abstract—**In the paper there is presented a stage of the research on a new solution for prototyping hip prostheses, using lightweight and durable materials. An IT application is presented through which it was proceeded to establish optimal materials in terms of masses and mechanical strengths, taking into account different situations regarding static and dynamic loads. Subsequently, the prototype of the hip prosthesis was designed and CAD-CAM modeled, including both basic elements: the implant rod and the acetabular cup. As research, the solution was established by which an implant rod was prototyped, made of a metal core, wrapped in a carbon fiber sleeve.**

Keywords—implant, simulation, materials, carbon fiber

I. INTRODUCTION ON HIP PROSTHESES IMPLANTS

It is well known that the implants are more and more used in the context of saving the function of concerned member. This is due to the fact that, over the years, the branch of orthopedics has developed and more and more problems such as fractures, dislocations or degenerations could be solved through prosthesis [1], [2], [3]. Maybe the most conclusive example in this regard is that of prosthesis of the hip joint. Frequent and serious accidents often have serious consequences: femoral neck fractures, pelvic fractures, etc., especially in the elderly. Also with aging, a major problem is damage to the hip joint. Thus, pathologies such as hip osteoarthritis, osteoporosis of the hip joint cause unbearable pain, severe damage to locomotion to the inability of the person to move, even the inability to sit or in an orthostatic position. Hence the growing need to implant prostheses and hip orthoses [4], [5], [6].

Generally, metallic materials are used for the rod, sometimes, depending on the tissues biocompatibility, high density plastics or metal powders obtained by sintering procedures can be used. The acetabular cup is usually composed of a metal coating (stainless steel), inside which there is an inner cup made of plastics, processed, corrected and sanded. This is recommended to be done at a roughness of up to 0.1 μm, so that the friction between the rod head and the bucket is minimal [6], [7]. In certain situations (eg, femoral neck fractures) a prosthesis is sufficient only by implanting the stem, not the acetabular cup [6], [8].

The trend is to reduce body weight, finding solutions to use lightweight but at the same time durable materials is a compelling challenge. A major problem, however, for this purpose is that bone structures have an anisomorphic structure, which greatly complicates the possibility of finding very similar materials [9], [10]. However there is more and more about the solution on implementation of lightweight materials, such as titanium or carbon fiber [10], [11]. When implanting the rod, either it made of plastic or metal, there is the problem of biocompatibility with neighboring tissues [8], [12]. It is known that the recovery of patients after any invasive or non-invasive operations is a very important aspect. In this context, the effects on the skin, on the tissues near the implant must be taken into account [13].

In the current context, the issue of a research internship has been raised regarding the possibility of implementing a solution through which, in order to increase biocompatibility, light materials such as carbon fiber can be combined with conventional materials such as stainless steel.

II. PROPOSED SOLUTION ON CARBON FIBER USING

A. Carbon fiber advantages

Carbon fiber is easier to accept by the body, being a basic constituent of tissues. One of the first medical uses of carbon fiber was the replacement or repair of ligaments and tendons, in that it does not inhibit tissue growth but, on the contrary, can act as a scaffold for tissue proliferation. Due to the good biocompatibility and that it has been scientifically proven that there are no toxic effects generated by the carbon fiber printing process, it could be successfully used for prosthetic components, especially for hip joint [9], [11].

B. Carbon fiber strength

A first aspect of the research focused on finding the most efficient solution that can make a comparative assessment of the advantages and disadvantages of using certain materials or combinations of materials in the composition of a hip prosthesis. In this case, the issue of a simple and quick evaluation for different prototyping materials, such as ABS plastic, stainless steel or carbon fiber, was raised. Moreover, the issue of efficient evaluation was considered not only from the point of view of materials, but also of the 3D printing resolution for prototyping.For this purpose, a simple and quick solution was found, namely that this assessment should be aided by PC ensured, using a user-friendly graphical interface, first of all accessible to orthopedic doctors.

For this, a flexible software application has been developed, through which it is possible to evaluate as efficiently as possible to what extent a hip prosthesis could or could not withstand different static and dynamic loads, in different situations (orthostatism, normal walking, running , jumping). The application was developed in the *LabVIEW* environment, because it allows an easy and user-friendly interface, having a very suggestive graphic interface. In this case with the orthopedic doctor who could use it [10], [14]. This would allow the most efficient simulation to what extent, for a certain patient (depending on age, weight, anthropometric dimensions) a hip prosthesis could withstand different situations, depending on the materials and the resolution of prototyping.

C. Developped software application description

The design of the application had the hypothesis of the possibility of complex simulations to cover as many of the possible real situations. These refer both to the body mass of the subject and to the different types and structures of materials that could be part of the hip prosthesis. [15]. Figure 1 there presents an example of the software interface.

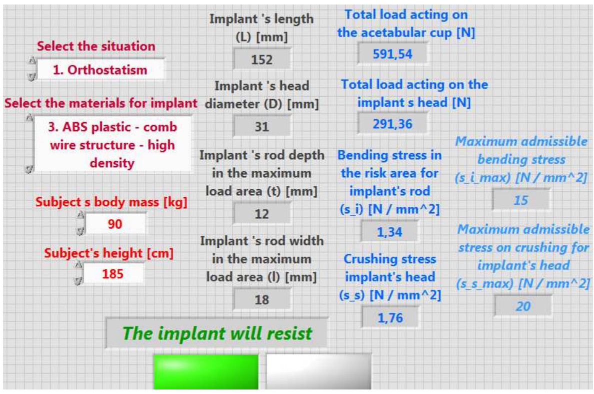

Fig. 1. Sample of using developped software application

The user can select from a predefined list the specific case. Since the purpose of this application is to establish assisted which would be the ideal combination of materials for making the implant, another dialog box for selecting different possible combinations of materials was envisaged.

As programming, a *Switch Case* selector was called, addressing a multiple case structure, each one invoking the calculus algorithm for each combination of materials. Two numeric inputs were defined for introducing the data of the concerning subject (the mass and the height). As output data, the implant tehnical charateristics can be done, like the length, head's diameter, rod's depth and width. Finally, the most important output data are referring to the loads and deformation acting on the hip implant, these being compared with the maximum admissible bending and crushing stress for the implant. For this purpose, we proceeded to the programming of the calculation algorithms, specific for each case, starting from the equations for determining the tasks in static and dynamic regime at the implant level. For this, first od all, the research had a theoretical, analytical approach, regarding the establishment of calculation equations specific to the determination of the bending stresses at the level of the hip prosthesis head, in different hypostases.

In this paper, only a part of the analytical study is presented and explained, namely the one regarding the determination of the loads and bending stresses that occure in the head of the prosthesis for orthostatism and for normal walking. In order to determine the static stresses on the endoprosthesis capsule, it could be considered that in the case of orthostatism, the forces acting on the prosthesis capsule are the body weight that rests on the hip joints and the reactions in the joints at the ends of the rods. Considering that Ns means the body weight acting for a single hip joint, and with Rs is the reaction force at the hip joint, means that the total force acting on the stent capsule will be $P = Ns + Rs$. In order to simplify and approximate the calculation of static loads, we will start from the following hypothesis: It is assumed that at the level of the hip joints the return weight is approximately equal to two thirds of the total weight of the person concerned. Considering G the body weight of the person, means that at the level of the hips will return a force of weight equal to $2/3 * G$. Given the fact that it is assumed that, in orthostatism, the weight on each hip joint is distributed equally, this means that at the level of a single hip joint will return a weight equal to one third of the total weight of the patient. So $Ns = 1/3 * G$. Taking into account the equation of the equilibrium of forces in static regime for a single hip prosthesis, it can be considered that the reaction force in the joint, acting in the opposite direction will have the same value with the weight force. So $Rs = -Ns$. Considering all forces acting simultaneously on the hip prosthesis head, it follows that the total load will be $P = 2 * Ns$. As $Ns = (0.33) * G$, it means that the total load acting on the hip prosthesis capsule will be $P = (0.67) * G$.

From the point of view of determining the static stresses on the head of the stent rod, the following reasoning was taken into account when determining the programming algorithm: Due to the static forces applying the endoprosthesis rod, especially in the area of the head that comes in contact with either the hip joint or the hip joint head joint, the problem is to determine both the bending tension in the maximum risk area (regarding rupture of the rod), as well as the static loads acting on the head of the prosthesis rod. It is assumed that the risk area according to the maximum load would be at a distance b from the center of the rod head. For the most unfavorable case, it was taken into account that although the cross section through the rod increases towards the head area, the force acting on it would give maximum bending moment in the area - limit of the rod curvature. According to the construction of a standard hip endoprosthesis, it can therefore be considered that this area would be about ¼ from the length of the rod to the head area, to which is added the half diameter of the head. Therefore it can be considered that the length of the arm is $b \cong ¼ * L + ½ * d$, where d means the head diameter. The bending moment acting on the risk area being $M_i = Fs * b$, means that it will be: $M_i = Ns * b$. The bending stress in the risk area is the ratio between the moment and the resistance modulus of the cross section in the risk area ($\sigma_i = M_i / W_z$), and $W_z = l * t^2$ where l and h are the width and the thickness of the cross section. The bending stress calculus equation becomes:

$$\sigma_i = [(2 * G) * ((0,25)*L+(0,5)*d)] / (t*l^2) \quad (1)$$

To determine the stress in the case of walking, as well as in the case of the static regime, the problem of establishing the loads on the prosthesis capsule and on the head of its rod was raised. To find the stresses on the endoprosthesis capsule, in dynamic regime (normal walking, running, etc.) it was considered that, in addition to the weight and reaction forces in the joint, a friction force must be considered. It occurs between the inside of the capsule and the outside of the rod head. Therefore, the equation of the balance of forces acting simultaneously on the capsule while walking is:

$$\sum F_{d1} = N_{d1} + R_{d1} + F_{fd1} \quad (2)$$

where $\sum F_{d1}$ represents the sum of all the loads acting simultaneously, at a certain moment, while walking, on the endoprosthesis capsule; N_{d1} - maximum gravitational force in dynamic regime, which returns to the capsule, while walking; R_{d1} - the dynamic regime reaction at the contact area with the surface of the rod head and F_{fd1} – the dynamic friction force, in the contact area between the capsule and the rod head. On the maximum weight instantaneous force acting on the prosthesis was considered the most unfavorable case, walking with the prosthetic foot. This meant that the entire body weight up to the hip joint area would act (when stepping) on the prosthesis head. On the other hand, however, being a dynamic regime (walking), it was considered that the dynamic force when walking can be at most equal to twice the static force given by the body weight on the prosthesis. Therefore, it would mean that the force in dynamic, walking regime, given by the maximum body weight, at the moment of step, will be $N_{d1} = 2 * \Psi * N_s$, where Ψ represents the factor of multiplication of the load. As a result, it means that the force for normal walking regime, will be given by the maximum body weight, at the moment of step, being $N_{d1} = 4 * N_s$. It follows that $N_{d1} = (1.33) * G$. Regarding the force given by the reaction in the capsule-rod contact area it is thought that the reaction could be at most equal to twice the orthostatic reaction in the contact area of the joint. This means that the reaction for walking will be $R_{d1} = 2 * R_s$. As it is known that the reaction in static regime is equal and opposite to the static gravitational force resting on the prosthesis, it means that the reaction on the go will be equal to $2 * (1/3) * G$. It follows that the walking reaction will be $R_{d1} = 0.67 * G$. The calculated friction force in the contact area for walking, will take into account the force which would appear in static regime (F_{fs}), this being multiplied by the amplification factor specific to the dynamic regime, normal running ($\Psi = 2$) . So $F_{fd1} = \Psi * F_{fs}$. As in the present case the force acting on the contact area is the one given by the maximum weight force, at the moment of step, it means that the static friction force would be $F_{fs} = (0,67) * G$. The friction force on travel in the contact area will be $F_{fd1} = \Psi * F_{fs} = \Psi * [(0.67) * G]$. It means that the friction force on the fly in the contact area is $Ffd1 = 1.33 * G$. By summing all the forces acting as loads on the stent capsule, it results that the total force will be:

$$\sum F_{d1} = [(1.33) * G] + [(0.67)*G] + [(1.33)*G] = 3.33*G \quad (3)$$

Regarding the dynamic, walking stresses on the head of the stent, the problem was posed, first of all, to determine the bending tension in the maximum risk area, from the point of view of the rod breaking. For this, the following hypothesis was taken into account, namely that the dynamic force when walking can be at most twice the force in orthostatism that acts on the head of the prosthesis rod: $F_{d1} = 2 * F_s = (0.67) * G$. As the bending moment is the product between the support force in dynamic regime and the arm up to the risk area, $M_{i\_d1} = F_{d1} * b$, that means $M_{i\_d1} = 2 * F_s * b$. It means that the bending moment will be twice the bending moment in the case of orthostatism. As a result, the bending stress in the risk area will be:

$$\sigma_{i\_d1} = [(4 * G) * ((0,25)*L+(0,5)*d)] / (t*l^2) \quad (4)$$

Using similar reasoning, we proceeded to calculate the loads and stresses on the prosthesis head and for the other dynamic regimes (running, jumping). Starting from the established relationships, we proceeded to program the calculation algorithms, in order to display the results on the resistance of the prosthesis for all cases (orthostatism, walking, running, jumping).

TABLE I. THE RESULTS FOR STAINLESS STEEL

Simulated situation	*Maximum admissible stress on implant's head*	*Maximum admissible bending stress*	*Bending stress in the risk area for the rod*	*Crushing stress implant's head*
orthostatism	120	120	1.34	1.76
walking			2.72	8.74
running			5.4	17.5
jumping			27.06	70.15

In the study, for repeated simulations on the determination of the resistance of the hip implant (different materials), the software application was used, taking into account an adult weighing 90 kg ($G \cong 883\ N$) and 185 cm in height. In tables 1 ÷ 3 it could be exemplified some results in [N/mm^2] obtained after the simulations for different combinations of materials:

TABLE II. THE RESULTS FOR ABS PLASTIC, WIRE STRUCTURE, LOW DENSITY

Simulated situation	*Maximum admissible stress on implant's head*	*Maximum admissible bending stress*	*Bending stress in the risk area for the rod*	*Crushing stress implant's head*
orthostatism			1.34	1.76
walking	7.5	8.0	2.72	8.74
running	7.5	8.0	5.4	17.5
jumping			27.06	70.15

TABLE III. THE RESULTS FOR CORE - FILLED ABS PLASTIC, HIGH DENSITY - SHELL - CARBON FIBER

Simulated situation	***Maximum admissible stress on implant's head***	***Maximum admissible bending stress***	***Bending stress in the risk area for the rod***	***Crushing stress implant's head***
orthostatism	18	21	1.34	1.76
walking			2.72	8.74
running			5.4	17.5
jumping			27.06	70.15

III. RESULTS

Based on the simulations, a first conclusion could be reached, namely that a metal-carbon fiber combination could be a good solution in terms of strength and mass. In order to verify the veracity of this, in another research stage, a finite element analysis of an implant type model was performed, having as metal a metallic material (stainless steel) and as a coating a carbon fiber sleeve. For this purpose, ANSYS was used as a finite element analysis software medium. The finite element analysis, both in static and dynamic regime, invoked the establishment of the determined constraints and loads in the analytical study stage. [4], [5].

It has been determined that, for example, for an adult weighing 90 kg and 1.80 meters in height, in the case of a stainless steel hip stent, its deformity in the case of walking could be about 5-10% higher than in the case of walking. a hip stent with a stainless steel core and a carbon fiber sheath of approximately 0.5 mm. Based on these studies, it was found that TEKA PEEK type carbon fiber has good mechanical properties but also better biocompatibility to be used as an implant material [4], [5]. In order to achieve such an implant prototype, several stages of work were highlighted, the first being the creation and modeling of CAD-CAM of an implant stem prototype, by using CATIA interface. After printing, a unidirectional carbon fiber was arranged, for this purpose an epoxy resin was used for a better adhesion of the material (figure 2). Separately, the acetabular cup representing the head of the implant was 3D printed [4], [5], [11].

Fig. 2. Coating the core in the carbon fiber sleeve with epoxy resin [4], [11]

The rod was placed in a polystyrene profile after which carbon fiber and epoxy resin were applied. After 24-48 hours, the final result was obtained, presented in figure 3 [4], [11].

IV. CONCLUSIONS

By disseminating important research steps in the paper, the novelty of the proposed solution stems from the fact that it has been proven that the combination of conventional materials with unconventional lightweight materials could involve a number of advantages over hip stents. A first advantage would be that the combination of unconventional light materials, such as carbon fiber with conventional metals such as stainless steel can be done successfully, depending on the composition of a hip prosthesis. Another advantage would be that a carbon fiber coating could interact much better with living tissues, in terms of increasing biocompatibility [9], [11], [12].

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