Modelling of the mechanical behavior of a polyurethane finger interphalangeal joint endoprosthesis after surface modification by ion implantation

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Abstract. Production of biocompatible implants made of polyurethane treated with plasma is very perspective. During plasma treatment the surface of polyurethane acquires unique physic-chemical properties. However such treatment may change the mechanical properties of polyurethane which may adversely affect the deformation behaviour of the real implant. Therefore careful study of the mechanical properties of the plasma-modified polyurethane is needed. In this paper, experimental observations of the elastic characteristics of plasma treated polyurethane and modelling of the deformation behaviour of polyurethane bio-implants are reported.

1. Introduction
The problem of creation of biocompatible implants is raised in many papers [1-5]. Variety of materials is used to create them: metals, polymers and ceramics. All of these materials are suitable for various classes of prostheses. In this study will be considered the usage of polymers that are suitable for creating soft tissue implants inter alia hearing implants, breast implants, heart valves, interphalangeal prosthesis, etc. [6-12]. The most commonly used polymer in soft tissue implantology is silicone [13, 14]. However, this polymer has a number of disadvantages: not all polymer degradation products are removed from the body and some accumulate there. It may lead to adverse consequences for the patient. Also silicone is bioinert material and bioinertness is sometimes not enough for optimum functioning of the prosthesis in the body. The introduction of the implant into the body starts the process of encapsulation, which eventually leads to the formation of calcified shell around the implant. Polyurethane has unimportant advantage over the silicone: being bioinert it can be subjected by ion-plasma treatment [15, 16]. As a result, the treated surface becomes active and protein adheres to it [17, 18]. Correctly adsorbed protein not only becomes strongly bound on the surface but will retain its native conformation [19]. In this state protein is able to serve as a link between the foreign body (implant) and the living organism. In other words, the biocompatibility of the implant in the organism will be achieved.
2. Formulation of the problem and method of solution

At high nitrogen plasma fluence (approximately \(10^{16}\) ions/cm\(^2\)) and 20 keV ion energy a carbon layer about 100 nm thickness is formed on the surface of the material [21]. Since the plasma treatment provides the surface of the material with unique physical and chemical properties, there is reason to believe that the treatment may also have an impact on the mechanical properties of the material.

Additionally, the presence of a carbonized layer on the surface presents a number of problems. The adhesion between the polyurethane and the carbon layer is extremely high because the layer is prepared by modification of the polyurethane surface layer. The carbonized layer in this case is very fragile. Studies using atomic-force and optical microscopy show that even a slight deformation of the polyurethane after treatment leads to cracking of the carbonized layer [22]. Since the adhesion between the layer and the material is high the carbon layer does not peel off from the surface after cracking but continues to be there as a set of small fragments of varying size and shape. The problem is that during deformation of the material the space between the fragments is a strain concentrator.

This can lead to the formation of micro-cracks between the fragments that affect the material’s macro strength. Furthermore, cracks open an area of unmodified polyurethane which is not covered by a protein and the cells of the immune system recognizes these areas as a "foreign body" and activate the foreign body reaction [20]. From this it follows that it is necessary to design such a shape of prosthesis which is close anatomically and in addition to minimize the deformation on the surface of the prosthesis.

This paper presents an experimental study of the mechanical properties of polyurethane-treated by plasma. Also the calculation of the stress-strain state of areal interphalangeal joint prosthesis is conducted and the estimation of efficiency of possible forms is held.

3. Experiment

3.1. Experimental study of treated polyurethanes

The study of the mechanical properties of polyurethane was carried out on the dynamic mechanical analyzer DMA / SDTA861e. The dynamic mechanical analysis (DMA) method has been chosen because it can be used to obtain information about not only the elastic, but also the dissipation properties of a material. Five groups of samples were studied: untreated (init) and 4 groups with different treatment fluences \(5 \times 10^{14}\); \(10^{15}\); \(5 \times 10^{15}\); \(10^{16}\). Such fluence values were obtained by treating with nitrogen plasma whilst applying pulsed negative bias of 20 keV to a sample electrode and covering mesh for 40, 80, 400 and 800 seconds respectively. The samples were rectangular strips of dimensions10x4x2 mm. Measurements were made at 1, 4 and 7 Hz frequencies. For each frequency, the amplitude loading was varied from 1 to 10% with 1.5% increments. For each amplitude \(E'\) (storage modulus) and \(E''\) (loss modulus) values were determined.

Analysis showed that characteristics of the material remain practically unchanged with different frequencies (therefore all the results will be given only for a frequency of 4 Hz). This is certainly a positive result since the viscosity is a negative characteristic of the material used for producing implants.

Figure 1 shows the values of storage modulus for samples treated with different fluences. \(E'\) values were found to depend on the deformation as is typical for polymers. \(E'\) dependence on the fluence of the ion treatment is very weak. The results of the measurement differ only within the instrumental error. On the basis of the measurements it can be concluded that the elastic properties of the material do not change after the plasma treatment.

Figure 2 shows the values of the dynamic loss modulus for samples treated with different fluences. It is observed that \(E'\) is about 5 times greater than \(E''\). This confirms the fact that the behavior of the polymer can be described as elastic and we can disregard its viscosity when modeling an object made of this polymer.

Taking into account the results obtained using DMA analysis it was decided to consider polyurethane as a hyperelastic material. The properties of untreated polyurethane are used in the
calculation due to the minimal influence of the ion surface modification on the elastic behavior of material.

![Graph](image)

**Figure 1.** The values of the storage modulus $E'$ for polyurethane samples that were ion-treated with various fluencies

![Graph](image)

**Figure 2.** The values of the loss modulus $E''$ for polyurethane samples that were ion-treated with various fluencies

### 3.2. Interphalangeal joint prosthesis modeling

For interphalangeal prosthesis modeling in a nonlinear-elastic formulation it is necessary to have a curve of the deformation behaviour of the material. To obtain this data an experiment where uniaxial tension was applied to samples of polyurethane was carried out. Ring-shaped samples were used for this testing (as ring-shaped samples eliminate the error in the measurement of deformation associated with slippage of the sample in the grips). The diameter of the ring is 7 cm, and it has a 6 mm width and 2 mm thickness. The loading was carried out at a rate of 0.2\%/min. As mentioned above, the effects associated with the viscosity of this material are negligible. However, such a low strain rate was chosen to completely eliminate the impact of temporary effects on the deformation curve. The results are shown in Figure 3.

Stretching was performed to 180% strain (the sample was not destroyed). The initial modulus corresponding to the linear portion of the curve is $E_{\text{initial}}=0.93$ MPa. The obtained curve was used in finite element calculations software ABAQUS. On the basis of the deformation behavior of the polyurethane sample, the stress-strain state of an interphalangeal joint prosthesis was calculated.

Figures 4a and 4b show photographs of the interphalangeal prosthesis prototype.
Figure 3. Loading curve of the polyurethane sample under uniaxial tension

The prosthesis parts are shown in figure 4: 1- polyurethane working part (performing the role of the joint) 2- bone fixators that are made of a carbon-carbon composite material. Carbon-carbon composites have unique compatibility with bone tissue. Their stiffness and strength properties are sufficiently high so that they are ideally suited for use in this type of prosthesis.

The prosthesis is implanted as follows: the interphalangeal joint is opened from the rear side. The damaged ends of the phalanges are removed; the bone anchor is inserted in the advanced medullary canals. The polyurethane working part is located at the position of the natural joint. The living tissue around the joint must remain at its natural location. That is why form of the prosthesis should be very close to the natural form of the joint.

The original version of the prosthesis is shown in Figure 5.

Figure 5. The scheme of the original version of the prosthesis. a- schematic representation (1- working part, 2- bone fixators), b- finite element implementation

Figure 6. Model of endoprosthesis with a cavity inside the working part. a- finite element model, b-deformed state of bending at an angle of 90° (cut view)

The disadvantage of this geometry is that the working part is too thick and therefore stiff. It was necessary to change the geometry so as to reduce the stiffness of the working part but retain the shape of the prosthesis as close to the physiological version. Additionally, a crease formed at the bending site of the working part when flexed which could compress the living tissue and cause discomfort to the patient during exploitation of the prosthesis. This fold concentrates compressive strain which can
negatively affect the state of the carbon layer. Therefore, the revised implant had the shape shown in Figure 6a.

Boundary conditions were set as follows: the end of one bone holder was fixed rigidly and an angular displacement of 90° was applied on the end of the second end. The results of the model calculations show that the presence of the cavity inside the working part doesn’t solve the problem of the appearance of a crease upon flexure. The maximum tensile deformation on the surface of the implant is approximately 42% (Fig. 6b). It was therefore decided to consider a different geometry of the prosthesis. Figure 7a shows the next model of the interphalangeal prosthesis where the cavity is opened on the rear side forming a cut-out section.

![Figure 7](image1)

**Figure 7.** Model of the interphalangeal prosthesis with a cavity cut into the working part to eliminate the crease and reduce the strain on the surface of the prosthesis. a- finite element realization, b- strained state calculation result

According to the calculation results it can be seen that the maximum tensile stress on the surface of the prosthesis decreased to 24%, and the compressive deformation inside the cut is 28% (which is 6% lower than that observed in the vicinity of creases in the previous model). Furthermore this geometry greatly reduced the stiffness of working part. This geometry eliminates some of the problems associated with high deformations at the sample surface. Reduction of compressive deformation is insignificant but they are not as critical for the normal functioning of the carbon layer as tensile deformations.

However, it should be noted that this cut-out version is not functional. Living tissue surrounding the prosthesis can be clamped inside the cut-out section causing pain to the patient. The problem can be solved by filling the cut-out with soft, hyperelastic and compressible material. In practice, it may be closed-pore polyurethane foam that can also be treated by the plasma to improve the biocompatibility of the prosthesis. Figure 8 shows the calculation of the strain state of this version of the prosthesis. The calculation shows that the maximum tensile strain corresponds to the previous version of the geometry of the prosthesis (24%). The maximum compressive strain in the porous part reaches 91% but this applies to areas under the surface. On the surface strain does not exceed 35% which is not critical to the carbon layer. Thus the introduction of a soft, compressible part solves the pinching problem of the tissue and additionally to reduces strain on the implant surface in comparison with the original version.

**Figure 8.** Calculation of strain state of interphalangeal prosthesis with soft, compressible hyperelastic material part

### 4. Conclusion

An investigation of ion-treated polyurethane using the DMA method showed that the treatment had an insignificant effect on its elastic properties. It should be noted that the DMA analysis is performed at strains that do not exceed 10%. Since the strain on the surface of the prosthesis is not much higher than the strain in the experiment this estimate can be considered to be adequate.
To design a different type of prosthesis working for large deformations more research is needed. Also it is necessary to carry out a series of studies to determine how the strength characteristics change. In this phase of the study a form of the implant which is geometrically close to a natural interphalangeal joint was proposed. Some simple design revisions managed to significantly reduce deformation of the surface. This reduction in surface deformation positively affects the functioning of the active carbon layer.

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