Strain-stress analysis of the new type composite hip joint endoprostheses

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INTRODUCTION

Hip joint endoprosthesis implantation is a very common operation in orthopaedics due to the high frequency of joint injuries and degenerative joint diseases. New construction solutions of hip joint endoprostheses are constantly introduced into the market however there are no new material propositions [1,2]. Currently available metal endoprostheses are typified by great stiffness, which can induce implant loosening caused by bone resorption (stress-shielding). Alternative for metallic endoprostheses are the composite ones. Strict control of mechanical and biological composite properties is possible in wide range because of many different parameters such as: type of matrix and reinforcements, volume fraction and distribution of fibres, surface microstructure.
The aim of the presented work was to evaluate and to analyse the flexural rigidity along the stem length for composite endoprosthesis depending on a type of matrix (carbon matrix – CC 1D and two types of polymer matrix: PSU+CF 1D, PEEK+CF 1D) and carbon fibre orientation (carbon-carbon composites: CC 1D and CC 2D), in comparison with titanium endoprosthesis.
Durability of tested materials was also estimated on the basis of *in vitro* studies (samples incubation in simulated biological environment).

MATERIALS AND METHODS

Polysulfon with carbon fibre composite (PSU+CF 1D 40% w/w) was obtained by injection moulding at the temperature of 340°C (PSU Mn=26000, σ_r = 72.8 MPa, E=2.59 GPa, ϱ=1.24g/cm^3). Carbon-carbon composites (CC 1D and CC 2D, 60% w/w) were obtained from phenolic-formaldehyde resin as matrix precursor, temperature of carbonization process was 1000°C. As reinforcing phase used carbon fibre (CF) FT 300B Toraya, ϱ=1.76 g/cm^3, σ_r = 3.2 GPa, E=235 GPa.
Three-dimensional numerical analysis was performed using the finite element method (FEM) in ANSYS software. For the numerical model, shape and dimensions of the selected cement endoprosthesis were assumed. Three-dimensional 10-noded elements were used for the whole structure, which was modelled as orthotropic material. Material constants used are listed in Tab. 1. Load was applied at the end of the endoprosthesis as shown on Fig. 1.
Material constants and mechanical properties for CC 1D, CC 2D, PSU+CF 1D composites were measured by using universal testing machine Zwick 1435. In this order ultrasonic tests were also performed. For the remaining materials (Ti, PEEK+CF 1D) mechanical parameters were used from literature [3,4].

In *in vitro* studies samples were incubated at the temperature of 37°C in distilled water and simultaneously in the Ringer Fluid made by Baxter Terpol Sp. z o.o. (of composition [g/cm³]: NaCl- 8.60, KCl- 0.30, CaCl- 0.48). Conductivity of distilled water and pH of Ringer fluid were measured.

Material	Young's modulus (EX) [GPa]	Young's modulus (EY) [GPa]	Young's modulus (EZ) [GPa]	Poisson's ratio	Tensile strength [MPa]
Titanium [3]	110	110	110	0.3	680
CC 1D 60% (w/w)	97	17	29	0.2	800
CC 2D 60% (w/w)	44	7	35	0.25	200
PSU+CF 1D 40% (w/w)	55	3	7	0.35	560
PEEK+CF 1D 68% (v/v) [4]	135	7	7	0.38	> 2000

Table 1: Mechanical properties of studied materials.

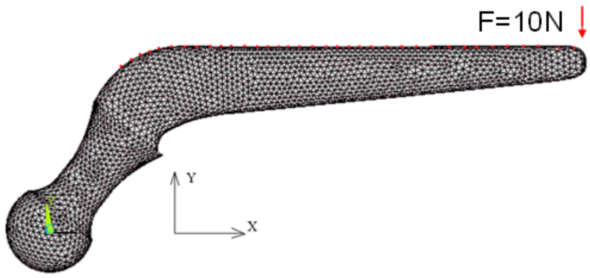

Figure 1: Loading of endoprosthesis and measuring points.

Results

The results indicate that maximum value of the flexural rigidity for composite endoprosthesis is over 10-times lower than for the titanium one (Fig. 2). The influence of matrix type and arrangement of carbon fibres on stress-strain relation was also observed. Maximum value and character of changes of the flexural rigidity are similar for composite with polymer matrices and values of this parameter is lower than for carbon-carbon composites (Fig. 3). Distribution of fibres in 1D direction significantly increases the value of the flexural rigidity in proximal part of endoprosthesis in comparison to 2D CC composite. However, differences between particular composite materials are not as significant as the differences of compared composite materials to titanium. Deformability measured, as total strain is greater for composite endoprostheses, while the values of the Von Mises stress are comparable for endoprostheses made of proposed materials, in comparison with titanium specimen (Fig. 4). However, due to simplified loading of endoprosthesis, stress and strain distribution enables only approximate comparison of analysed materials. The above analysis as well as comparison of strength of tested materials allows the conclusion that the most promising materials are CC 1D and PEEK+CF 1D composites (Tab. 1).
Preliminary evaluation of durability of composites in a simulated biological environment was performed on the basis of *in vitro* studies. In determined observation period no significant changes of Ringer pH or conductivity of distilled water connected with samples degradation were measured (Fig. 5). Changes of mechanical properties in analysed period were also not observed. In the next studies the life-time of selected composites should be tested under cyclic loads in *in vitro* conditions and during longer time of observation.

Conclusions

Maximum value of the flexural rigidity for composite endoprosthesis is over 10-times less than for the titanium one. The influence of matrix type and arrangement of carbon fibres on stress-strain relation was also observed. Composites degradation and changes of mechanical properties were not observed during the observation period. Lower flexural rigidity as well as low von Mises stresses and high deformability of composite endoprostheses gives hope of reducing the complications caused by their implantation.

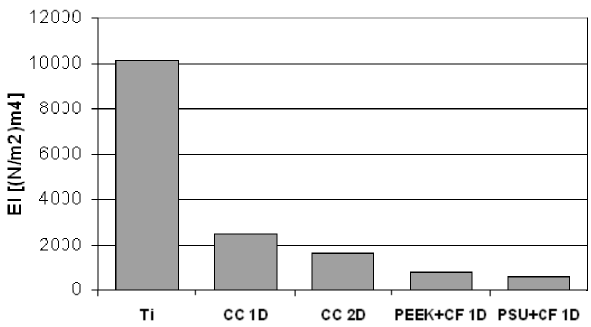

Figure 2: Maximum value of the flexural rigidity in selected point of proximal part of endoprosthesis.

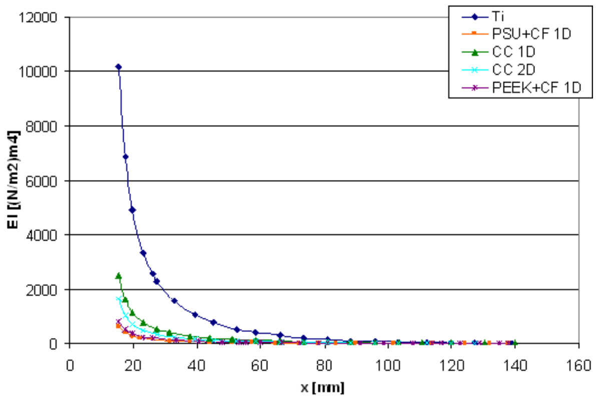

Figure 3: Flexural rigidity distribution (x-distance of particular nodes on upper edge of endoprosthesis stem from the beginning of co-ordinate system, which is located in the middle of endoprosthesis head).

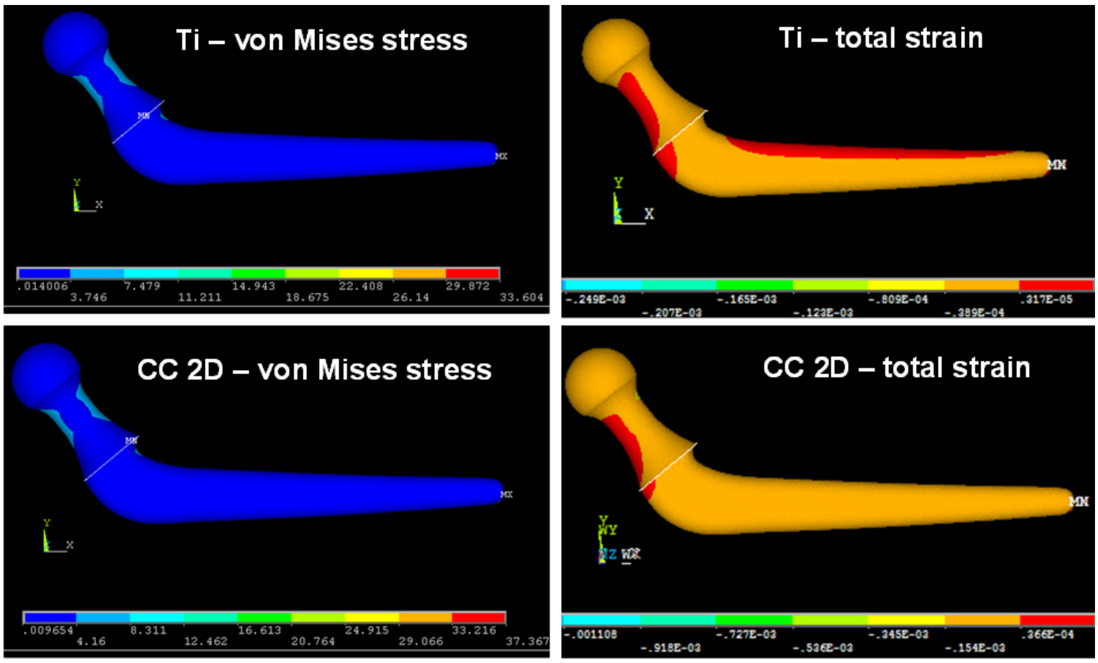

Figure 4: Von Mises stress and total strain distribution for selected materials.

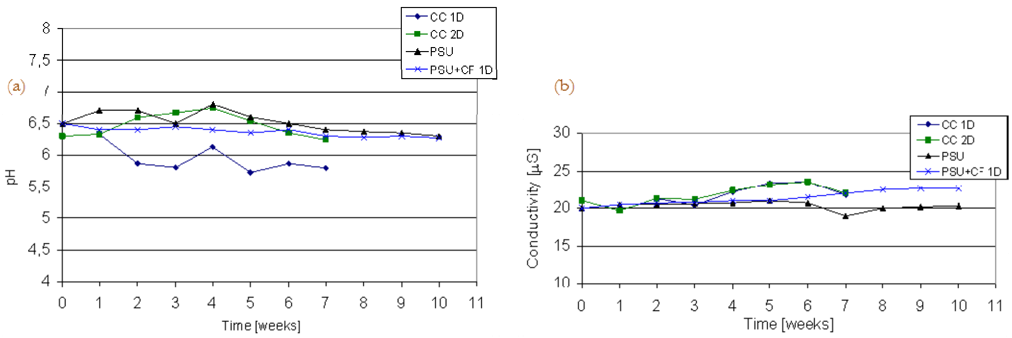

Figure 5: Changes of: a) pH of Ringer fluid, b) conductivity of distilled water, in a function of incubation time.

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