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FRACTURE PROPERTIES OF TITANIUM BONE IMPLANTS CORRODED BY SIMULATED BODY FLUID

LOMNE LASTNOSTI TITANOVIH KOSTNIH IMPLANTANTOV KORODIRANIH S SIMULIRANO TELESNO TEKOČINO

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Keywords: titanium, bone implants, hip prosthesis, body fluid, fracture test, fracture properties, crack tip opening displacement

Abstract

Titanium has been widely used for medical implants and restorations due to its excellent osseointegration, corrosion resistance, biocompatibility in biological fluids, and high resistance/weight ratio.

The aim of this paper is to determine the fracture properties of two bone implants of a hip prosthesis made from commercially pure titanium and Ti-6Al-4V titanium alloy. Both bone implants were before fracture mechanics testing corroded by simulated body fluid. The presence of different microstructures along the pre-crack fatigue front has significant effects on the critical crack tip opening displacement (CTOD). The CTOD values were calculated in accordance with standard ASTM E1290-08e1. This value can be the relevant parameter for the safe servicing of hip prostheses.

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Povzetek

Titan se v veliki meri uporablja za medicinske implantante in obnavljanje že vgrajenih sklopov zaradi dobre trdnosti, korozijske obstojnosti in biokompatibilnosti v bioloških tekočinah in visoke odpornosti glede na njegovo težo.

Namen članka je določitev lomnih lastnosti dveh kostnih implantantov, ki sta vgrajena v protezo kolka in narejena iz komercialno čistega titana in titanove zlitine Ti-6Al-4V. Kostna implantanta sta bila pred lomnomehanskim preizkušanjem korodirana s simulirano telesno tekočino. Prisotnost različnih mikrostruktur na fronti utrujenostne razpoke ima pomemben vpliv na kritično odpiranje konice razpoke (CTOD). CTOD vrednosti so izračunane v skladu s standardom ASTM E1290-08e1. Ta vrednost je lahko relevanten parameter za varno uporabo protez kolka.

1 INTRODUCTION

The high strength, low weight, and excellent corrosion resistance of titanium and titanium alloys have led to an extensive and diversified range of successful applications which demand high levels of reliable performance in surgery and medicine as well as in aerospace, automotive, chemical plant, power generation, oil and gas extraction, sports, and other major industries. More than 1000 tonnes of titanium devices of every description and function are implanted in patients worldwide every year. Requirements for joint replacement continue to grow as people live longer or damage themselves more through strenuous sports or jogging, or are seriously injured in road traffic and other accidents. Light, strong and totally biocompatible, titanium is one of the few materials that naturally match the requirements for implantation in the human body. Medical grade titanium alloys have a significantly higher strength-to-weight ratio than competing stainless steels. The range of available titanium alloys enables medical specialists and designers to select materials and forms closely tailored to the needs of the application. The full range of alloys reaches from high ductility commercially pure titanium used where extreme malleability is essential, to fully heat treatable alloys with strength above 1300 MPa. Shape-memory alloys based on titanium, further extend the range of useful properties and applications. A combination of forging or casting, machining and fabrication are the process routes used for medical products. Surface engineering frequently plays a significant role, extending the performance of titanium several times beyond its natural capability, *[1]*.

Approximately one million patients worldwide are treated annually for total replacement of arthritic hips and knee joints. The prostheses come in many shapes and sizes. Hip joints normally have a metallic femoral stem and head that is placed in an ultrahigh molecular weight low friction polyethylene socket, both secured in position with polymethyl methacrylate bone cement. Some designs, including cement-less joints, use roughened bioactive surfaces (including hydroxyapatite) to stimulate osseointegration, limit resorption and thus increase the implant lifetime for younger recipients. Internal and external bone-fracture fixation provides a further major application for titanium as spinal fusion devices, pins, bone-plates, screws, intramedullary nails, and external fixators, *[2-3]*.

The aim of this paper is to determine the fracture properties of two bone implants of a hip prosthesis made of commercially pure titanium and Ti-6Al-4V titanium alloy, corroded by simulated body fluid. The presence of different microstructures along the pre-crack fatigue

front has important effects on the critical crack tip opening displacement (CTOD) which can be the relevant parameter for the safe servicing of hip prostheses.

2 EXPERIMENTAL PROCEDURE

Bone implants of hip prostheses (Figure 1) are medical devices intended to restore mobility and relieve pain usually associated with arthritis and other hip diseases or injuries. Every bone implant has a distinct set of benefits and risks. The key design features of each implant including size, material and dimensions make each system unique. In addition, the same bone implant system will have different outcomes in different patients. It is also important to recognize that bone implants may need to be replaced after a certain amount of time. Factors that influence the longevity of the device include the patient's age, sex, weight, diagnosis, activity level, conditions of the surgery, and the type of implant chosen, *[3]*.

In the European Union, there are currently five types of total bone implant replacement devices available with different bearing surfaces. These are:

- Metal-on-Polyethylene: The ball is made of metal, and the socket is made of plastic (polyethylene) or has a plastic lining.
- Ceramic-on-Polyethylene: The ball is made of ceramic and the socket is made of plastic (polyethylene) or has a plastic lining.
- Metal-on-Metal: The ball and socket are both made of metal.
- Ceramic-on-Ceramic: The ball is made of ceramic and the socket has a ceramic lining.
- Ceramic-on-Metal: The ball is made of ceramic and the socket has a metal lining.

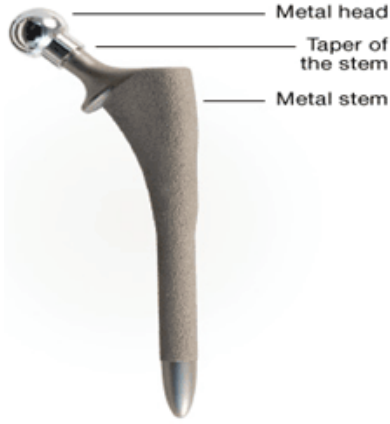


Figure 1: *Bone implant of hip prosthesis*

For fracture mechanics testing single edge notch bend (SENB) specimens were used, and they were loaded in a three-point bending to fracture (Figure 2).

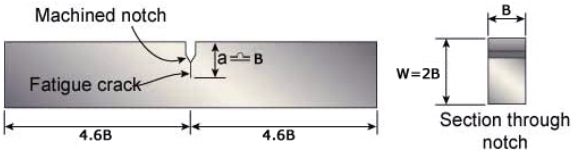


Figure 2: *Fracture mechanics SENB specimens with B=15 mm*

All specimens were extracted from the stem of the bone implant in the direction of loading (Figure 1 and Figure 3) and immerged for 720 hours (30 days) in the simulated body fluid. Ringer's solution (9.0 g/l NaCl, 0.43 g/l KCl, 0.24 g/l CaCl2 and 0.2 g/l NaHCO3) is used as body fluid and maintained in the p_H as 7.4 and a temperature of 37.4 ±1 °C to simulate the body fluid condition, *[6]*.

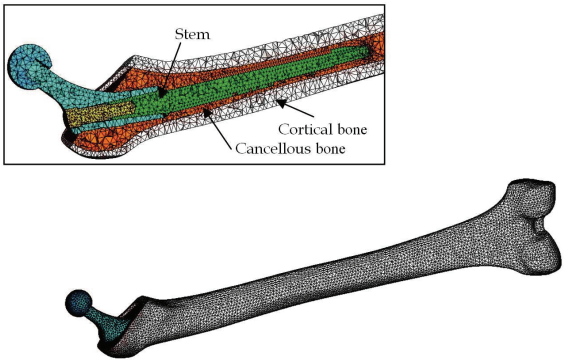


Figure 3: *Fitting of titanium bone implant into the thighbone*

The fracture toughness of fracture specimens made of commercially pure titanium (TiCP) and titanium alloy Ti-6Al-4V was evaluated using a standard static Crack Tip Opening Displacement (CTOD) test, *[4-5, 8]*. Mechanical properties of bone implant materials for this research are shown in Table 1.

Table 1: *Mechanical properties of Ti CP (ASTM F 67) and Ti6Al4V alloy (ASTM F 136)*

Material	Yield strain (MPa)	Tensile strain (MPa)	Elongation (%)
TiCP-grade 4	485	550	15
Ti6Al4V	795	860	10

All CTOD tests were conducted using Zwick and Schenk testing machines. Specimen loading was carried out with constant crosshead speed v = 0.5 mm/min. The test temperature was 37 °C, following the recommendation of the ASTM (the American Society of the International Association for Testing and Materials). For CTOD testing, the single specimen method was used. To evaluate the fracture toughness of titanium bone implants, standard bending specimens, *[4]*, with shallow (a/W = 0.2) notches were used. For all specimens, fatigue pre-cracking was carried out with the Step-Wise High R ratio method (SHR) procedure, *[5]*. Shallow fatigue notches simulate the real cracks the can appear in the stem of bone implants during hip prosthesis operations. Specimens with fatigue notches were immerged for 720 hours (30 days) in the simulated body fluid causing the real corrosion at the fatigue crack front. Titanium implants usually fail as a result of high cyclic loading resulting in peri-implant bone resorption; increased bending moments on implants and eventual metal fatigue and implant fracture.

During the CTOD tests, the potential drop technique was used for monitoring stable crack growth, *[7,9]*. The load line displacement (LLD) was also measured with a reference bar to minimize the effects of possible indentations of the rollers. The CTOD values were calculated in accordance with ASTM E1290-08e1, *[4]*.

3 RESULTS AND DISCUSSION

The critical CTOD was obtained via the clip-gauge displacement *Vg* measured across the notch mouth by using a specific converting equation. In the current ASTM E1290-08e1 standard, the CTOD be calculated from the new equation as follows:

$$\delta = \frac{K^2}{2\sigma_Y E'} + \frac{0.4(W - a)\, V_p}{0.4W + 0.6a + z} \tag{3.1}$$

where the first term is the elastic component of CTOD, the second term is the plastic component, and *Vp* is the plastic component of the clip-gauge displacement. The stress intensity factor for the elastic CTOD calculation is obtained from the following relationship.

$$K = YP / BW^{1/2} \tag{3.2}$$

where *P* is the applied load, and *Y* is the stress intensity coefficient given as a function of the crack length-to-width ratio.

In this standard, the type of critical CTOD was clearly defined according to the nature of the observed fracture event. The four kinds of critical CTOD, i.e. δ_c, δ_u, δ_m and δ_i, are measured (Fig. 4). At low temperatures, the steel fails by cleaving and δ_c is measured empirically.

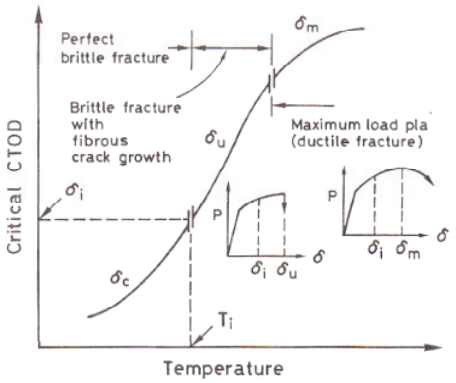


Figure 4: *Definition of critical CTODs in ASTM E1290-08e1.*

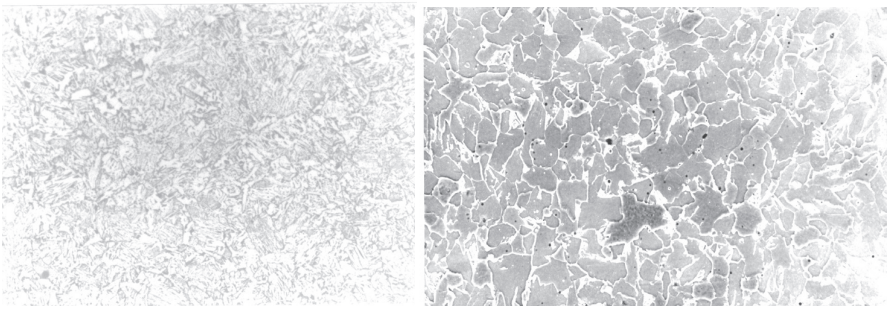
As the test temperature increases, cleavage becomes less favourable, and the fracture toughness increases. The fracture mode changes to micro-void coalescence, and the crack grows in a stable manner. δ_i is defined as the value of CTOD at the onset of tearing. At temperatures slightly above the fracture mode change, stable tearing can be followed by unstable cleavage. In this case, the critical measure is δ_u at the instability point. On the upper level of toughness, the steel reaches a point of plastic collapse when the work-hardening cannot keep pace with the decrease in ligament area caused by stable crack growth. δ_m is then measured at the point of maximum load in a bend test. The real fracture properties and results of fracture toughness CTOD are shown in Table 2.

Table 2: *The real fracture properties of titanium bone implants and results of CTOD fracture toughness at temperature testing at* 37 °C

Specimen SENB	Yield strain (MPa)	Tensile strain (MPa)	CTOD (mm)	Event
TiCP-grade 4	479	542	0,201	δ_m
TiCP-grade 4	465	555	0,234	δ_m
Ti6Al4V	761	834	0,292	δ_m
Ti6Al4V	743	871	0,301	δ_m

All specimens reached the CTOD event δ_m measured at the point of maximum load in a bend test. CTOD fracture toughness of Ti6Al4V specimens is slightly higher than CTOD fracture toughness of TiCP-grade.

Microstructures at the brittle fracture initiation point and around it, as well as the nature of crack path deviation, were evaluated using the fracture surface cross-section method, *[7,8]*, through the brittle fracture initiation point.



a) b)

Figure 5: *Microstructure at the crack tip of a) TiCP-grade 4 and b) Ti6Al4V specimen during* δ_m fracture event

A detailed analysis of material at the crack tip region and along a deviated crack path was made with an optical microscope and scanning electron microscope (SEM). In this way, the critical microstructure (local brittle zones) at the fatigue crack tip surroundings, where the brittle fracture was initiated, and the microstructure where it propagated later, were identified. The SEM analysis of fractured titanium implants reveals consistent uniformity of the microstructure with no indications that major inclusions or porosities are present and refutes the possibility of implant failure due to manufacturing errors.

4 CONCLUSIONS

Bone implants of hip prostheses are medical devices intended to restore mobility and relieve pain usually associated with arthritis and other hip diseases or injuries. Every bone implant has a distinct set of benefits and risks. The key design features of each implant including size, material and dimensions make each system unique. Exact evaluation of real material mechanical properties of the bone implant is essential for the safe servicing of hip prostheses. All fracture specimens reached the CTOD event δ_m measured at the point of maximum load in a bend test. Shallow fatigue notches simulate the real cracks which can appear in the stem of bone implants during hip prosthesis operations. The CTOD fracture toughness of Ti6Al4V specimens is slightly higher than CTOD fracture toughness of TiCP-grade.

The SEM analysis of fractured titanium implants reveals the consistent uniformity of the microstructure with no indications that major inclusions or porosities are present; thus, both titanium bone implants made of commercially pure titanium (TiCP-grade 4), and titanium alloy Ti-6Al-4V can be used for the manufacturing of bone implants of hip prostheses.

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