Chapter 4

Mechanical Properties Comparison of Engineering Materials Produced by Additive and Subtractive Technologies for Dental Prosthetic Restoration Application

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Additional information is available at the end of the chapter

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Abstract

Comparative investigations are presented into the structure and properties of selected engineering materials used for dental prosthetic restorations manufactured alternatively by the subtractive method by milling on computerised numerical control (CNC) milling machines and by the additive selective laser sintering (SLS) method of solid and porous elements using computer-aided design/computer-aided manufacturing (CAD/CAM) techniques; moreover, an original technology was presented of manufacturing the elements used in prosthodontics, produced with titanium and Ti6Al4V alloy powders with the SLS technique. Suitability was confirmed of applying the manufacturing technologies used in prosthodontics from powders by the SLS technique. The results were compared of the executed tests of strength micro-samples and of the selected prosthetic Bridges. The SLS technology for titanium, and even more for Ti6Al4V alloy, ensures the achievement of mechanical properties comparable or better than a reference Co-Cr alloy commonly used for prosthetic restorations, including prosthetic bridges fabricated by milling solid discs with CNC milling machines. For all the examined engineering materials, the milled and then sintered ZrO_2 material exhibits the lowest strength properties. The results presented in this chapter can be directly applied in dental practice.

Keywords: prosthodontics, machining, additive machining, selective laser sintering, titanium and its alloys , Co-Cr alloys, zirconium oxide

1. Justification and scope of comparative investigations of selected alloys and technologies of dental manufacturing of prosthetic restorations

Practical dental engineering encounters numerous problems with selecting the most suitable material and technology of dental manufacturing of a prosthetic restoration by a dental surgeon conducting treatment. One of the important reasons is inability to compare directly the properties and structure of analogous prosthetic restorations but produced using various technologies and from different alternative engineering materials accepted for medical, especially dental, uses as compared to the restorations used most often until now. It is beyond any doubt that dental surgeons are responsible for the strategy and quality of dental treatment. The literature studies carried out as well as the Author's long-term own engineering experience in running medical practice and a dental engineering workshop show that—parallel to clinical experience of a dentist conducting treatment of dental system diseases—the successfulness of such activities also depends on the results of a dental engineer's activities, consisting of the appropriate execution of prosthetic restorations and implants, depending on the selection of the most appropriate engineering material and appropriate manufacturing technology, ensuring, respectively, the required biocompatibility, mechanical properties and corrosion resistance, as well as high production accuracy and aesthetic requirements. Such activities require, therefore, close cooperation of a dentist and a dental engineer having extensive and interdisciplinary engineering knowledge of the principles of construction, technology and science of materials and computer-aided methods of engineering and dental works.

The dental system, especially teeth, is still subject to numerous disease processes, especially tooth decay, so widespread in many countries and practically across the globe [1–4]. This disease has a considerable social range and is a serious medical problem, not only due to local mouth cavity disorders but also due to a high risk of systemic complications stemming from the dental pulp disease, and is a direct cause of a majority of tooth losses among a wide group of people in many countries. The extraction of unhealthy teeth was a dominant practice in such cases for a long time. There may also be inborn causes of missing teeth, for example, due to missing tooth buds. The basic method of producing permanent prosthetic restorations, which is still often used, is to cast metal alloys requiring successive machining, mainly to remove elements of the gate system, and to ensure required surface roughness and dimensional accuracy. Apart from conventional technologies of manufacturing prosthetic restorations including especially powder metallurgy, casting technologies, metallic foam manufacturing, also additive technologies of fabricating solid and microporous materials in medicine and dentistry are exceptionally useful. The potential of additive technologies is much larger than of other technologies and is represented chiefly by opportunities ensured by three-dimensional designing and the related control of the structure, sizes and shape of the materials produced and the repeatability of geometric features of the elements produced, and because no wastes are produced and because the simplicity of the technology is comprised of two key stages only: designing and manufacturing.

Considering the additive technologies employed most extensively in industry, only few have found their application in prosthetics, with selective laser sintering/selective laser melting (SLS/SLM) offering the biggest opportunities [5–15]. This issue is even more important to clarify whether such additive technologies, applied in dental engineering, are competitive for conventional technologies, because original own works [5, 16–36] have been undertaken concerning constructional solutions and fabrication technologies of a new generation of custom, original, hybrid, microporous, high-strength engineering and biological materials with microporous rigid titanium and titanium alloy skeletons manufactured by selective laser sintering, whose pores are filled with living cells. The aim of such constructional and technological solutions is to ensure conditions for natural ingrowth of living tissues in the connection zone of prosthetic/implant elements, with bone or organ stumps, to eliminate the need to apply for patients the mechanical elements which are positioning and fixating the implants. In order to manufacture engineering and biological materials and implant scaffolds, it is required to seek the most advantageous proliferation conditions of living cells inside the pores of a microporous skeleton made of titanium and titanium alloys, because the adhesion and growth of living cells are dependent on the type and characteristics of the substrate. A porous zone requires surface treatment inside pores to improve proliferation conditions of living tissues, which are not advantageous for a substrate made of titanium or Ti6Al4V alloy. It appears that the improvement of proliferation conditions of cells is ensured by a substrate made of fully compatible materials, including TiO_2, Al_2O_3 oxides and a hydroxyapatite, $Ca_{10}(PO_4)_6(OH)_2$, used as coatings of internal surfaces of pores of a microporous skeleton fabricated by SLS. Thin layers inside pores were deposited by two technologies ensuring the uniform thickness of coatings on all the walls and openings of a substrate, that is, the so-called atomic layer deposition (ALD) [19, 37–43], and the sol-gel technology of coating deposition from the liquid phase by the immersion method [37, 40, 44–52]. In order to be able to accomplish the above-mentioned general aims of the investigations, it has to be confirmed that additive technologies, especially SLS, do not lead to unacceptable deterioration of useful properties of the so-produced prosthetic restorations.

Apart from permanent prosthetic restorations made traditionally on a metallic substructure, new technologies of introducing ceramic materials have been launched, including so-called fully porcelain systems made of zirconium oxide and aluminium oxide. For dimensional accuracy and aesthetics as well as biocompatibility and dental strength of prosthetic restorations, the constant development of manufacturing technologies of prosthetic restorations is required along with the evolvement of the engineering materials currently employed in prosthetics. Highly varied conditions of joining the artificially produced prosthetic restorations and implants with a body's natural tissues, mainly in the oral cavity, are driving especially high requirements for the engineering material applied for prosthetic restorations and implants. The reason for this is not only special conditions of using various prosthetic and implantological restorations but also different economic and technological conditions and differentiated availability and level of clinical methods and also important aesthetic reasons. Computer-aided design/computer-aided manufacturing (CAD/CAM) techniques [53–59], allowing to improve dimensional accuracy of the fabricated prosthetic restorations, have become increasingly vital over the last decade. CAD/CAM techniques find their application in dental engineering in machining, in plastic forming, in additive manufacturing technologies and in special technologies and both in metal and ceramics working. Computer-aided design enables, in particular, to produce a digital prosthetic model by scanning with an intra-cavity scanner and also to customise the designing and manufacturing of each prosthetic restoration under constant supervision of a doctor conducting the treatment. Cone beam computed tomography (CBCT), used often in dental surgeries these days as the basic diagnostic tool for comprehensive dental treatment, mainly implantological, prosthetic and endodontic treatment, creates excellent possibilities in this field. A widespread application of CBCT also permits to utilise the information acquired about the condition of a patient's tissues in planning and performing implant and prosthetic treatment [55, 60].

The assumptions made require special interest in the selected engineering materials finding application in dental engineering, including mainly casting alloys of cobalt, titanium and its alloys and ceramics on the zirconium oxide matrix [61–72]. The essential aspects underlying the investigations described here were the questions whether—in each case—it is feasible to apply prosthetic restorations made on a metallic substructure of zirconium oxide, guaranteeing currently best aesthetic results, for prosthetic restorations and to explain if it is possible to use a zirconium oxide substructure in the side sections and in circular restorations together with recognising the limitations for the use of this material. Another practical aspect is to develop a methodology for application of titanium and its alloys as compared to Co-Cr alloys of the Vitalium type for implantoprosthetic restorations so as to guarantee the lowest possible mass of the completely manufactured prosthetic restoration while ensuring aesthetic requirements and the maximum, possible strength properties. For production of metallic substructures, it is necessary to ensure space for veneering porcelain, among others, in intermolecular spaces on abutments between crowns and bridge spans but also on the vestibular space, so that the dark colour of the substructure does not influence the colour of the ready veneering ceramics. Considering that disadvantages and limitations are known of casting and machining technologies such as computerised numerical control (CNC) milling currently used in dental engineering laboratories, it was decided to compare the state-of-the-art manufacturing techniques currently applied in dental engineering, including completely innovative additive techniques which—due to high implementation price, but very low price of producing a single prosthetic point—are rarely used in practice. As Co-Cr alloys are most often used in dental practice and, therefore, most often described in the literature, such were used as a reference point for the works performed already described in literature. This will allow practitioners, especially dentists, to easily transfer the results presented in this article to everyday uses and to consider such results for designing the extensiveness of prosthetic restorations and for cost valuation. The matter is very complex and requires multi-criteria optimisation, which cannot be performed individually by any prosthetic laboratory. The matter requires, therefore, methodologically planned research which can demonstrate—to the practising therapists and prosthetic workshops cooperating with them—the advantages of some material and technological solutions over those commonly in use these days. This will surely have a major impact on the outcomes and durability of dental treatment and on satisfaction, comfort and patients' improved health condition. The aspects highlighted were the reasons for outlining the topic of the chapter presenting the results of the work [18], after making a thorough literature analysis of the status of research into the dental system, its diseases and treatment methods of tooth losses as well as materials and technologies applied in prosthetics. A concept presented in this article is covered by a broader range of works followed in the recent years [5, 7, 16, 17, 21–26, 34, 55, 56, 73–79], in which the author has participated actively. The works concern

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principally the possibility of producing solid and porous titanium parts by additive technology, for use in dentistry and reconstructive medicine.

This chapter depicts the outcomes of comparative investigations into the structure and properties of selected engineering materials used for dental prosthetic restorations manufactured alternatively by the subtractive method by milling on CNC milling machines and by the additive SLS method of solid and porous items using CAD/CAM techniques; moreover, an original technology was presented of manufacturing the elements used in prosthodontics, produced with titanium and Ti6Al4V alloy powders with the SLS technique. The assumptions made in this chapter are focussed on the selected engineering materials finding application in dental engineering, including mainly casting alloys of cobalt, titanium and its alloys and ceramics on the zirconium oxide matrix.

Materials for milling were delivered in the form of discs with the diameter of 98.3 mm and height of 10–16 mm, and the chemical composition, according to suppliers' attestations, is shown in **Table 1**.

Two types of powders with a spherical shape were used, respectively, for selective laser sintering and with the composition shown in **Table 2**, also confirmed with spectral examinations with the energy dispersive spectrometry (EDS) method:

- Titanium powder with Grade 4 and grain size of up to 45 μm, oxygen concentration reduced to 0.14%, while average oxygen content in titanium powders is about 0.5%, the aim of which is to ensure process safety,
- Ti6Al4V alloy powder with the grain diameter of 15–45 μm, for medical applications.

The results of comparative investigations are presented into the structure and properties of the mentioned engineering materials used for dental prosthetic restorations manufactured alternatively by the subtractive method by milling on CNC milling machines and by the additive SLS method of solid and porous items using CAD/CAM techniques; moreover, an original developed

Material	Mass concentration/fraction, %				
Ti grade 2	Ti	Others N, C, H, Fe, O altogether			
	> 99.6	< 0.4			
Ti6Al4V	Ti	Al	V	Others N, C, H, Fe, O altogether		
	89.4	6.2	4.0	≤ 1.0		
Co-Cr	Co	Cr	W	Mo	Si	Others C, Fe, Mn, N altogether
	59.0	25.0	9.5	3.5	1.0	< 1.0
ZrO_2	$ZrO_2 + HfO_2 + Y_2O_3$					Including Y_2O_3
	> 99.0					4.5-6.0

Table 1. Chemical composition of the examined materials manufactured by CAD/CAM by milling solid discs.

Powder	Mass concentration of elements, %									
	Al	V	C	Fe	O	N	H	Others total	Others each	Ti
Ti	–	–	0.01	0.03	0.14	0.01	0.004	< 0.4	< 0.01	remainder
Ti6Al4V	6.35	4.0	0.01	0.2	0.15	0.02	0.003	≤ 0.4	≤ 0.1	

Table 2. Chemical composition of the powders used for selective laser sintering.

technology was presented of manufacturing the elements used in prosthodontics, produced with titanium and Ti6Al4V alloy powders with the SLS technique. Experimental comparative investigations were carried out of selected engineering materials recommended for clinical use for permanent dental prostheses manufactured by CAD/CAM. The properties and structure were also compared directly of analogous prosthetic restorations but produced using the various technologies mentioned and from different alternative engineering materials already mentioned, as well, as compared to the restorations made of Co-Cr alloy by milling solid discs, used most often until now. In order to create a ranking of the materials and technologies applied, analogous elements were produced by milling from solid discs made of ZrO_2 ceramics and then by sintering. Such material and technology are commonly regarded by dentist practitioners as the most avant-garde solution. Elements were produced from titanium and Ti6Al4V alloy by machining by way of milling solid discs, and analogous elements were made from such materials by the additive method in an SLS process. An original technology was established of manufacturing elements used in prosthodontics, produced with titanium and Ti6Al4V alloy powders with the SLS technique.

2. Additive and machining technology of fabrication of materials and dental restorations and research methodology

The elements used for the investigations were fabricated by the machining method by milling solid discs on a CNC milling machine and by the additive SLS method. In the case of the engineering materials manufactured by the machining method by milling on CNC milling machines, technological conditions of manufacturing were chosen experimentally. The technological processes mentioned have no influence on any structural changes, except that ceramic ZrO_2 material of the Cercon type has to be sintered already after milling, so the selection of manufacturing conditions does not have to be verified experimentally and structural changes do not have to be confronted with the mechanical properties obtained of the elements produced. The situation is opposite if the SLS additive technology is applied. Investigations over the selection of technological conditions of selective laser sintering involve the selection of laser power, laser beam diameter, a dependency between the laser beam diameter and the distance between laser beams and the distance between laser remelting paths to desired geometrical features of solid and porous elements produced from titanium and Ti6Al4V alloy and the structure formed depending on such technological conditions and its influence on mechanical properties. Each time before producing the examined elements, design was performed using

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the CAD method, and relevant data in the stereolithography (STL) format is transferred for the execution of computer-aided manufacturing in a relevant technological device.

The solid elements used for the investigations, fabricated by the machining method by milling solid discs, were prepared with special CAD software, dental wings operation system (DWOS), by Dental Wings and 3D Marcam Engineering AutoFab software (software for manufacturing applications) for geometrically simpler shapes of test specimens.

The software features the width offset of the tools used for milling, to ensure the highest possible design reproduction in reality. The so-designed elements were verified by the software for surface cohesion and developed as ready designs in the stereolithography (STL) format. The design files are then implemented to a CAM module, Mayka Dental, by Picasoft. The CAM module enables to position the design prepared on a material disc according to the optimum three-axial milling axis and ensures the placement of the milled elements in a way least interfering with the design and to determine the optimum axis of precision five-axial milling to exploit optimally the capabilities of the milling machine. The so-prepared design is transformed into a machine language, and the file produced is transferred to the milling machine's software.

Another group of the examined elements was manufactured by an additive technology of selective laser sintering by designing a given element virtually, in order to develop a 3D CAD model in the STL format, transferred to the machine's software and then by producing in reality the so-designed element layer by layer until producing a ready product. 3D Marcam Engineering AutoFab (software for manufacturing applications) software was used for model designing as a CAD/CAM tool. The software enables to select model dimensions, constructional features and the type of the model volume filling either as solid or porous and to choose the size of a cell unit making up the entire model. A 3D model featuring a layered structure is produced as a result of designing. The surface of the elements modelled in the STL format, which is accommodated to SLS manufacturing conditions, is shown with a net of triangles, with the accuracy of the surface being represented being inversely proportional to their size.

The model size and structure have to be assumed each time and divided into layers with the expected thickness. The number of layers in a virtual model matches the actually designed number of powder layers when the given element is really produced. A set of "hexagon cross" unit cells was used, selected in a geometrical analysis and as a result of preliminary studies, from the set available in the software. By duplicating them, the entire elements were designed, consisting of nodes and single-lattice fibres, linking the particular skeleton nodes (**Figure 1**).

Fabrication conditions were also selected, at the stage of virtual design of elements, available in the selective laser sintering system, including layer thickness, laser power, laser beam diameter, scanning rate and distance between particular remelting paths. A spatial orientation of unit cells of 45° relative to the axis x of the system of coordinates was also selected experimentally, because other orientations analysed initially, including at the angle of 0°, at the angle of 45° relative to the axis y, at the angle of 45° relative to the axis x and 45° to the axis y, at the angle of 45° relative to the axis y and 45° relative to the axis x, turned out to be less advantageous due to the mechanical properties of the elements produced (**Figure 1**).

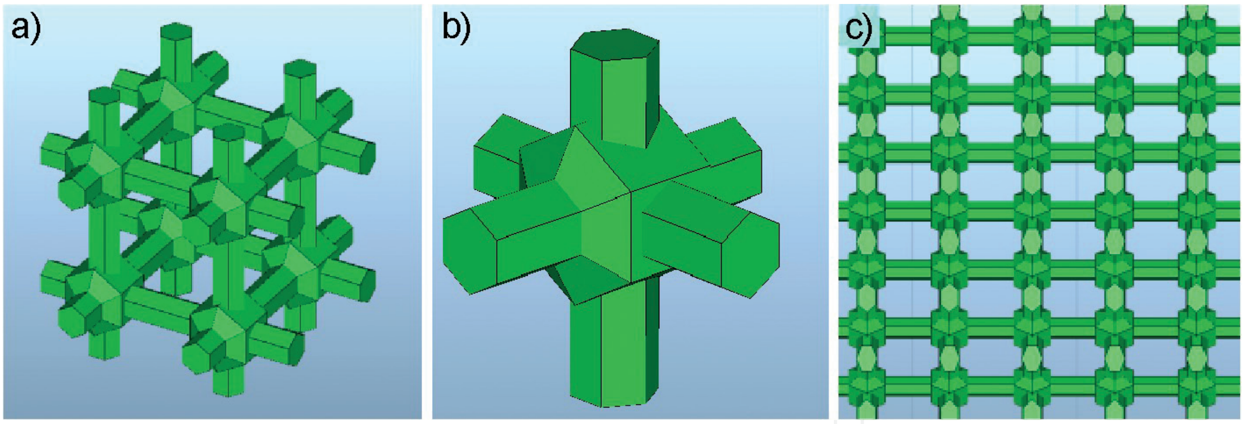

Figure 1. (a, b) "Hexagon cross unit cell"; (c) the arrangement of unit cells in the space of the system of coordinates at the angle of 45° relative to the axis *x*.

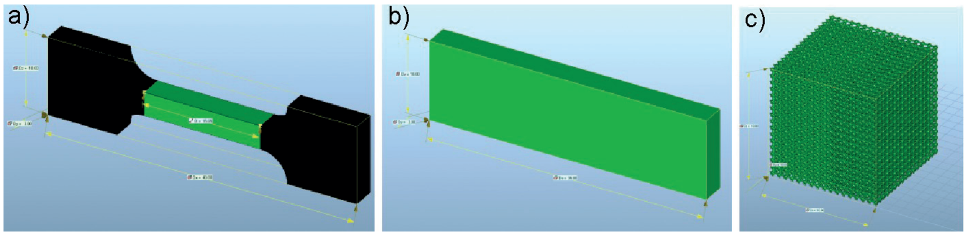

Figure 2. Computer model image of solid samples for static test: (a) tensile test, (b) three-point bending test and (c) compression test.

Micro-samples for static tensile, bending and compression tests were specially selected for the investigations (**Figure 2**), made of all the chosen materials manufactured using, respectively, the additive SLS method from powders with individual CAD/CAM and with the machining technology using CAD/CAM techniques by milling solid discs on CNC milling machines. Micro-samples for tensile strength, bending strength and compression strength tests were also designed for the conditions given above with the pore size of 200–250 μm.

In order to perform bending strength tests, models were created of three-point prosthetic bridges from three different sections of a dental arch (**Figure 3**), including the front, front-wing and wing section, taking into account first of all the highest possible stresses which may occur in such type of prosthetic restorations. They were made as solid from all the examined engineering materials for bending strength tests with the three-point bending method with the both analysed technologies, that is, machining method by milling solid discs on a CNC milling

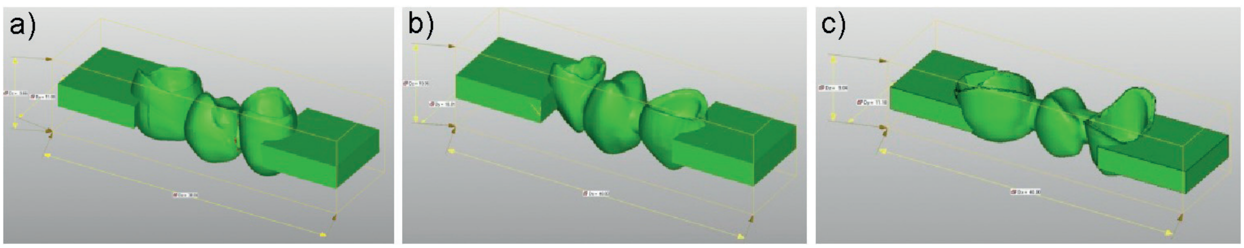

Figure 3. Computer model of bridges nos. 1, 2 and 3 for bending strength tests with the three-point bending method.

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machine and by the additive SLS method. The bending strength was compared to dental bridges made of solid materials, that is, sintered titanium, sintered Ti6Al4V alloy, milled Ti6Al4V alloy, milled Co-Cr alloy and milled and sintered ZrO_2 sinter. The results of strength tests of dental bridges selectively sintered by laser in optimised conditions were compared with the results of tests for bridges milled with CAD/CAM methods from discs of other tested materials, additionally sintered after milling in the case of ZrO_2. The samples prepared correspond to the prosthetic restorations designed in the author's own clinical practice and in a dental engineering laboratory for particular patients and used by them. Geometrical features were taken into account, such as the fitting of a prosthetic restoration to a pillar, of 0.03 mm; the width of the threshold area; the space for prosthetic cement in the occlusal region, of 0.1 mm; the reduction of the crown shape considering the space for porcelain, of 1.5 mm; and the minimum thickness of the substructure according to the material manufacturer's recommendations, of 0.3 mm, for a substructure made of Co-Cr alloys, titanium and its alloys and 0.6 mm for a zirconium oxide substructure. The software features the width offset of the tools used for milling.

The so-prepared micro-samples and bridges were verified by software for surface cohesion and prepared as a ready design in the STL format (stereolithography), processed into a machine language of a CNC milling machine by Yenadent D43W or AM 125 machine by Renishaw for selective powder laser sintering. After designing the shape of the elements produced, short machining of such elements takes place with a CNC milling machine from a disc of the selected material and additional sintering in a resistance furnace for ceramic prostheses (**Table 3**). In the case of additive treatment, after designing the produced element the same as previously, the particular layers of metals and alloys are joined layer by layer in the form of powder in an SLS process (**Table 3**).

As reference results, the results were used of tests of micro-samples and prosthetic bridges fabricated by the machining method with CAD/CAM techniques by milling—with CNC milling machines—solid discs made of Co-Cr alloy of the Vitalium type, used most often in dental treatment, mainly because the alloys are commonly used in casting methods. In order to create a ranking of the materials and technologies used, analogous elements were produced by milling from solid discs made of ZrO_2 ceramics and then by sintering. Titanium elements were fabricated by the machining method by milling solid discs and by the additive SLS method. Elements from Ti6Al4V powder by SLS (**Table 3**) were also produced for comparative purposes. Apart from examinations of solid materials, examinations of porous elements with the pore size of 200–250 μm manufactured by SLS from titanium and Ti6Al4V alloy powder were also carried out for comparative purposes.

Milling from disc on CNC milling machine			**Manufacturing technology**
Co-Cr	ZrO_2	Ti6Al4V	Solid engineering material
Sintered solid engineering material		Ti6Al4V	Ti
Manufacturing technology		Selective laser sintering of powders	

Table 3. Diagram of selection of engineering materials and manufacturing technologies of the examined elements.

3. Comparison of strength properties of the examined solid engineering materials for dental prosthetic restorations manufactured by additive and machining methods

The investigations of mechanical properties of the examined engineering materials were performed in two series in each case. According to the assumed aim of the investigations, mechanical properties were compared in the first series, starting with tensile strength, each time mechanical properties of solid materials produced by the SLS technique for different laser power, that is, sintered titanium and sintered Ti6Al4V alloy. Mechanical properties were compared in the second series of all the examined solid materials, that is, sintered titanium, sintered Ti6Al4V alloy, milled Ti6Al4V alloy, milled Co-Cr alloy and milled and milled ZrO_2 sinter. The results of the examinations were presented in such order for, respectively, tensile strength, bending strength and comprehensive strength.

The impact of laser power, ranging 70–110 W, was investigated in the first place on tensile strength values of sintered titanium and sintered Ti6Al4V alloy (**Figure 4**). The results of tensile strength tests for sintered titanium, sintered Ti6Al4V alloy, obtained in optimum conditions of production, milled Ti6Al4V alloy, milled Co-Cr alloy and milled ZrO_2 sinter, are shown in **Figure 5**.

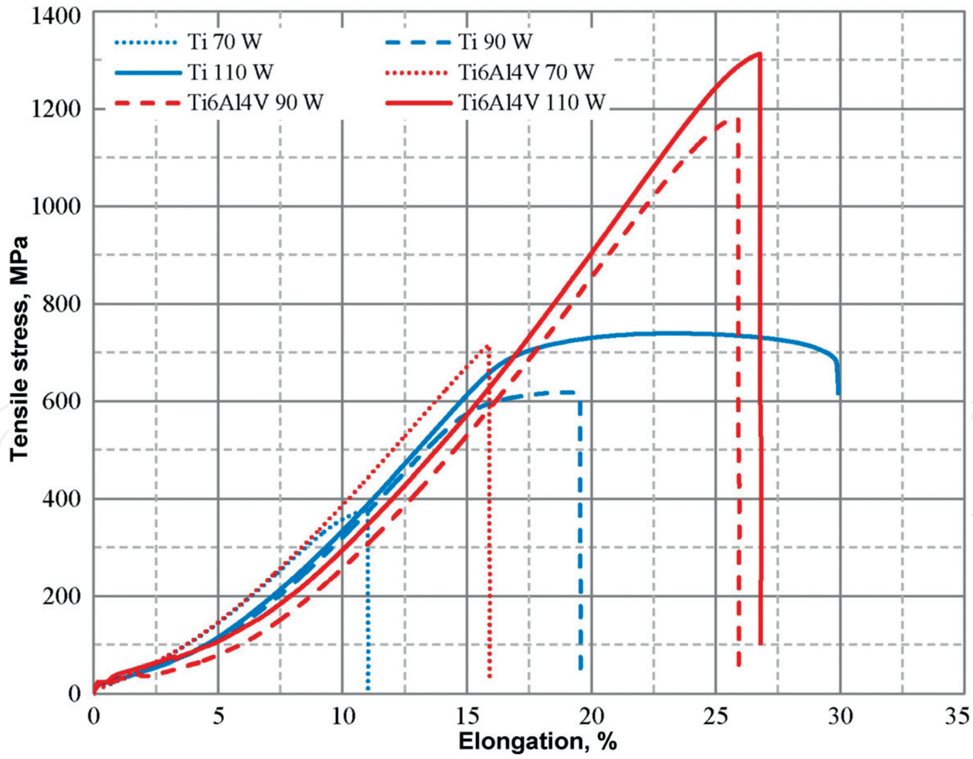

Figure 4. Comparison of diagrams of dependency between tensile stress and elongation for solid samples made of Ti6Al4V alloy and pristine titanium sintered at different laser powers.

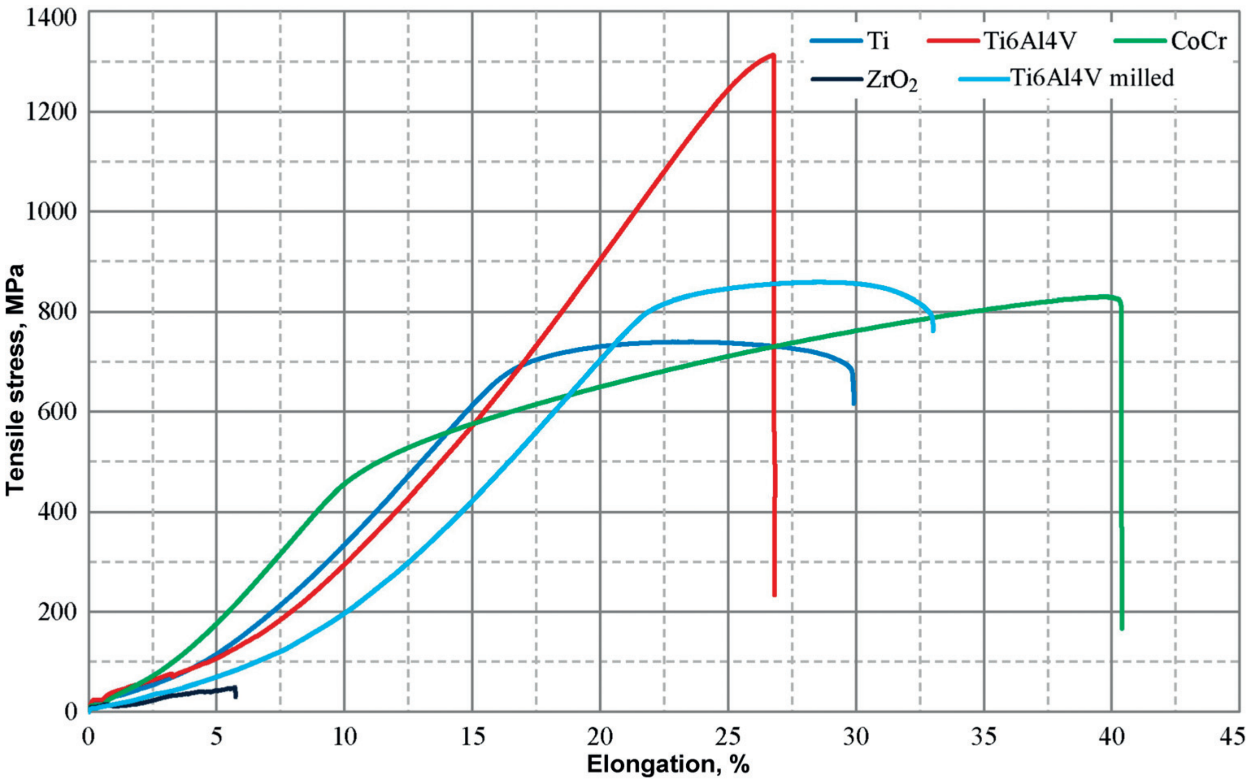

Figure 5. Comparison of diagrams of dependency between tensile stress and elongation for various examined materials, that is, for sintered Ti, sintered Ti6Al4V alloy, milled Co-Cr alloy, milled ZrO_2 sinter and milled Ti6Al4V alloy.

The test results given in **Figure 5** are an interesting reason for application of the tested materials in dental prosthetics. A result of approx. 47 MPa for tensile strength of milled and then sintered ZrO_2 alloys is certainly concerning. Selectively laser sintered titanium exhibits the tensile strength of approx. 740 MPa, slightly smaller than specific for milled Co-Cr alloy samples, approx. 825 MPa, and Ti6Al4V alloy, approx. 858 MPa, and much lower than the tensile strength of approx. 1312 MPa of the selectively laser sintered Ti6Al4V alloy. It is the highest tensile strength value for all the examined engineering materials.

Bending strength tests were performed the same as tensile strength tests. The investigations were performed in relation to solid materials manufactured by SLS for different laser power, that is, sintered titanium and sintered Ti6Al4V alloy (**Figure 6**), as well as in relation to the investigated solid materials, that is, sintered titanium, sintered Ti6Al4V alloy, milled Ti6Al4V alloy, milled Co-Cr alloy and milled ZrO_2 sinter (**Figure 7**). It is noted that the bending strength of the sintered titanium of approx. 1682 MPA is only about 3% smaller than the bending strength of about 1740 MPa of the Co-Cr alloy but strongly smaller than the bending strength of approx. 1959 MPa of the milled Ti6Al4V alloy and much higher than the bending strength of approx. 2374 MPa of this alloy after SLS. The smallest is the bending strength of approx. 605 MPa of sintered ZrO_2 alloys milled from discs and then sintered.

Figure 8 is the comparison of diagrams of dependency between compressive stress and deformation for selectively laser sintered Ti samples and Ti6Al4V alloy.

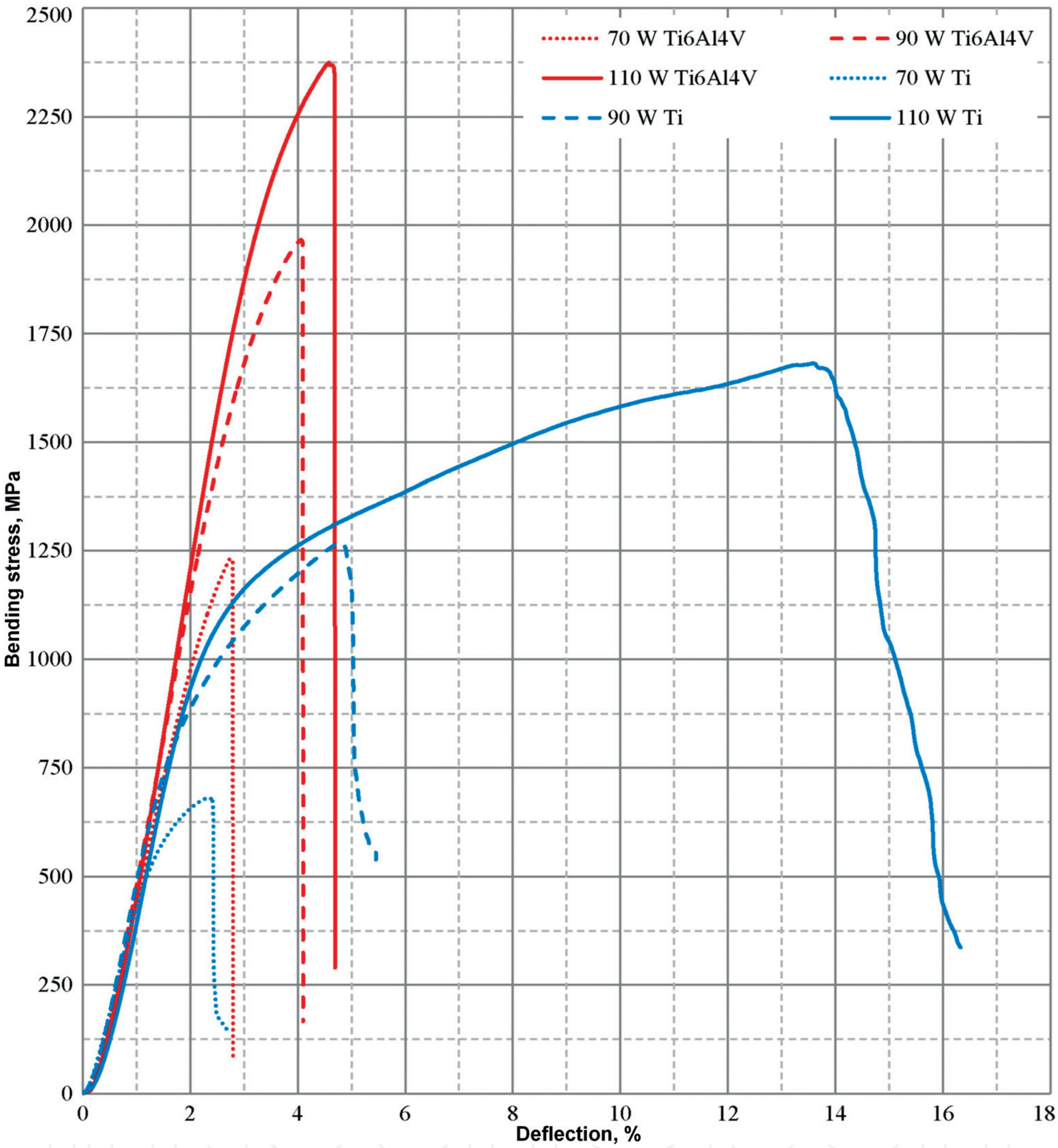

Figure 6. Comparison of diagrams of dependency between bending stress and bending for solid samples made of Ti6Al4V alloy and pristine titanium sintered at different laser powers.

4. Comparison of results of strength tests of dental bridges manufactured from the investigated materials with the additive and machining technology

The results of comparative investigations are presented of mechanical properties of dental bridges manufactured from the investigated engineering materials. The results of strength

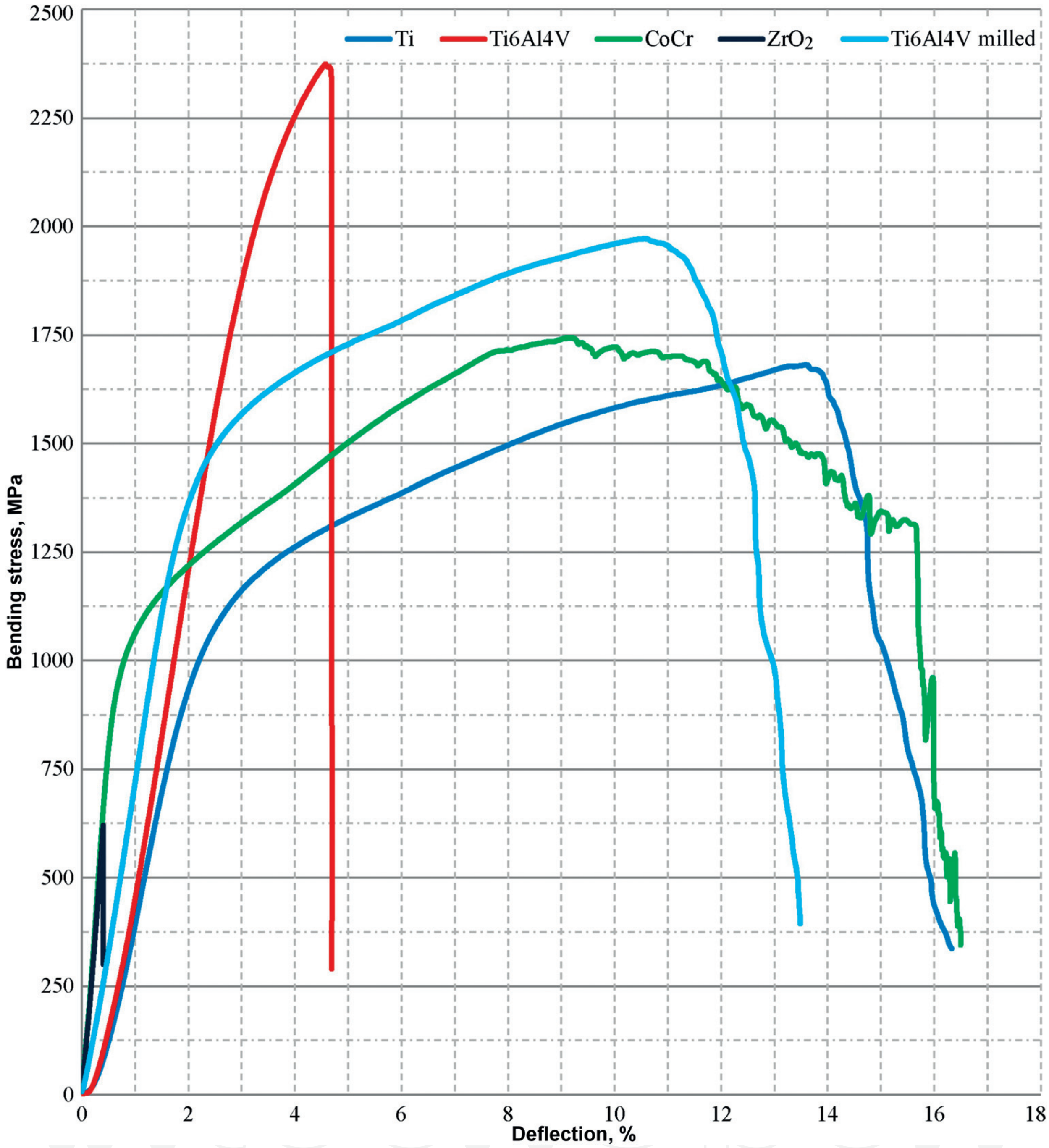

Figure 7. Comparison of diagrams of dependency between tensile stress and bending for various examined materials, that is, for sintered Ti, sintered Ti6Al4V alloy, milled Co-Cr alloy, milled ZrO_2 sinter and milled Ti6Al4V alloy.

tests of dental bridges selectively sintered by laser in optimised conditions were compared with the results of tests for bridges milled with CAD/CAM methods from discs of other tested materials, additionally sintered after milling in the case of ZrO_2. The impact of laser power, ranging 70–110 W, was investigated in the first place on bending strength values of three bridges manufactured by SLS from sintered titanium and sintered Ti6Al4V alloy. The results of bending strength tests of a dental bridge no. 1 produced from titanium and Ti6AlV4 alloy were compared directly in **Figure 9**. Regardless the material produced by selective laser

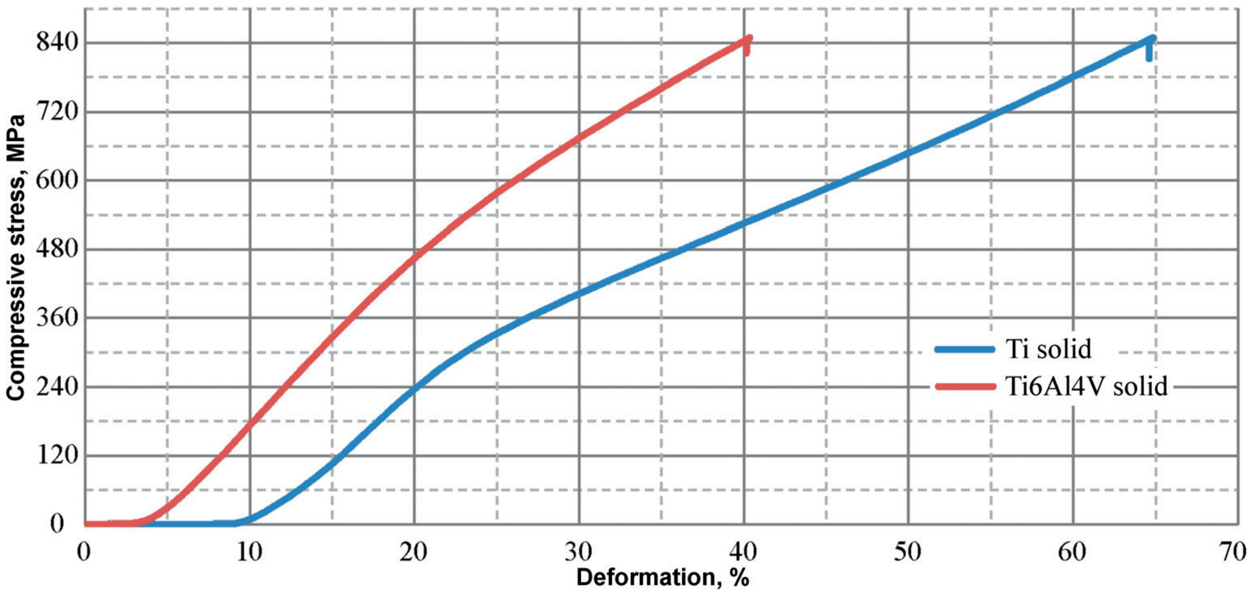

Figure 8. Comparison of diagrams of dependency between compressive stress and deformation for selectively laser sintered Ti and Ti6Al4V alloy samples.

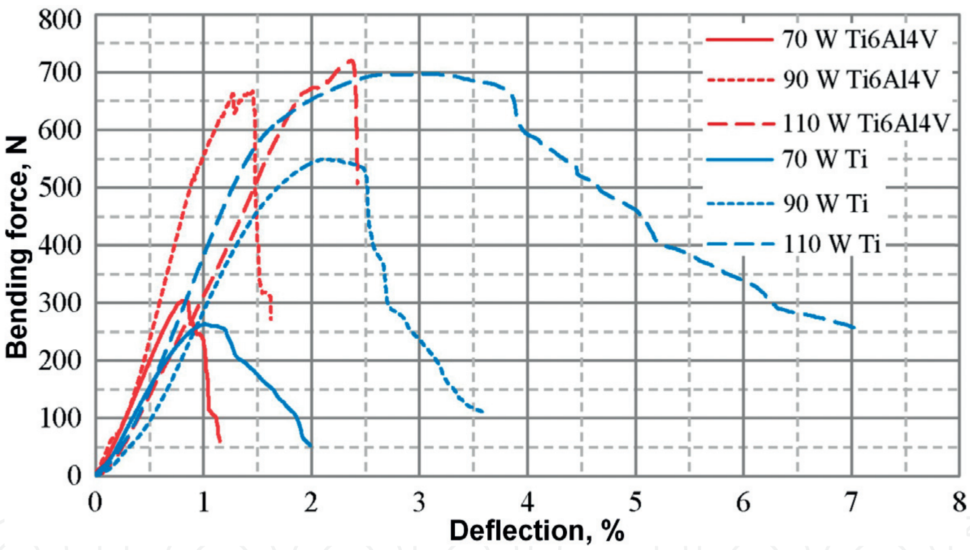

Figure 9. Comparison of diagrams of dependency between bending strength and bending of solid bridge no. 1 manufactured by SLS from titanium and Ti6Al4V alloy for different laser power.

sintering, the maximum bending strength strongly depends on laser power. Sintering with the laser power of 70 W ensures small maximum bending force of approx. 260 and approx. 310 N, respectively, for titanium and Ti6Al4V alloy and is rising almost proportionally to, respectively, approx. 700 and approx. 730 N with laser power increase to 110 W. The maximum bending of the produced dental bridges depends very strongly on the applied laser power and a selectively sintered material. For titanium, this bending—depending on the laser power—varies between approx. 0.9 and approx. 2.4%, when it is much higher for Ti6Al4V alloy and ranges within approx. 2 to approx. 7%, and bending strength is systematically falling as bending is rising (**Figure 9**).

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The analogous strength properties were obtained for a dental bridge no. 2 produced from titanium and Ti6AlV4 (**Figure 10**). In the case of a titanium bridge, if laser power is increased from 70 to 110 W, this has a considerable effect on the value of the maximum bending strength, which increases on average from approx. 350 to approx. 1315 N. In the case of a bridge made of Ti6Al4V alloy produced by SLS, if laser power is raised in the same range, the maximum bending strength rises from approx. 525 to approx. 1560 N. In the case of Ti6Al4V alloy, each time, for each laser power, the maximum bending strength causing sample fracture is higher than for bridges produced from titanium, and this takes place in each case for relatively smaller bending. For example, selectively laser sintered solid bridges with laser power of 110 W from Ti6Al4V alloy are damaged at approx. 3% bending, when bending for titanium exceeds 4.5% (**Figure 10**).

As laser power is increased from 70 to 110 W, the value of the maximum bending strength for dental bridge no. 3, produced from titanium, increases from approx. 240 to approx. 580 N, in

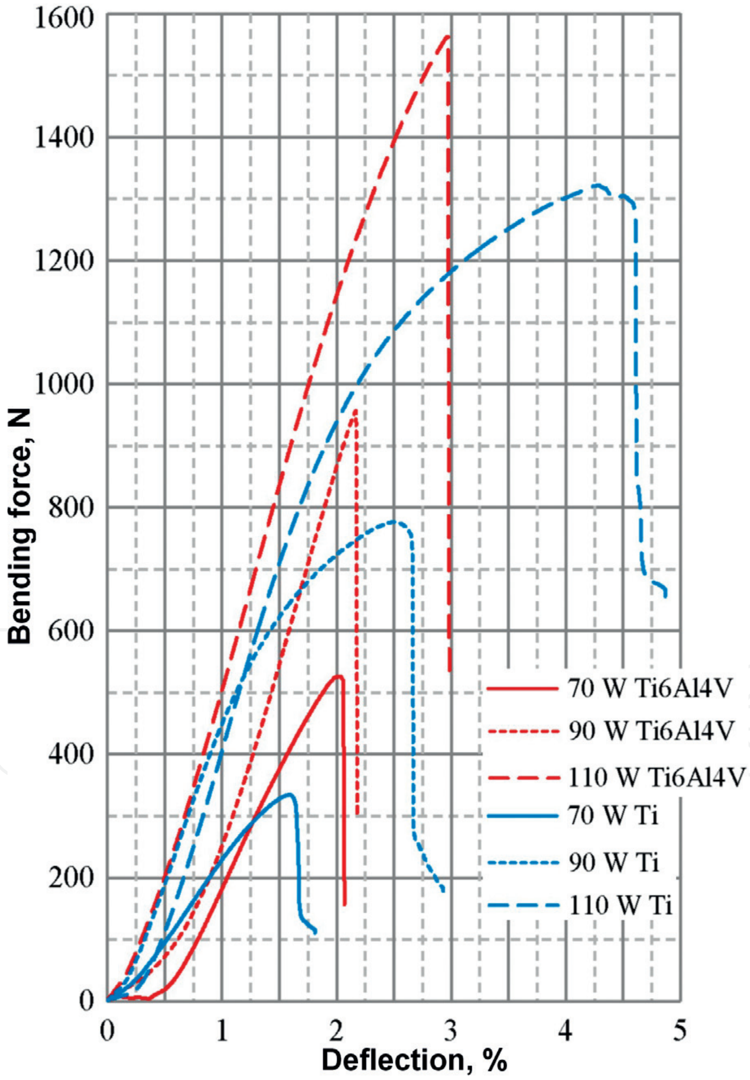

Figure 10. Comparison of diagrams of dependency between bending strength and bending of solid bridge no. 2 manufactured by SLS from titanium and Ti6Al4V alloy for different laser power.

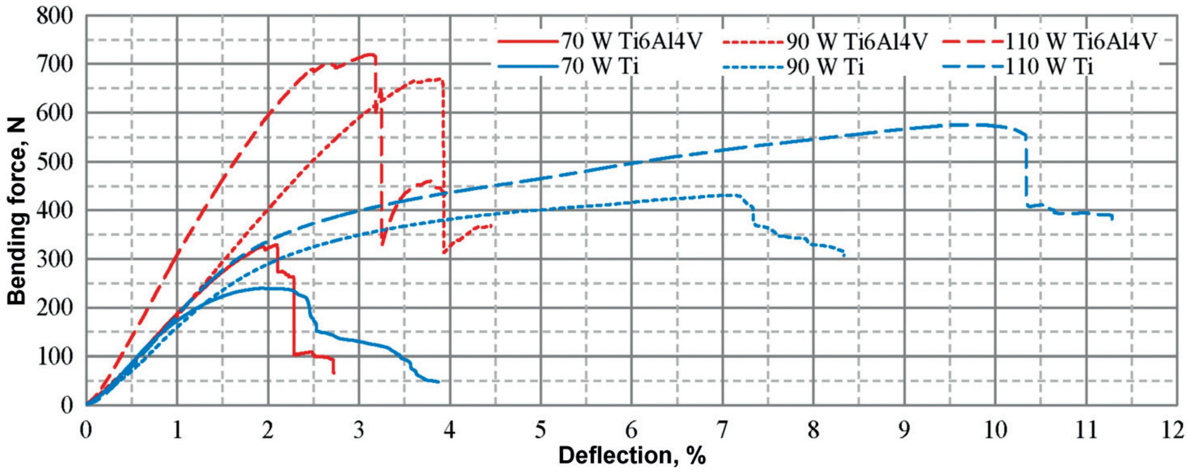

Figure 11. Comparison of diagrams of dependency between bending strength and bending of solid bridge no. 3 manufactured by SLS from titanium and Ti6Al4V alloy for different laser power.

the case of a bridge made of Ti6Al4V alloy, from approx. 340 to approx. 720 N (**Figure 11**). In the case of selectively laser sintered Ti6Al4V alloy, each time, for each laser power, bending strength causing sample fracture is higher than for bridges produced from titanium, and this takes place in each case for relatively smaller bending (**Figure 11**).

Diagrams were compared of dependency between bending strength and bending of solid bridges nos. 1, 2 and 3 manufactured from all the examined materials, that is, from selectively laser sintered titanium and Ti6Al4V alloy and milled from Ti6Al4V alloy discs, Co-Cr alloy and milled ZrO_2 sinter and then sintered (**Figure 12**). The maximum bending strength highly relies on the dental bridge type and varies from approx. 580 N for bridge no. 3 to approx. 1320 N for bridge no. 2. On the other hand, bridge no. 3 exhibits the maximum bending of approx. 10.4% while bridge no. 2 smallest bending of approx. 4.6%. Bridge no. 1 shows indirect properties. In the case of solid bridges nos 1, 2 and 3 manufactured by SLS from Ti6Al4V alloy with laser power of 110 W (**Figure 12**), the maximum bending strength of approx. 1570 N occurs for bridge no. 2, and the two other bridges show the strength of approx. 720 N. Maximum bending for bridges nos 3, 2 and 1 is, respectively, approx. 3.2, approx. 3 and approx. 2.4%. In the case of bridges nos. 2, 1 and 3 manufactured by CAD/CAM by milling discs from Ti6Al4V alloy, as presented in **Figure 12**, the maximum bending strength is much higher and is, respectively, approx. 2970, approx. 2030 and 1030 N, and maximum bending is, respectively, approx. 5, approx. 3.4 and 5.6%. Diagrams were also compared of dependency between bending strength and bending of solid bridges nos. 1, 2 and 3 manufactured by CAD/CAM by milling discs from Co-Cr alloy and ZrO_2 sinter (**Figure 12**). The comparison of diagrams of dependency between bending strength and bending of solid bridges nos. 1, 2 and 3 manufactured from all the examined materials, that is, from selectively laser sintered titanium and Ti6Al4V alloy and with the CAD/CAM method by milling discs from Co-Cr alloy and ZrO_2 sinter, shows that the highest maximum bending strengths are specific for bridge no. 2 regardless the material applied, and highest bending is seen for bridge no. 3.

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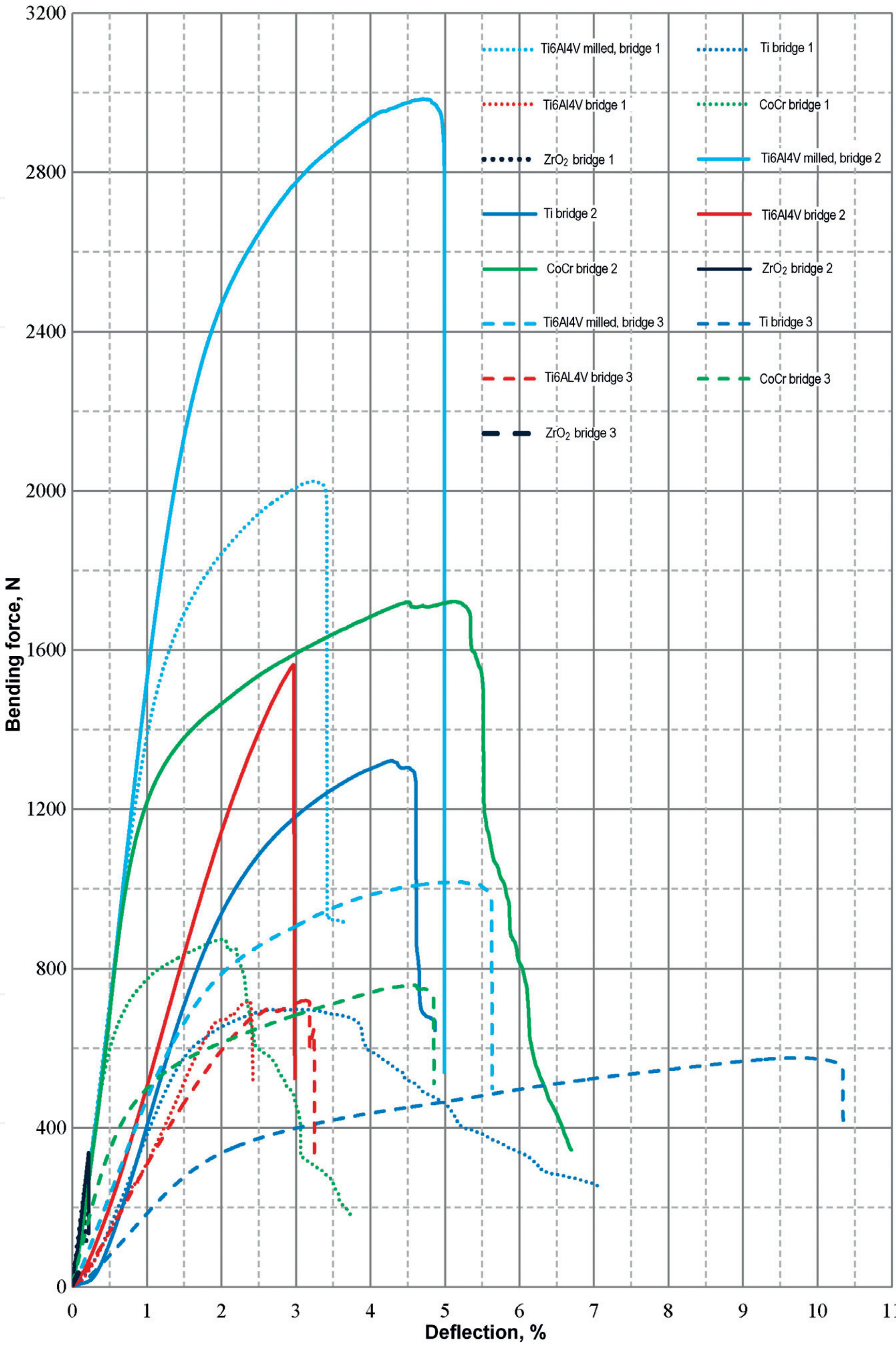

Figure 12. Comparison of diagrams of dependency between bending strength and bending of solid bridges nos. 1, 2 and 3 manufactured from all the examined materials, that is, from selectively laser sintered titanium and Ti6Al4V alloy and with the CAD/CAM method by milling discs from Ti6Al4V alloy, Co-Cr alloy and sintered ZrO_2 sinter.

5. Structure of the investigated engineering materials manufactured by additive and machining technologies

It was revealed in examinations of the structure of the investigated materials in a scanning electron microscope that in the case of Co-Cr and Ti6Al4V alloy, the surface of the samples produced by CAD/CAM, by milling solid discs, exhibits roughness distinctive for precision machining (**Figures 13** and **14**), the same as for ceramic ZrO_2 material which is sintered after performing machining (**Figure 13**). The examinations of chemical composition with the EDS method confirm the presence, in the first case, of Co, Cr and W (**Figure 13**) and in the second case of, respectively, Ti, Al and V (**Figure 14**) and in the third case of Zr and O (**Figure 13**), corresponding to the chemical composition of the materials applied. **Figure 14** shows a surface structure of pristine titanium and Ti6Al4V alloy manufactured by selective laser sintering with

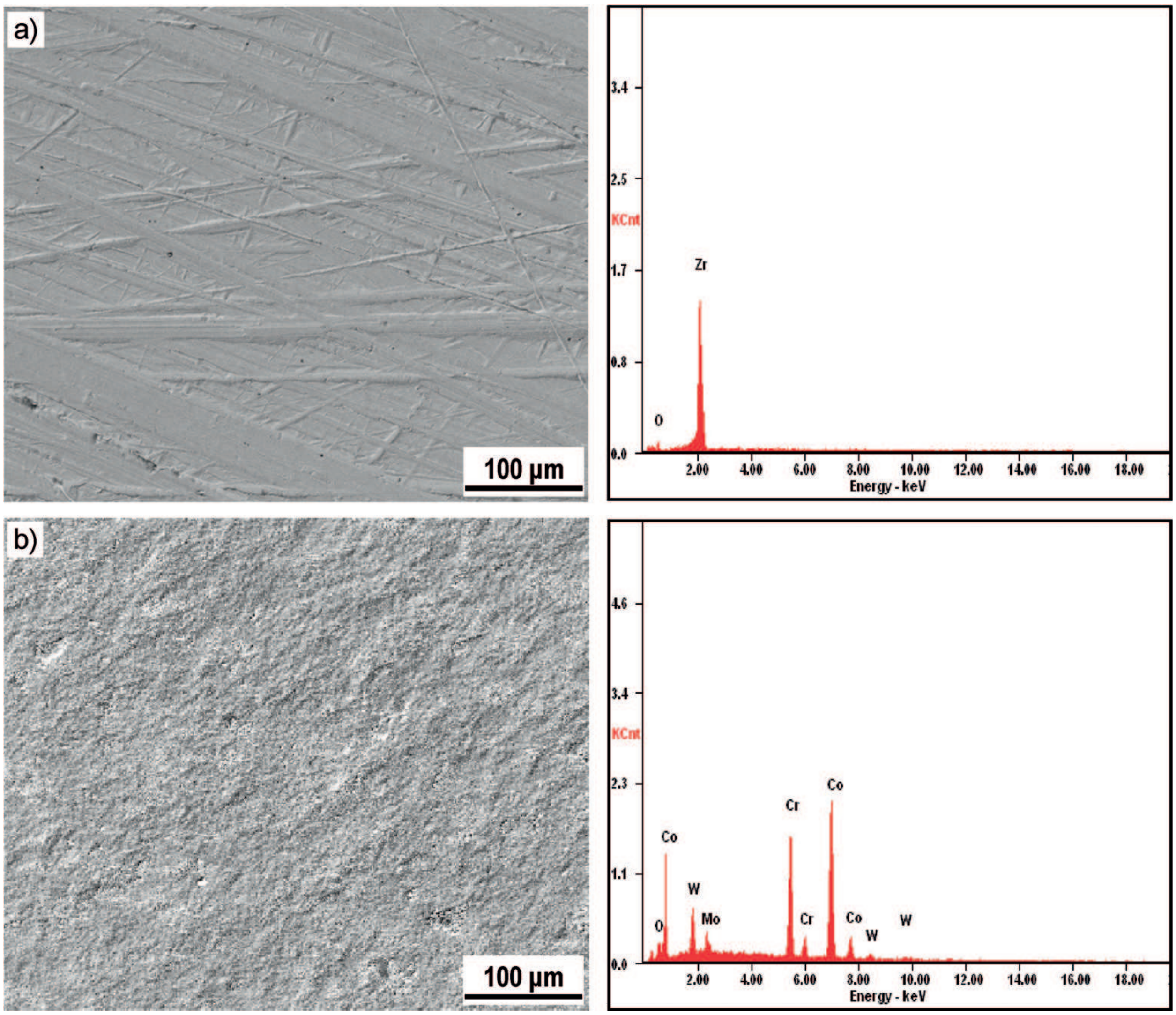

Figure 13. Surface structure of the samples produced by CAD/CAM by milling solid discs with the results of examinations of EDS chemical composition of (a) Co-Cr alloy, (b) sintered ZrO_2 sinter after machining (SEM).

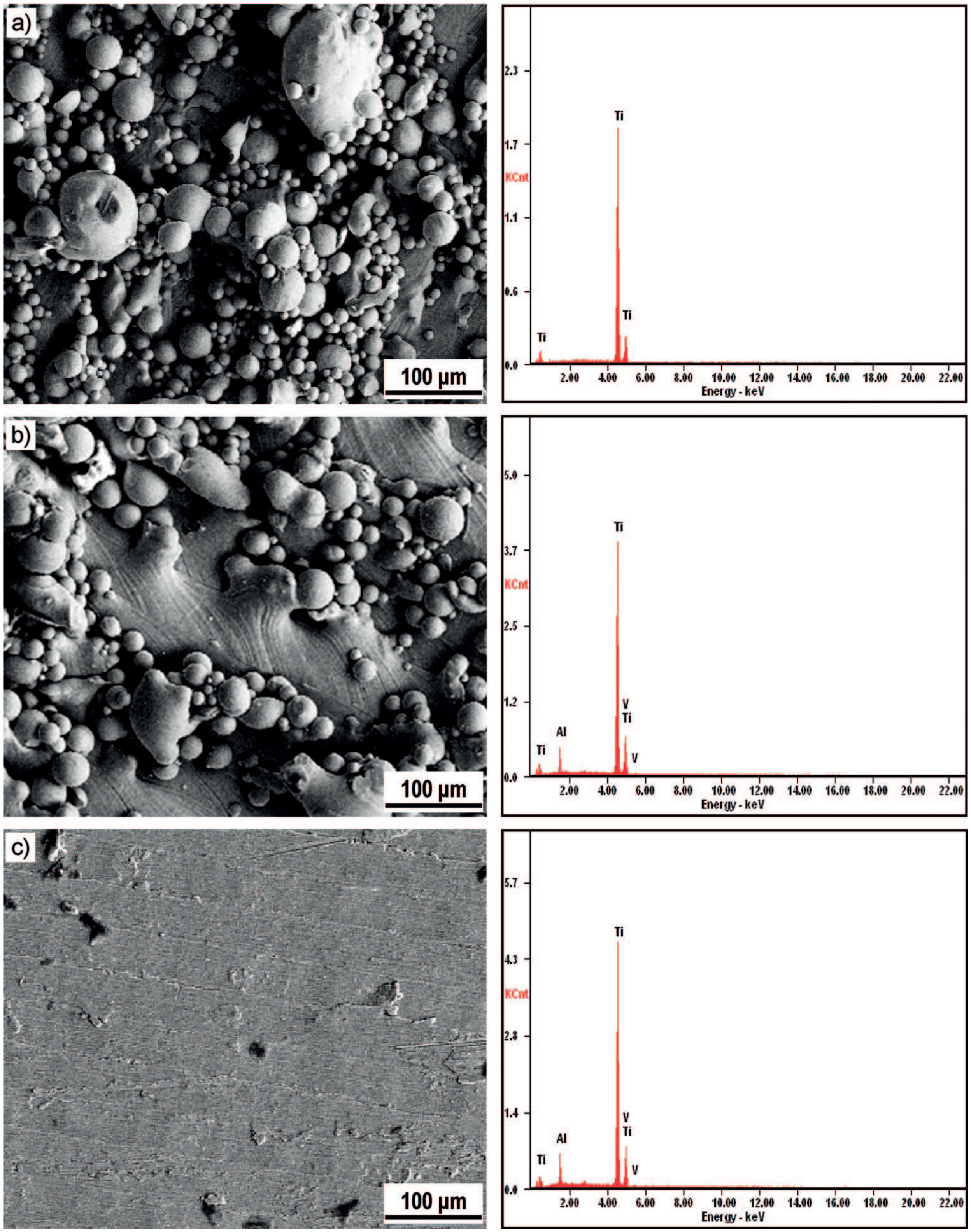

Figure 14. Surface structure of samples manufactured by selective laser sintering with the use of laser beam of 50 μm with laser power of 110 W of (a) solid titanium, (b) Ti6Al4V alloy and (c) Ti6Al4V alloy by CAD/CAM method by milling solid discs with results of examinations of EDS chemical composition (SEM).

the laser beam size of 50 μm and laser power of 110 W. It was revealed in the both cases that apart from the completely sintered materials, with the shown paths of laser beam transition, fine powder particles exist, as well, which—after sintering—should be removed mechanically or by chemical etching.

X-ray tests allow to establish the phase composition of the investigated materials based on the data derived from international files. The following was identified, respectively, in particular samples: Co and Cr in Co-Cr, ZrO_2 alloy, Ti in pristine titanium and $Al_{0.3}Ti_{0.7}$ phase and $Ti_{0.8}V_{0.2}$ in Ti6Al4V alloy irrespective of the manufacturing technology by milling from discs or selective laser sintering.

Investigations were carried out into the structure of the examined engineering materials subjected to the examinations of mechanical properties, and the structure was compared, especially of fractures of the examined materials produced by SLS, that is, sintered titanium and sintered Ti6Al4V alloy, and also manufactured by CAD/CAM by milling discs from solid Ti6Al4V, Co-Cr alloys and ZrO_2 sinter milled and then sintered. The selectively laser sintered materials have a continuous structure without pores, as for titanium (**Figure 15a**), in which

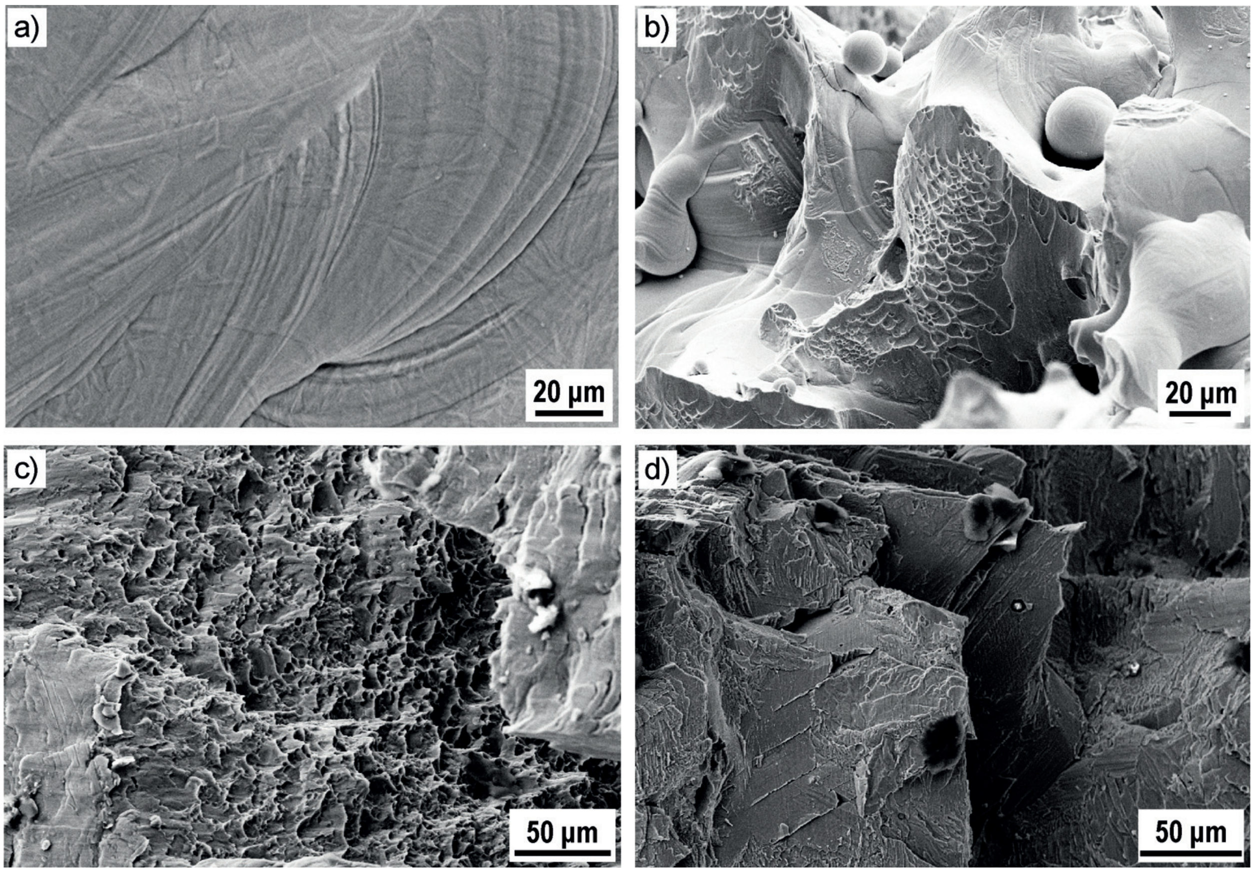

Figure 15. Surface structure of solid samples manufactured by selective laser sintering (a) with the use of laser beam of 50 μm with laser power of 110 W, (b) fracture structure after a static bending test of samples of solid Ti6Al4V alloy manufactured by selective laser sintering with laser beam size of 50 μm after sintering with laser power of 70 W and (c, d) fracture structure after a static bending test of the samples produced by CAD/CAM by milling discs from solid alloys of (c) Ti6Al4V and (d) Co-Cr; SEM.

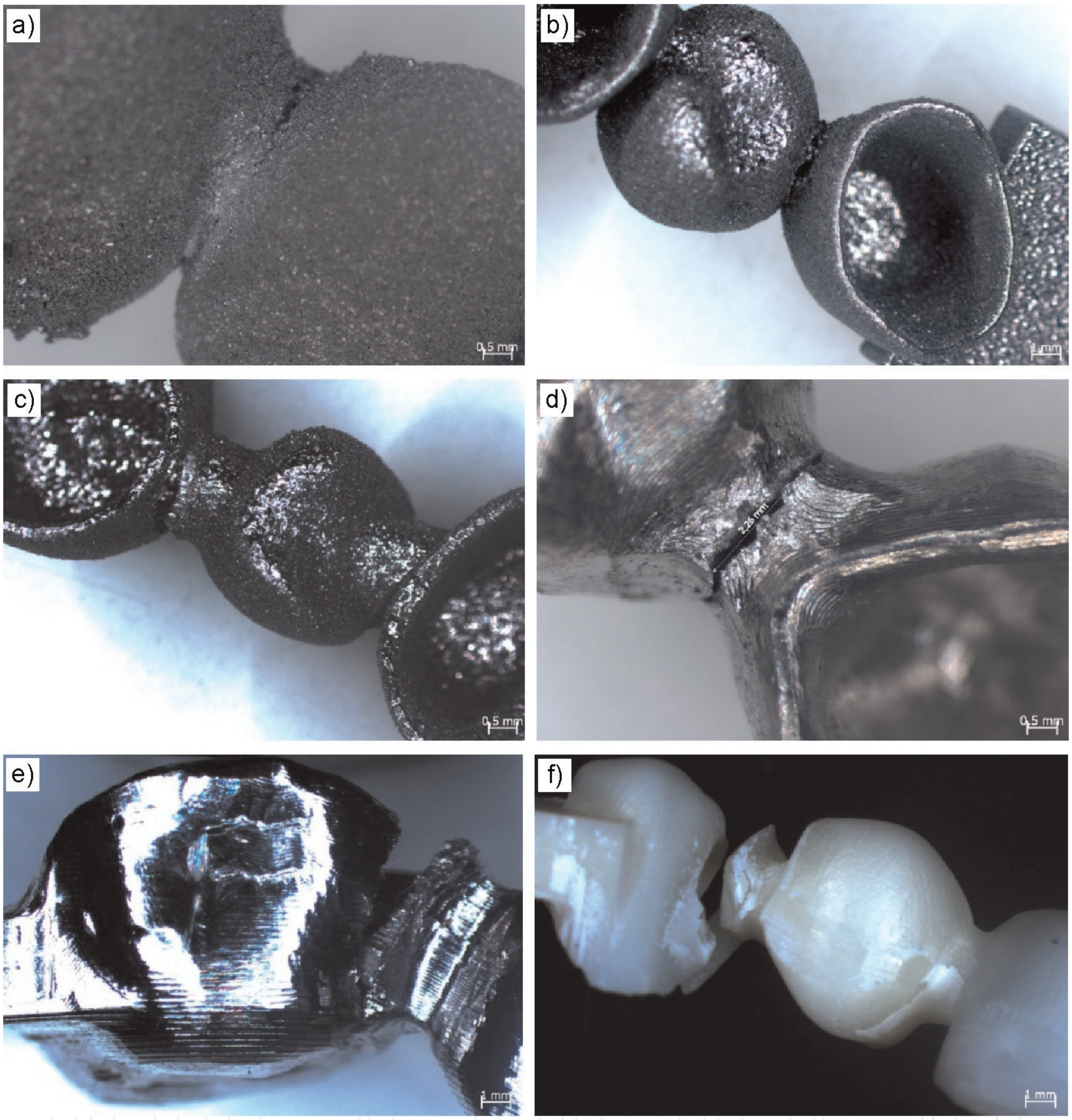

Figure 16. View of dental bridges after static bending test, manufactured from (a,b) titanium by SLS, (a) bridge no. 2 and (b) bridge no. 1; (c) from Ti6Al4V alloy by SLS bridge no. 3; and (d–f) fabricated by CAD/CAM by disc milling (d) from Ti6Al4V alloy, bridge no. 2; (e) from Co-Cr alloy, bridge no. 3; and (f) from ZrO_2 sinter manufactured by CAD/CAM by disc milling and then sintering, bridge no. 3; stereoscope microscope.

particular laser paths can be distinguished, provided that laser power for a laser beam with a diameter of 50 μm is 110 W. It is revealed by analysing a fracture of titanium, laser sintered with the laser power of 70 W, subjected to a static bending test, that—especially—finest powder particles are not sintered fully and also that a ductile fracture occurs in bridges of the sintered material. The samples sintered with the laser power of 110 W are solid and show a ductile fracture. A fracture structure of the samples from the laser sintered Ti6Al4V alloy after a static bending test is analogous to this obtained for titanium (**Figure 15b**) and—after sintering with the laser power of 110 W—shows a ductile fracture. A ductile fracture also exists in the case of fractographic tests of Ti6Al4V alloy but in the form of discs for sample milling (**Figure 15c**). The samples milled from Co-Cr alloy show, however, a brittle fracture, only with the local presence of a ductile fracture (**Figure 15d**).

In the case of all the examined metallic materials, and independently from the applied manufacturing technology, a fracture of samples occurs most often on a root of the abutments linking successive teeth, while in the case of bridges made of ZrO_2, such bridges undergo brittle cracking in different places (**Figure 16**).

6. Suitability of application of additive manufacturing methods of prosthetic restorations of the dental system as compared to machining methods

Comparative investigations are presented in this chapter of the structure and properties of selected engineering materials used for dental prosthetic restorations manufactured alternatively by the machining method by milling on computerised numerical control (CNC) milling machines and by the additive selective laser sintering (SLS) method of solid and porous elements using CAD/CAM techniques; moreover, an original technology was presented of manufacturing the elements used in prosthodontics, produced with titanium and Ti6Al4V alloy powders with the SLS technique. The suitability was confirmed of applying the manufacturing technologies used in prosthodontics from powders by the SLS technique. The results of the executed tests of strength micro-samples and of the selected prosthetic bridges were compared. The bending strength of the titanium sintered selectively by laser, of the milled Ti6Al4V alloy and of the milled Co-Cr alloy is within the range of 1739–1959 MPa and tensile strength is within 739 and 858 MPa. It points out that the materials can be employed as prosthetic restorations, similarly as a selectively laser sintered Ti6Al4V alloy with the bending strength of approx. 2374 MPa and tensile strength of approx. 1312 MPa and pristine Ti with the bending strength of approx. 1682 MPa and tensile strength of 739 MPa. The SLS technology for titanium, and even more for Ti6Al4V alloy, ensures the achievement of mechanical properties comparable or better than a reference Co-Cr alloy which is commonly used for prosthetic restorations, including prosthetic bridges fabricated by milling solid discs with CNC milling machines. For all the examined engineering materials, the milled and then sintered ZrO_2 material exhibits the lowest strength properties. The average tensile strength of this material is approx. 47 MPa, that is, it is nearly 20 times smaller than of Co-Cr alloy, and the average bending strength is smaller nearly 3 times than this of Co-Cr alloy and is approx. 605 MPa. The maximum elongation does not exceed, respectively, 6%, and maximum bending is approx. 0.3%, and the both values are much smaller than such appropriate for Co-Cr alloy. The microscopic examinations of the surface of prosthetic bridges fabricated from this material by the machining method of disc milling and then sintering show more advantageous roughness than bridges produced from Co-Cr alloy, and their fracture after bending is decisively brittle, and cracking takes place in different places, not as in the case of Co-Cr alloy. Prosthetic restorations on a substructure of sintered ZrO_2 are, however, used more and more often, especially because it is necessary to ensure the best aesthetic values of the solution employed.

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As a zirconium oxide substructure can be completely white, but also after colouring with special paints before sintering, it can have a colour consistent with the VITA colour palette, a fully natural aesthetic effect can be achieved of a ready prosthetic restoration. It is also noteworthy that a translucence effect can be achieved, especially on secant edges of such a prosthetic restoration, responding to the most demanding patients' needs. Such effects can be attained on a substructure made of Co-Cr alloy. On the other hand, it is practically impossible to use titanium as a substructure for prosthetic crowns and bridges because the material tends to change the shade of porcelain to darker and colder; hence it is very difficult to ensure the manufacturing repeatability of prosthetic restorations and their quality. A solution ensuring the best possible aesthetic values is to use zirconium oxide as a substructure. Due to a substantial technological and therapeutic risk, a solution with the lowest strength properties is most expensive, though. Therefore, a dental engineer needs to be aware of the limitations of this material and has to design prosthetic restorations on a zirconium oxide substructure differently than when metallic materials are used. The author of this work prepares in his daily engineering practice all kinds of bridges with ¾ crowns, that is, such which are made on all the walls, except for the vestibular wall and the chewing surface, with a full structure, as so-called full contoured. Such a crown is coated from the vestibular part with veneering porcelain to ensure the best aesthetic values, and special paints and glaze are applied onto the other surfaces. The so-designed bridge, in the example shown in this work, would have walls with a thickness of approx. 2.5 mm instead of 0.6 mm, and a section area of abutments would be approx. 20–22 mm^2 instead of 10–12 mm^2, as for the designs presented in this work and essentially relating to metallic materials. A much greater active section will allow to transfer higher loads, accordingly. The designer also has to keep in mind to support bridge spans on the periodontal tissue, never using the saddleback-like gum shape, but using an elliptic support with subgingival correction in each case. The so-designed prosthetic restoration will surely feature much higher strength properties than a prosthetic restoration with a reduced structure, being a reference point used in this work. Unfortunately, prosthetic restorations made on a zirconium oxide substructure with a simplified structure, as so-called caps with simple abutments, are still commonly used by dentists and dental technicians. The test results obtained seem not to confirm the applicability of the bridges, made of this material, designed this way. This puts into question a potential application of ZrO_2 sinter bridges in dental practice without using a special design method of such dental restorations. This fact can be confirmed by cases, known from medical practice, when such bridges are damaged during normal use by patients. The results presented in this chapter can be directly applied in dental practice to remove consequences of teeth lost due to tooth decay.

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