Microstructure and surface investigations of TiAl6V4 and CoCr28Mo6 orthopaedic femoral stems

M Crackau1, K Harnisch1, T Baierl1, P Rosemann1, C H Lohmann2, J Bertrand2 and T Halle1

1 Otto-von-Guericke University Magdeburg, Faculty of Mechanical Engineering, Institute of Materials and Joining Technology, Universitätsplatz 2, 39106 Magdeburg, Germany
2 Otto-von-Guericke-University Magdeburg, Faculty of Medicine (FME), Department of Orthopaedic Surgery, Leipziger Straße 44, 39120 Magdeburg, Germany

maria.crackau@ovgu.de

Abstract. Total hip arthroplasties (THA) achieve very good clinical results and show annually increasing numbers of implantation. Interactions of the bone with the implants surface are of major importance for a stable fixation and longevity of the implant. Therefore, manufacturing of the material and the implants as well as their surface properties can have a decisive influence on the functionality of the implant. The aim of the present study is the investigation of two commercially available femur stems with analytical methods. One stem is made of a TiAl6V4 wrought alloy for cementless application and the other one is made of a CoCr28Mo6 cast alloy for cemented fixation. The change of the production-related microstructure within the implant, differences between surface and bulk properties and potential correlations between the production-related changes to predict failures are addressed. Longitudinal cross sections of tested stems were prepared metallographically, investigated using optical and scanning electron microscopy including EDS and EBSD and correlated with micro hardness depth profiles. Due to production and processing, a subsurface layer is formed in both alloys. The TiAl6V4 wrought alloy stem exhibits a homogenous recrystallization microstructure with fine grains of micrometre size. The subsurface layer of the stem is deformed in parts with embedded corundum particles within the depth of 10 µm. Corundum residues were detected on the entire stem surface and partially covered by the applied calcium phosphate spray coating. The CoCr28Mo6 cast alloy stem shows a dendritic microstructure with chromium- and molybdenum-rich interdendritic precipitations and a surface layer of smaller globular grains in the size of 50 µm to 200 µm. The face centred cubic (fcc) crystal structure was predominantly detected in the phase analysis. The brittle hexagonal close packed (hcp) phase was evident at the implant surface. Similar to the TiAl6V4 alloy stem, embedded residues of corundum particles were detected on the implant surface. This study shows different surface integrities for both stems in comparison to the base material. The observed residues from the manufacturing processes are generally well-embedded into the implant surface, however if released they could impair the functionality of the endoprosthesis as the particles might negatively affect the sterilization process or might reduce metal corrosion resistance.
1 Introduction

Total hip arthroplasties (THA) are successful options for the treatment of joint diseases, such as osteoarthritis or fractures due to trauma. As a standard surgery of modern orthopaedics, artificial hip replacement shows excellent clinical outcomes for the patients with increasing numbers of implantation [1]. In Germany, there are approximately 230,000 primary THA implanted annually [2]. All THA consist of a femoral stem component that is inserted into the femur, a modular head and an acetabular cup, as monobloc or with a liner being placed into the acetabulum. The modular systems allow the reconstruction of the centre of rotation with stable anchorage of stem and socket. Standard metals used for femoral stems are cobalt-chromium-molybdenum alloys (Co-28%Cr-6%Mo cast/wrought alloy ISO 5832-4/-12) with high fracture toughness, wear and corrosion resistance [3] or titanium-based alloys (Ti-6%Al-4%V ISO 5832-3, Ti-6%Al-7%Nb ISO 5832-11) with excellent biocompatibility, high corrosion resistance, low density and a comparatively low elastic modulus [4].

Implant-bone fixation of the femoral stem can either be cementless (press-fit), cemented by applying an acrylic bone cement, or hybrid as a combination of both methods. The fixation type predominantly depends on the patient's bone structure, bone quality, age and activity [5–7]. Cementless fixation has the advantages of lower intramedullary pressure, fewer risk of emboli and hemodynamic disturbances [5], but increases the risk for periprosthetic fractures [8]. Cemented fixation is mainly used in older patients or in patients with minor bone quality [9]. It allows an immediate loading after implantation, and is cost-effective [10]. In 2018, more than 75% of all primary THA in Germany had a complete cementless fixation whereas only 5% were fully cemented [11]. In cementless application, primary implant stability is based on a rigid press-fit macro-lock and secondary stability by bony apposition and growth, known as osseointegration [12]. The biological fixation requires an appropriate implant design and surface finish, such as porous or roughened surfaces [13, 14]. In addition, osteoconductive surface coatings of calcium phosphates, such as hydroxyapatite (HA), are applied. These coatings promote cell attachment, calcification and adaptive bone remodelling by continuous coating degradation [15, 16]. Titanium-based alloys are favourable for cementless fixation since their elastic modulus of 110 GPa is closer to the elastic modulus of bone thus ensuring a better interface stress transfer and lower stress shielding [17, 18]. In the cemented fixation method, bone cement, usually PMMA (polymethylmethacrylat), forms a bond between the cortical bone and the implant and ensures a firm cement-bone connection. PMMA tolerates compressive loads, but is sensitive to tensile or shear stresses that can cause breakage [19]. Therefore, rounded stem designs with polished surfaces are envisioned to reduce stress peaks and overloads in the PMMA and adjacent bone [17, 20].

The interface between implant, cement and bone tissue is of uppermost importance. A permanently firm, load-bearing connection to the bone is decisive for the long-term functionality of the endoprosthesis [15, 21]. Implant weight-bearing causes inevitable micromotion that is associated with failure of the femoral stem fixation due to cement debonding, fracture, wear generation, corrosion and prevention of osseointegration that can lead to the need of revision surgery [5, 22–24]. Common damage mechanisms initiate at the stem surface and are influenced by multiple factors. Manufacturing of titanium- and cobalt-based alloy stems determines the microstructure and consequently the mechanical properties. Defined surface roughness is achieved by post-machining and finishing. Common finishing techniques for stems are shot peening, grit blasting and vibratory grinding to produce a matte, smooth or roughened surface for improved cemented or cementless fixation [25, 26]. Roughened surface modifications promote the subsequent HA coating adhesion that is commonly deposited in a plasma spray process [15]. All finishing processes affect surface near material layers and can cause microstructural changes, metal distortion, contamination and micro damages that alter the material properties. While some studies report an effect of surface contaminants on early implant failure, production-related changes in material and surface properties of stems are not well known [27, 28].
This study investigates the effects of stem manufacturing on the microstructure and properties of a TiAl6V4 wrought alloy stem for cementless application and a CoCr28Mo6 cast alloy stem for cemented fixation. In particular, the following questions are addressed:

- How does the production-related microstructure change with component geometry?
- How do the material properties differ between the bulk and the surface?
- Is there any connection between the production-related change in the properties of the implants and known failure mechanisms?

2 Materials and Methods

2.1 Material and sampling

Two commercially available femoral hip stems were investigated. One stem was made of TiAl6V4 wrought alloy (ISO 5832-3) from the company ImplanTec (ImplanTec Deutschland GmbH (today ARTIQO GmbH), Lüdinghausen, Germany) for cementless application. The arch-shaped ANA.NOVA Alpha stem having a rectangular cross-section exhibits a polished neck and the holohedral, bioactive plasma spray coated calcium phosphate coating (BIONIT®). In contrast, the CoCr28Mo6 cast alloy (ISO 5832-4) stem is intended for cemented application from the company Link (Waldemar Link GmbH & Co. KG, Hamburg, Germany). Rounded surfaces medially and laterally serve to protect the cement bed and improve the implant anchoring. The stem has a collar that counteracts the sintering of the stem. Technical data of the investigated femoral stems are summarized in table 1. Length, width and height (L x W x H), CCD-angle (centrum- collum- diaphyseal- angle) and taper design describe geometrical characteristics of the stems.

### Table 1. Technical data of the investigated femoral stems.

|                       | TiAl6V4 stem                                  | CoCr28Mo6 stem                     |
|-----------------------|-----------------------------------------------|------------------------------------|
| Manufacturer          | ImplanTec Deutschland GmbH                    | Waldemar Link GmbH & Co. KG        |
| Implant model         | modular ANA.NOVA Alpha                        | modular Lubinus Classic Plus       |
| Material standard     | DIN EN ISO 5832-3:2017-03 (TiAl6V4 wrought alloy) | DIN EN ISO 5832-4:2015-12 (CoCr28Mo6 cast alloy) |
| Size (L x W x H [mm]) | 150 x 10-38 x 7-19                            | 150 x 5-20 x 5-10                  |
| CCD-angle[°]          | 127                                           | 126                                |
| Taper design          | 12/14                                         | 12/14                              |
| Surface finish        | porous calcium phosphate coating              | matte                              |

The stems were sectioned perpendicular to the stem axis into 12 samples each with a thickness of 10 mm (figure 1). The TiAl6V4 stem was customized by cut-off grinding with liquid lubrication. For the CoCr28Mo6 cast alloy stem, wire-cut electrical discharge machining (Mitsubishi MV2400S, Mitsubishi Electric, Japan) was used.

![Figure 1. Macro photographs of the femoral stems with marking of the samples. (A) Cementless TiAl6V4 alloy stem with calcium phosphate coating and (B) CoCr28Mo6 cast alloy stem.](image-url)
2.2 Surface roughness
Surface roughness of the implant-bone/cement interface was determined with a confocal microscope (μsurf expert, NanoFocus AG, Oberhausen, Germany). The measuring area of $1.57 \times 1.57 \text{mm}$ was acquired with a $\times10$ magnification objective and a vertical resolution of $20 \text{nm}$. Each stem surface was measured at three different locations (sample T12 and C12, on medial, anterior and lateral side). Besides 3D topography images, the roughness parameters arithmetic mean roughness $R_a$ and average maximum profile height $R_z$ were obtained according to ISO 25178. Based on the obtained data, the Abbott-Firestone curves were calculated and the functional bearing ratio parameters $R_k$, $R_{pk}$ and $R_{vk}$ were determined according to European project report EUR 15178 EN.

2.3 Metallographic preparation
The samples were hot mounted in a conductive phenol resin with carbon additive (PolyFast, Struers GmbH, Dresden, Germany). TiAl6V4 samples were ground with resin bonded diamond discs (MD-Piano, Struers GmbH, Dresden, Germany) of different grits ($125 \mu m$ – $10 \mu m$). For the CoCr28Mo6 samples, a manual grinding process was conducted with decreasing grit ($200 \mu m$ – $8 \mu m$) of abrasive silicon carbide paper under flowing water. Subsequently, a chemical-mechanical polishing process with a mixture of silicon oxide and hydrogen peroxide (35 %) on synthetic fibre for 10 minutes generated a scratch- and deformation-free surface. In order to visualize the microstructure, TiAl6V4 samples were etched in an immersion bath (100 ml deionized water, 5 ml hydrogen peroxide and 2 ml hydrofluoric acid) for 30 seconds. For the CoCr28Mo6 samples, electrochemical etching was applied. The electrolyte was ethanol with 60% perchlorid acid at a voltage of 7 V for 15 seconds (Electrolyte A2 Struers GmbH, Dresden, Germany).

2.4 Optical Microscopy
The optical microscopes Axiovert 200M with the camera system AxisCam MRc5 (Carl Zeiss Jena GmbH, Jena, Germany) and Leica MEF4A (Leica Microsystems GmbH, Wetzlar, Germany) with the camera unit AxioCam 503 were used for optical analyses. Differential interference contrast was applied for a better relief representation of the polished sample surfaces. Three different locations on each sample were examined as shown in figure 2 enabling to compare changes in depth direction (e.g. see A-B) and longitudinal direction of the stem (e.g. see T1-B, T2-B, T3-B).

![Figure 2](image-url)

**Figure 2.** Locations of the optical microscopic examination on cross sections of the (left) TiAl6V4 stem and (right) CoCr28Mo6 stem.

2.5 Scanning Electron Microscopy
The electron microscopic examination was carried out using a scanning electron microscope (FEI Scios DualBeam, Thermo Fisher Scientific, Waltham, MA, USA) equipped with a TEAM Trident system (energy dispersive X-ray spectroscopy EDS, electron backscatter diffraction EBSD, wavelength dispersive X-ray spectroscopy WDS; EDAX, AMETEK GmbH, Weiterstadt, Germany). Secondary electron (SE) imaging and backscattered electron imaging (BSE) were used to show microstructural and surface features. EDS was used to determine the local chemical composition of different phases and
precipitations. The crystal structures were investigated using EBSD regarding crystallographic orientations, grain sizes and phase distribution.

2.6 Microhardness measurements

Microhardness (HV$_{0.05}$) was measured according to ISO 6507-1 using an automated microhardness tester VH3300 (Buehler, ITW Test & Measurement GmbH, Esslingen am Neckar, Germany).

3 Results and Discussion

3.1 Characterization of TiAl6V4 wrought alloy stem

Figure 3 shows a micrograph of the original surface of the TiAl6V4 stem revealing a strongly inhomogeneous coating. The plasma sprayed surface coating exhibits a characteristic crystalline structure, inherent porosity and cracked, deformed as well as chipped-off areas. Different grey shades in the BSE micrograph indicate different chemical elements according to the atomic number. Fine needle shaped structures correspond to the crystalline calcium phosphate (CaP) coating. Bright features are related to the TiAl6V4 substrate not being covered with CaP. The darker features within the red frame in figure 3 were classified in detail using EDS mapping. The results of the elemental mappings verify the features to consist of aluminum and oxygen.

![Figure 3. Electron micrograph of the surface structure of the CaP coated TiAl6V4 stem with elemental mappings of titanium, aluminum, oxygen, calcium and phosphor](image)

Presumably, these particles being partly covered with CaP originate from former processing steps. TiAl6V4 implants are commonly grit blasted in order to generate a macro topography that increases the surface area and improves primary bony fixation. Previous studies showed that an insufficient surface cleaning after grit-blasting resulted in the adhesion of Al$_2$O$_3$ particles, which are commonly used in the grit blasting process [25–28]. Schuh et al. investigated thermomechanical and pH-dependent cleaning techniques to remove contaminant grit particles and highlighted the relevance of a post-cleaning process for implant application [29, 30].
The inhomogeneous CaP coating, embedded Al₂O₃ particles and the surface structure of the TiAl6V4 substrate influence the resulting roughness of the outer osteoconductive surface. Figure 4 is summarizing of the results of surface roughness measurements.

**Figure 4.** Surface of the TiAl6V4 stem with an osteoconductive CaP coating. (A) 3D topographic confocal microscopic image showing macroscopic waviness. (B) Bearing area curve of the measured surface area in A.

The macroscopic surface morphology of the TiAl6V4 stem exhibits regular distributed peak ripple of 100 µm height with a spatial distance of 800 µm (macro roughness). The steep drop in the corresponding areal material ratio curve (figure 4B) demonstrates the resulting pronounced peak height above the core roughness. In addition, microroughness is generated by the porous plasma sprayed CaP coating that exhibits an arithmetic mean roughness $R_a$ of $2.77 \pm 0.43$ µm and roughness height $R_z$ of $16.57 \pm 3.45$ µm. Various studies reported that complex roughness features in macro, micro and nano length scale exhibit an enhanced osseointegration in comparison to polished or uniform patterned surfaces [26, 31, 32]. The measured roughness parameters for plasma sprayed CaP coatings correspond to other studies that report a wide range of $R_a$ value between 2 µm to 9 µm according to coating thickness and pre-treatment processes [13, 33, 34].

The TiAl6V4 stem exhibits a fine-grained and regular microstructure of a globular $\alpha$ phase and a lamellar $\alpha+\beta$ phase, as shown in figure 5A. The average grain size of the homogeneously distributed $\alpha$ phase is smaller than 10 µm. In the SEM image (figure 5B), the lamellae of the $\beta$ phase are shown and exhibit sub-micrometre thickness. Large $\alpha$-phase regions often consist of many smaller $\alpha$ grains adjacent to each other and being separated by their orientation. The structure within the $\alpha$ grains that becomes apparent using SEM is based on orientation dependent responses to the etching of the samples.
Figure 5. Microstructure of the investigated TiAl6V4 stem. (A) Optical micrograph showing two different phases. Light grains represent the globular α phase that is surrounded by a lamellar α+β phase. (B) Corresponding SEM image displaying the fine structure of the α+β phase between α grains.

There are no changes in the microstructure in the longitudinal and depth direction (see Appendix figure A1). The results of the phase analysis by optical greyscale distribution reveal that the majority with 75.3 ± 2.3% comprises of α phase and 24.7 ± 2.3% of the lamellar α+β phase. Local EDS measurements revealed that the α phase exhibits a slightly increased concentration of aluminum (8 wt-%) and a slightly decreased concentration of vanadium (1 wt-%) in comparison to integral measurements with about 7 wt-% aluminum and about 4 wt-% vanadium. In contrast, the lamellar α+β phase shows a slightly increased vanadium concentration (6 wt-%) and a slightly decreased concentration of titanium (88 wt-%).

Cross sections of the corresponding stem surface are illustrated in figure 6 confirming the presence of particles within the stem surface and below the CaP coating.
Figure 6 shows the processing caused imperfections of the stem surface. Blasting induced the generation of a rough inhomogeneous surface with a deformed subsurface within the first 10 µm. Embedded particles of Al₂O₃ that have been detected within the previous surface investigation (see figure 3) can be seen in the cross sections as well. In figure 6 B a fractured particle that is incorporated into the surface is pictured with the corresponding EDS spectrum of the particle showing distinct peaks of aluminum and oxygen. Figures 6 C and 6 D are showing topography and colour-coded elemental mapping (EDS mapping) of a surface near cross section with a high amount of Al₂O₃ particles. The titanium rich TiAl6V4 substrate, several Al₂O₃ particles embedded in the irregular surface structure and a thin layer of CaP can be recognized and separated. Several measurements of Al₂O₃ particles revealed a particle size between 5 µm and 25 µm. The thickness of the CaP coating varies between few microns up to about 15 µm depending on the surface topology and subjacent Al₂O₃ particles. Schuh et al. determined the particle size distribution on different Al₂O₃ blasted titanium hip stems and detected that more than 75 % are smaller than 40 µm [25]. Based on Gburec et al., up to 35 % of the titanium surface can be covered by the blasting grit [35]. The areal amount and size depend on the blasting pressure, grit diameter, process control and cleaning method [25, 26]. The thickness of CaP coating determines the degradation time and consequently the complex dynamic osseointegration process [34]. Recommended layer thickness for the CaP coating is < 50 µm since it provides a good adhesion to the implant, easy resorption and a sufficient strength [34, 36].
In contrast to the surface characteristics of the investigated TiAl6V4 stem such as topology, roughness and present surface composition, hardness measurements did not indicate major deviations. Figure 7 shows a characteristic microhardness depth profile of the surface near zone of 0.7 mm.

![Figure 7](image.png)

**Figure 7.** Microhardness depth profile of the TiAl6V4 stem. Distance delineates the perpendicular offset to the surface of the sample.

There is no significant change in hardness with increasing distance to the surface. The mean hardness of the measurement is $337 \pm 10.5 \text{ HV}_{0.05}$, being in the order of previous general studies of TiAl6V4 wrought alloys [37]. Slight deviations may result from inhomogeneity of the surface and the structure hit by the indenter due to phase composition and grain boundaries. The comparatively low hardness for a distance of 0.05 mm ($329 \text{ HV}_{0.05}$) may be due to an edge effect.

### 3.2 Characterization of CoCr28Mo6 cast alloy stem

The matte surface of the CoCr28Mo6 cast alloy stem shows a macroscopic inhomogeneity. A SEM BSE image illustrates a characteristic surface area in figure 8 A.

![Figure 8](image.png)

**Figure 8.** Surface of the CoCr28Mo6 stem. (A) BSE image of the surface. (B) EDS mapping of cobalt, aluminium, oxygen and silicon.

The fissured surface exhibits particulate inclusions, deformation, scratches and cracks. Different grey scale values in the BSE image confirm variations in the chemical composition. Based on EDS, aluminium and silicon rich particles are present and embedded in the CoCr28Mo6 alloy surface. Based on previous publications, we assume that these particles are residues from the surface finishing applying corundum grit and glass bead blasting to generate a matte surface [26, 30]. Ricci et al. detected similar residues of silicon- and aluminium-rich particles on explanted cemented cobalt-chromium hip implants [27].
The results of the measurements of surface roughness are presented in figure 9.

![Figure 9](image_url)

**Figure 9.** Confocal microscopic measurements of the CoCr28Mo6 alloy stem. (A) 3D topographic image of the matt surface. (B) Bearing area curve of the measured surface area in A.

The matte stem surface is characterized by an irregular surface texture as shown in **figure 9 A**. The corresponding material bearing ratio curve with a low curve inclination represents a pronounced core roughness of the surface that allows a homogenous load distribution. The determined micro-roughness $R_a$ value of $1.12 \pm 0.17 \mu m$ and the mean value of roughness height $R_z$ with $7.18 \pm 1.1 \mu m$ are in accordance to cementless matte surface roughness of other manufacturers [20]. For cementless fixation surface finish and roughness determine the critical implant-cement bond and thus affect mechanical stress transfer, magnitude of post-operative stem subsidence and possible cement mantle damage leading to PMMA and metal wear causing osteolysis [24, 38–40]. Verdonschot et al. emphasize that polished cemented stem surfaces ($R_a < 0.04 \mu m$) are favourable in comparison to unpolished, matte surfaces since the amount of wear debris due to micromotion is reduced [41].

Figure 10 is illustrating the microstructure of the CoCr28Mo6 cast alloy stem. A coarse dendritic microstructure with a considerable amount of interdendritic precipitations can be identified in optical and electron micrographs.
Figure 10. Top: Microstructure of the CoCr28Mo6 cast alloy stem. (A) Overview of a cross section showing a coarse dendritic structure. (B) Micrograph illustrating segmented interdendritic precipitations. (C) SEM BSE image indicating the segmented structure of interdendritic precipitations. EDS mapping of the area shown in (C) revealing the structure of the interdendritic precipitations consisting of either fractions rich in chromium and carbon or increased concentration of molybdenum.

Dendrites with a length of a few millimetres can be recognized over the complete cross-sectional area (figure 10 A). Numerous homogeneously distributed precipitations within the interdendritic regions and at grain boundaries are visible at higher magnifications (figure 10 B). The segmented structure of the precipitations is shown in the electron micrograph using the BSE contrast in figure 10 C. A detailed analysis of the precipitations has been performed using EDS mapping and the results are shown in the bottom line of figure 10. Fractions of the precipitations that appear in darker grey values in BSE image are rich in molybdenum, lighter segments are rich in chromium. The entire precipitations are characterized by a lower cobalt concentration. Based on the determined characteristic composition, it is assumed that the chromium and carbon rich phase represents carbides. Grains within the base material exhibit large grain sizes due to the dendritic structure. In surface near areas, grain size is decreased considerably from few millimetres in the centre to about 150 µm at the surface (figure 11 A), respectively.
Figure 11. Microstructure of the CoCr28Mo6 cast alloy stem. (A) Surface area exhibiting finer grain size. (B) Twin boundaries within surface near grains. (C) Corundum particles embedded in the surface of the stem and the corresponding EDS spectrum.

Depth-related reduction of the grain size, as shown in figure 11 A, might be a result of the faster cooling and solidification with more nuclei at the crucible interface. The grain size reduction along the component contour was observed in all cross sections. There are no further changes in the dendritic microstructure in the longitudinal direction (see Appendix figure A 2).

Twin boundaries are a common feature in CoCr28Mo6 cast alloys due to a low stacking fault energy [42]. Figure 11 B illustrates the present twin boundaries in the stem investigated within this study. Similar to the TiAl6V4 stem, the surface of the CoCr28Mo6 stem exhibits irregularities as shown in figure 11 C. An Al2O3 (corundum) particle presumably originating from a previous blasting or grinding in the manufacturing process is embedded in the deformed surface. Clarification of the present crystal structure was provided using EBSD. Figure 12 A-C illustrate the central area of a cross section, figure 12 D-F show a surface near area.

Figure 12. EBSD results of the CoCr28Mo6 stem from central (A-C) and surface area (D-F): (A) Image Quality map (IQ). (B) Inverse Pole Figure (IPF map) indicating crystallographic orientations. (C) Phase map assigning crystal structure. (D) SEM image of the surface with a red frame highlighting the area for EBSD in (E) and (F). (E) IQ map showing deformation marks (thin lines). (F) Phase map revealing hcp phase being present along the determined lines and surrounded by the fcc matrix.
Grains within the material do not show a preferred orientation as proven by IPF map in figure 12 B. EBSD measurements revealed that the face centred cubic (fcc) structure is present in the investigated stem, see figure 12 C. This correlates with common studies that characterized CoCr28Mo6 cast alloys [43–45]. Small amounts of hexagonal close packed (hcp) phase can only be located in surface near areas in the vicinity of deformation marks (thin lines) indicating a strain-induced phase transformation during manufacturing process of the stem. Figure 12 D-F are visualizing these features. The thin lines exist in hcp structure (red) while the matrix is present in fcc structure (green). Plastic deformation of cobalt base alloys leads to work hardening due to twin formation or strain-induced phase transformation from fcc to hcp (known as ε-martensite) [42]. The volume fraction of deformation-induced formation hcp phase is limited to 0.4 to 0.5 and causes an increase of hardness, yield and tensile strength as well as improved wear properties [46, 47].

The corresponding microhardness depth profile of a cross section of the CoCr28Mo6 alloy stem is illustrated in figure 13.

![Figure 13](image)

**Figure 13.** Microhardness depth profile of the CoCr28Mo6 stem. Distance delineates the perpendicular offset to the surface of the sample.

Besides a 10% increase of hardness in the outer surface layer (distance of 0.5 mm) there is a uniform profile of the hardness with increasing distance to the surface. The mean hardness of the measurement illustrated in figure 13 is 402 ± 12.3 HV0.05. The increased hardness (442 HV0.05) at the surface can be attributed to the observed lower grain size according to the Hall-Petch equation [48]. In addition, hcp phase was detected that is associated with a strain-induced phase transformation due to stem processing and results in an increased hardness.

4 Conclusion

The aim of this study was the examination of the microstructure and basic properties of a TiAl6V4 stem for cementless application and a CoCr28Mo6 stem for cemented application. The TiAl6V4 wrought alloy exhibits a homogenous two-phase microstructure and constant hardness in comparison to a characteristic cast microstructure with interdendritic precipitations and an increased surface hardness possessed by the CoCr28Mo6 cast alloy. Residues from the manufacturing process were detected in the surface and sub-surface on both new prostheses. Corundum blasting was applied to roughen the surface for a better CaP adhesion and as surface finishing after investment casting thereby causing areal grit inclusions of Al2O3 particles. Individual observations suggest that there could be a possible negative impact of residual particles on femoral stems on implant longevity. Böhler et al. observed adverse local tissue reactions in periprosthetic tissue with presence of corundum particles in early retrieved titanium based hip implants [28]. Silicon and aluminium particles were also detected on explanted hip stems of titanium and cobalt based alloys examined by Ricci et al. [27]. If corundum particles are not cleaned from the surfaces after the blasting process and liberated from the surfaces, these hard particles might migrate and could act abrasive [26, 29, 30, 40]. It was shown that adjoining cracks and crevices around embedded particles deteriorate the metal corrosion resistance and promote bacterial adhesion [27, 49].
Despite these individual investigations, the long-term results of corundum blasted stems with ultrasonic post-cleaning (e.g. SPII stem prostheses) verify good clinical results for more than 30 years [50].

A stable femoral stem fixation is particularly relevant since aseptic loosening of the stem component represents one of the main reasons for revision surgery [11, 51]. Micromotion caused by cyclic implant loading are inevitable due to differences in stiffness between stem metal, cement, cancellous and cortical bone. Therefore, surface and material properties should be adjusted. In order to minimize the risk of critical crevices around embedded particles, it is suggested to enhance the finishing process of cemented and cementless stems with regard to the avoidance of residual particles by ensuring good adhesion of the CaP coating. In particular, blasting with metallic particles instead of abrasive grit followed by pickling is a suitable method to generate a contaminant free roughened surface. In addition, the metal’s potential of a targeted adjustment of the subsurface microstructure that aims at an increased wear resistance should be exploited. In this context defined work hardening (e.g. deep rolling) or thermomechanical processing (e.g. solution annealing, quenching) should be envisaged. Research in the area of postprocessing after blasting and customized surface treatments in order to develop a contamination free, wear resistant implant surface offers great potential for improvement of resulting properties.

Acknowledgements
The authors Maria Crackau and Karsten Harnisch contributed equally to this work.

The authors would like to thank Michael Reppin and Oliver Michael for their excellent technical assistance and valuable support. This work was in part conducted within the context of the International Graduate School MEMoRiAL at Otto von Guericke University (OVGU) Magdeburg, Germany, kindly supported by the European Structural and Investment Funds (ESF) under the program “Sachsen-Anhalt WISSENSCHAFT Internationalisierung” (project grant no. ZS/2016/08/80646).

Finally, the authors would like to thank the German Research Foundation (DFG) for the funding of the used DualBeam FEI Scios within the framework of the funding of large-scale facilities according to GG§91.

Appendix

Figure A 1. Comparison of the microstructure of samples from the central (B) and edge positions (C) of various sections from the TiAl6V4 stem (location see figure 1 and figure 2).
Figure A 2. Comparison of the microstructure of samples from central (B) and edge position (C) of various sections from the CoCr28Mo6 stem (location see figure 1 and figure 2).

References
[1] Pilz V, Hanstein T and Skripitz R 2018 Acta Orthop.
[2] Statistisches Bundesamt 2019 Die 20 häufigsten Operationen insgesamt (OPS 5): Vollstationär behandelte Patientinnen und Patienten in Krankenhäuser 2018 https://www.destatis.de/DE/Themen/Gesellschaft-Umwelt/Gesundheit/Krankenhaeuser/Tabellen/drg-operationen-insgesamt.html
[3] Kaivosoja E, Tiainen V M, Takakubo Y, Rajchel B, Sobiecki J, Konttinen Y T and Takagi M 2012 Woodhead Publ. Ser. Biomater. 178–218
[4] Davis J R (ed) 2006 Handbook of materials for medical devices 4th ed (Materials Park, Ohio: ASM International)
[5] Karuppal R 2016 J. Orthop. 13 190–2
[6] Bleß H-H and Kip M 2018 White Paper on Joint Replacement (Berlin, Heidelberg: Springer)
[7] Konan S, Abdel M P and Haddad F S 2019 Bone Joint Res. 8 604–7
[8] Abdel M P, Houdek M T, Watts C D, Lewallen D G and Berry D J 2016 Bone Joint J. 98-B 468–74
[9] Moskal J T, Capps S G and Scanelli J A 2016 Arthroplast. Today 2 211–8
[10] Abdulkarim A, Ellanti P, Motterlini N, Fahey T and O’Byrne J M 2013 Orthop. Rev. 5 e8
[11] Alexander Grimberg, Volkmar Jansson, Oliver Melsheimer, Arnd Steinbrück 2019 0–66
[12] Albretksson T, Bränemark P I, Hansson H A and Lindström J 1981 Acta Orthop. Scand. 52 155–70
[13] Wong M, Eulenberger J, Schenk R and Hunziker E 1995 J. Biomed. Mater. Res. 29 1567–75
[14] Jones C P and Kelley S S 2001 Curr. Opin. Orthop. 12 52–6
[15] Herrera A, Mateo J, Gil-Albarova J, Lobo-Escolar A, Ibarz E, Gabarre S, Más Y and Gracia L 2015 Biomed. Res. Int. 2015
[16] Kienapfel H, Sprey C, Wilke A and Griss P 1999 J. Arthroplasty 14 355–68
[17] Claes L, Kirschner P, Perka C and Rudert M 2012 AE-Manual der Endoprothetik (Berlin, Heidelberg: Springer)
