Praxis Technology: Enhancements to Ti-MIM processing bring medical implants a step closer

Praxis Technology, a US-based manufacturer focused exclusively on titanium products has made substantial progress in the area of titanium Metal Injection Moulding (Ti-MIM). In this article Joe Grohowski and Jobe Piemme provide an insight into the capability and performance of Praxis' validated Ti-MIM process in relation to ASTM F2885, as well as additional technologies that have been developed to enhance Ti-MIM's applicability to the orthopaedic market and other markets demanding high fatigue performance.

The Metal Injection Moulding (MIM) of titanium for medical applications has been referred to as the "holy grail" of MIM. The challenges of forming a highly reactive, very fine metal powder are well understood within the industry. Titanium has the dual challenge of being both reactive and very sensitive to contamination. Over the years there has been tremendous activity in academia and some activity in industry with regards to overcoming the challenges of Ti-MIM [1]

The most relevant of those challenges is meeting the chemical and mechanical requirements of the Grade 5 (Ti-6AL-4V) alloy. In 2011, ASTM adopted a standard for the Metal Injection Moulding of titanium (ASTM F2885). This standard contemplates two versions of the product. Type 1 is a densified version having higher mechanical properties meeting conventional Ti-6Al-4V ASTM requirements. Type 2 is an as-sintered version having lower mechanical requirements. The interstitial requirements for both are the same. Praxis' development efforts have focused on Type 1 because the specification for Type 1 is most similar to other ASTM specifications used for titanium implantable devices, presenting less adoption challenges to the device OEMs.

During validation, the consistency of the Ti-MIM process was evaluated at many points along the process. This article focuses on testing the outputs of the process from both a perspective of interstitial content and mechanical properties. Both of these characteristics cannot be non-destructively inspected and must be monitored by a statistical sampling plan to ensure quality during production. In order to develop a sampling plan that meets the quality requirements of the customer, it is necessary to determine the capability of the process.

	Oxygen	Carbon
Average	0.174	0.0375
Standard deviation	0.007	0.005
Difference	0.0052	0.0085
Based on 95% confidence and a power value.95		
Calculated sample size	25	5

Table 1 Data set for basis of sample size determination

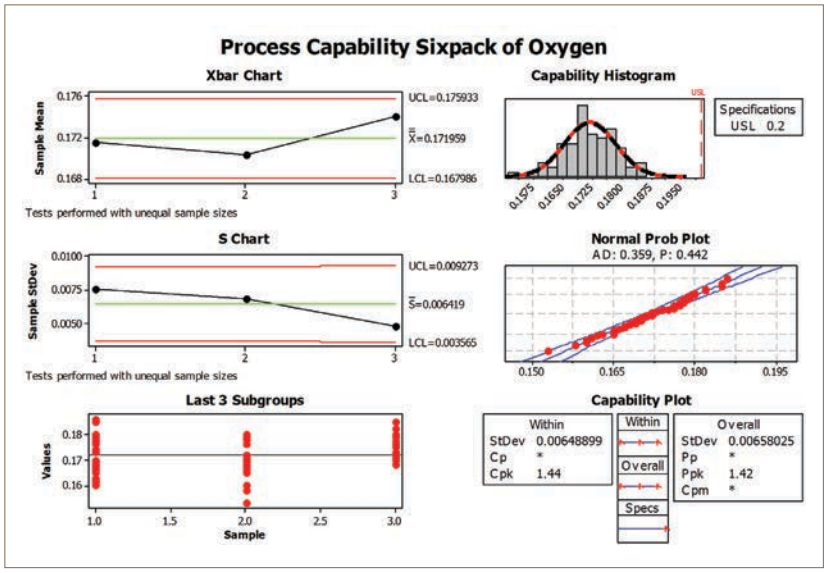

Fig. 1 Results of a Minitab analysis for oxygen capability

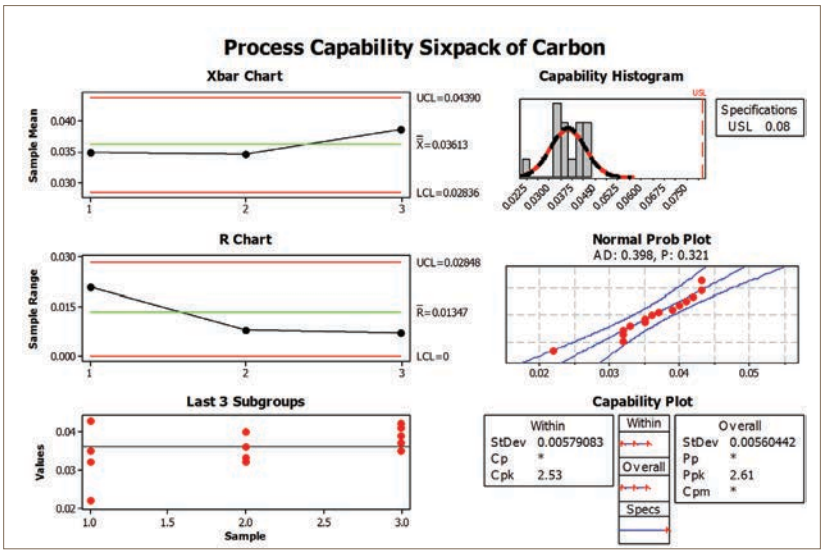

Fig. 2 Results of a Minitab analysis for carbon capability

Controlling oxygen and carbon

With respect to interstitial contamination, oxygen and carbon are widely understood to be the most challenging to control. To evaluate our ability to control these elements a sample size for the capability studies must first be established. Sample size calculations were based on historical data that was collected during an engineering study performed prior to the validation; the data was used to calculate sample sizes based on 1-sample z-tests. A summary of the data used to determine sample sizes is presented in Table 1.

The calculation of average and standard deviation is straightforward. Next the difference value had to be calculated. Because a batch operation is being evaluated, the difference calculation was based on the desire to detect a shift in the mean value. The mean shift difference was selected to be 20% of the span between the average and the upper specification limit. This yields roughly a 50 ppm mean shift for oxygen and an 85 ppm mean shift for carbon. In order to enhance the accuracy of detecting a true mean shift the power value was increased to 0.95 from the widely used value of 0.80. Based on 95% confidence, the samples sizes for both oxygen and carbon were 25 and 5 respectively.

Consistent with performance qualification (PQ) requirements capability studies for oxygen and carbon were based on three full, consecutive furnace runs. Oxygen content was determined using inert gas fusion according to ASTM E1409-08 and carbon content was determined using combustion according to ASTM E1941-10.

Prior to the capability analyses, normality and equal variance tests were conducted on the three data sets. All three were normally distributed with equal variance. Fig. 1 shows the results of Minitab analysis for oxygen capability. Analysis of the oxygen data indicated that the distribution was normal and that the process had a Ppk of 1.42, exceeding the objective of 1.33.

Samples for carbon capability were randomly selected from the sample sets used for oxygen capability. Based on the preliminary results a much smaller sample size of five tests per run was used. Fig. 2 shows the results of the Minitab analysis for carbon capability. The capability analysis provided a Ppk of 2.61, exceeding the PQ objective of 1.33.

Meeting the requirements of the medical industry

Developing a sintering process window and proving capability at the boundary conditions is critical to meet the stringent requirements of the medical industry. When qualifying equipment and processes, not only is it important to quantify the temperature variation within the processing window but it is equally important to understand the effect the variation imparts on final product properties.

Although the determination of the sintering window is based on numerous characteristics, capable mechanical properties are paramount. After target sintering temperature and the temperature window were established, boundary condition tests were conducted to determine mechanical property capability.

The approach used to determine the sample size for mechanical properties was different to that of

interstitial content. In the case of mechanical properties, confidence that the properties met a certain minimum was the objective. In the case of interstitial content, the detection of a mean shift was of the most interest.

Using engineering studies developed to determine the target sintering temperature window, sample sizes were selected based on the baseline Cpk's for tensile strength, yield strength and elongation. The lowest baseline Cpk exceeded 1.41 but was less than 1.55. Using Wayne Taylor's sampling plan tables, a minimum sample size of 20 is needed for variable data, one-sided, applying 95% confidence and 99% reliability assuming high risk [2]. Tensile tests were conducted on a sample size of 25 at both low and high sintering temperatures to determine capability at the boundary conditions of the sintering window.

The acceptance criteria for a minimum sample size of 20 is a Ppk = 1.10. The mechanical properties determined outside of the sintering temperature window proved process capability outside of the temperature variation when processing at the target sintering temperature. Table 2 summarises the properties and statistical results of the process at the upper and lower temperature limits. All of the resulting Cpk values exceed the qualification objective of 1.33.

The challenges of orthopaedic devices

Orthopaedic devices are often cited as applications that could benefit from commercially viable Ti-MIM. However, there are several barriers to Ti-MIM being widely adopted in the orthopaedic industry. Among these are fatigue performance and the increasing demand that orthopaedic implants have complex integration surfaces. In order to overcome these barriers Praxis developed two technologies that complement the Ti-MIM process, expanding its applicability into various orthopaedic markets.

Increasing fatigue strength of Ti-MIM

As mentioned above, a major challenge to using Ti-MIM to manufacture orthopaedic devices is that conventional Ti-MIM components do not have adequate fatigue strength for load bearing applications. The commonly accepted minimum for load bearing applications is around 620 MPa (90 ksi) at 10 million cycles When measured in rotating beam fatigue (ASTM E468-11) typical Ti-MIM fatigue strength is around 480 MPa (70ksi) at 10 million cycles. In order to overcome this limitation, Praxis developed a processing route to improve the final microstructure of the sintered titanium. This process, branded "TiRx™" provides fatigue strengths in excess of 620 MPa. TiRx achieves this performance while still meeting the chemical and mechanical requirement of ASTM F2825. Fig. 3 compares the rotating beam fatigue life of conventional Ti-MIM material and Praxis TiRx material.

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		Low temperature			High temperature		
Property	**ASTM F2885 limit**	**Avg.**	**Std. Dev.**	**Cpk**	**Avg.**	**Std. Dev.**	**Cpk**
UTS (MPa)	900	982	2.81	9.80	964	3.77	5.70
Yield (MPa)	830	871	9.65	1.42	860	6.68	1.47
Elongation (%)	10	19.9	1.05	3.13	19.8	1.19	2.74

Table 2 Summary of mechanical properties and capability for high and low temperature

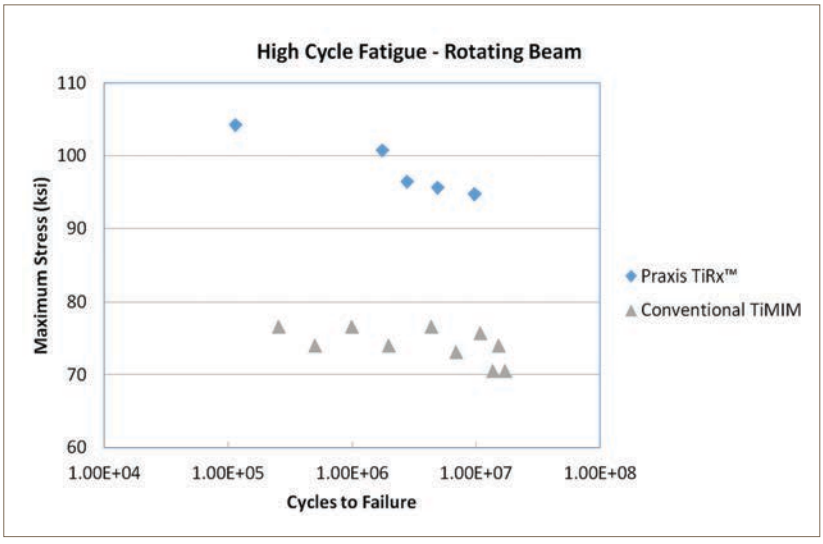

Fig. 3 Comparison of high cycle fatigue performance for TiRx and conventional Ti-MIM

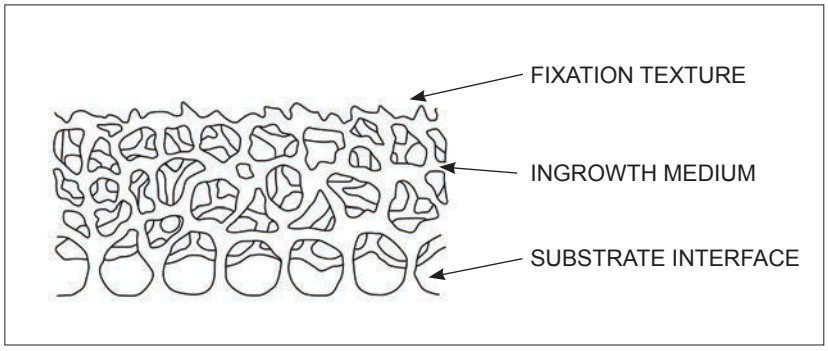

Fig. 4 Schematic of an idealised ingrowth surface

Improved osseointegration using Additively Manufactured sacrificial inserts

A further challenge to Ti-MIM in orthopaedics is that many orthopaedic devices are required to have integration surfaces to provide improved osseointegration. These surfaces are typically added to the device after it is initially formed by conventional methods, most typically forging followed by machining. This additional step represents additional manufacturing costs. The cost of adding a surface to an implant is often a substantive portion of the cost of manufacturing the implant and often also incorporates an additional thermal cycle that compromises the microstructure and diminishes the fatigue performance of the device.

The possibility of net-shaping the integration surface with the implant body is an opportunity for Ti-MIM to add value to the end product. This offers both additional savings to the customer and makes Ti-MIM more competitive. In addition, this surface is formed in the same thermal cycle allowing the possibility of complex ingrowth surfaces on implant bodies having superior fatigue performance. For these compelling reasons, Praxis developed its 3DT process for the net-shape forming of a variety of complex surfaces on injection moulded titanium articles.

The patent pending 3DT process uses an Additively Manufactured sacrificial insert to form the desired surface on a Ti-MIM part. The sacrificial insert can be removed after moulding by a variety of steps, preferably during first stage debinding. The technique can be adapted to many different applications requiring complex surfaces, among them uncemented, cemented and monoblock or unitised devices. Because the surface is formed by an insert and is removed with the moulded article, these surfaces can be applied to areas of a part that cannot be addressed by in-mould texturing. In addition to the insert being created by Additive Manufacturing, the surface can be very complex, easily possessing attributes such as highly interconnected porosity or multiple undercuts.

The approach of using a sacrificial insert enables the creation of very complex surfaces and allows the surfaces to be applied to areas of a part such as undercuts and inside diameters that would otherwise be very challenging to access using conventional line-of-sight processes.

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Complex surfaces for specific applications

Different types of orthopaedic devices require surfaces of different natures. A cemented device is often required to have a rough but not porous surface, often referred to as a 2D or ongrowth surface. An uncemented device not only requires a certain roughness to the surface, but also needs a layer of interconnected porosity that communicates with the outer surface. Surfaces intended for uncemented devices are often referred to as 3D or ingrowth surfaces. In addition to surfaces intended for contact with bone, some devices need surfaces intended to contact and bond with a polymer, most typically UHMWPE. Unitised or monoblock devices incorporate permanently bonded polymers as their bearing surface and require a surface that can securely bond to the polymer.

Ingrowth surfaces

Ingrowth surfaces currently demonstrate the fullest potential of the 3DT technology. A schematic of an idealised ingrowth surface is shown in Fig. 4. These surfaces are very complex and possess separate requirements for each surface region. The outermost portion is engineered to be rough with very open porosity. Rough surfaces are preferred by surgeons to provide initial fixation during implantation. This rough surface provides a "grippiness" to the outer surface and allows the surface to scrape bone and tissue matter into the porosity, which is thought to give osseointegration a head start and improve healing times. It is also important that this surface not be too sharp; while roughness is desired, extremely sharp edges are detrimental to bone growth. This fixation texture is a critical attribute of any ingrowth surface.

Ingrowth medium

Situated underneath the fixation texture is the ingrowth medium. This provides the network of interconnected pores to act as a scaffold for bone to grow onto and then into the ingrowth surface. Among

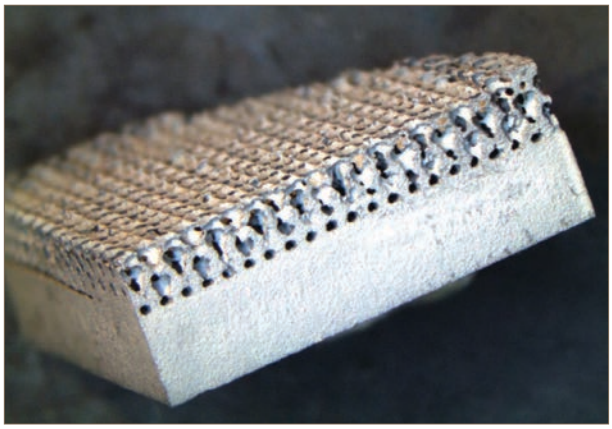

Fig. 5 Three dimensional ingrowth surface manufactured using 3DT technology

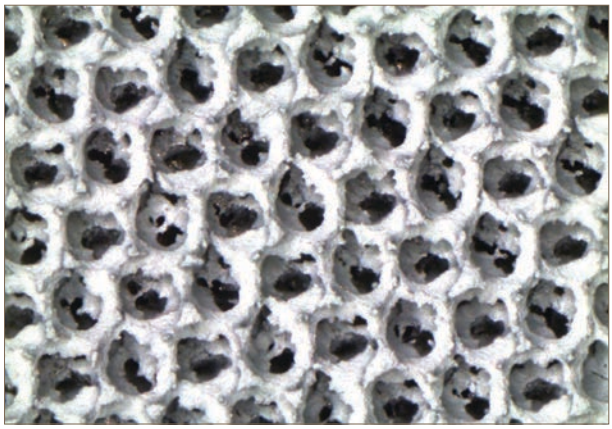

Fig. 7 Close-up of highly interconnected ingrowth medium and fixation surface

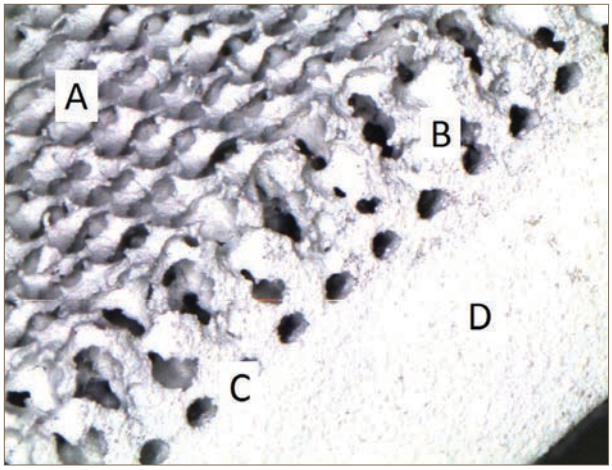

Fig. 6 Cross-section of a titanium integration surface manufactured using the 3DT technology

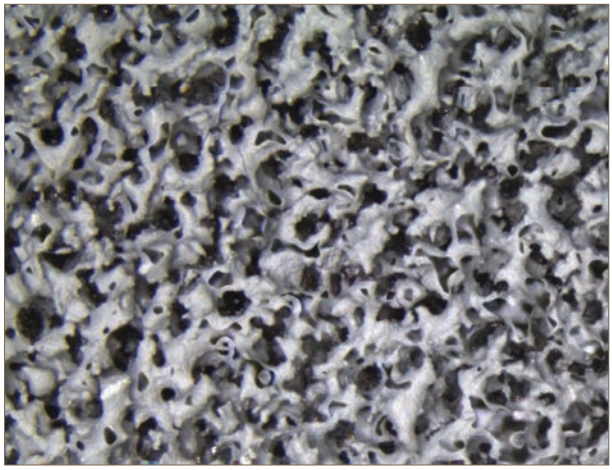

Fig. 8 3DT surface with finer porosity and more organic appearance

the critical attributes of a titanium ingrowth medium are major and minor pore diameter, porosity and tensile strength. The major pore diameter of these regions can range from 50 – 750 microns. The minor pore diameter, or interconnecting pore diameter, is generally preferred to have an average value over 100 microns although ingrowth has been demonstrated below 50 microns. Porosity can range from 55% to 75%. A minimum tensile strength of 20 MPa is set forth in FDA guidance documents. Ingrowth mediums are generally between 0.8 mm (0.03 in.) and 2.3 mm (0.090 in.) in thickness.

Substrate interface

Lastly, underneath the ingrowth medium is the substrate interface. In this region the ingrowth medium attaches to the solid implant body. This region is of importance because titanium is especially notch sensitive, and small stress concentrators on the surface can substantially diminish the fatigue performance of an implant. Most methods of forming an ingrowth surface focus on creating the porosity and the fixation texture. The substrate interface is generally not the result of an intentional design, but rather of the method used to form the porosity of the ingrowth medium.

Shown in Fig. 5 is a cross-section of an integration surface manufactured using a sacrificial insert. The fixation texture, ingrowth medium and the substrate interface are all defined precisely, repeatably and independently by different sections of the sacrificial insert. Furthermore, these surfaces can be created on geometries without line-of-sight access.

Fig. 6 is a close up of the cross-section visible in Fig. 5. Area A indicates the aggressive net-shape fixation texture, area B indicates the highly interconnected ingrowth medium, area C indicates the tailored substrate interface and area D indicates the substrate.

The three dimensional ingrowth surface made using 3DT technology is seen in Fig. 7. The pores are highly interconnected and the struts are well formed. The ingrowth medium is 70% porous and the major pores have an average diameter of 500 microns and the minor pores have and average interconnecting pore size over 100 microns. Tensile bond strength testing performed on these surfaces yielded an average bond strength of 64MPa.

The surface depicted in Fig. 7 is highly regular and repeating with

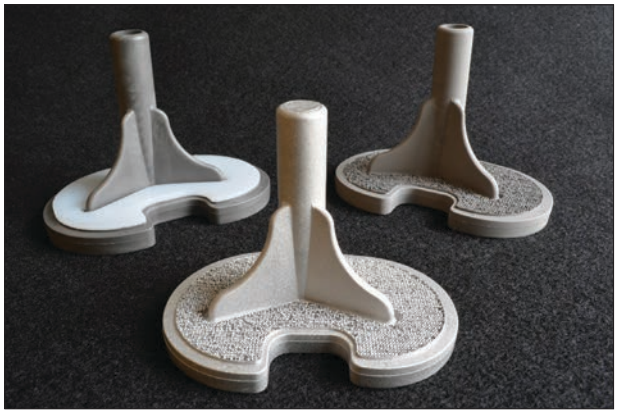

Fig. 9 Ti-MIM tibial tray with net-shaped 3DT ingrowth surface

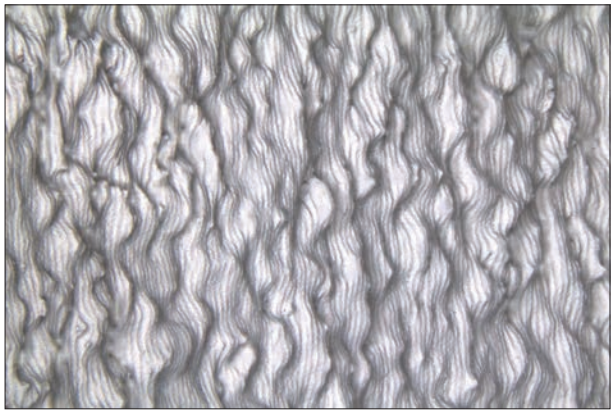

Fig. 10 Ongrowth surface created using 3DT technology

respect to pore size and location. From a marketing perspective it is often desired that three dimensional ingrowth surfaces have a less regular and more organic appearance. This can be accommodated by adjusting the program used to generate the sacrificial insert. Fig. 8 shows the versatility of the process, this surface has finer porosity and a much more organic appearance.

The 3DT technology has been demonstrated to enable conforming of ingrowth layers on large orthopaedic devices. Fig. 9 shows three stages of the 3DT process. The left background shows a green moulded part that has been co-moulded with a sacrificial insert using an insert moulding operation. The right background shows the green part after the sacrificial insert has been removed, and the foreground show the device after being debound and sintered.

Ongrowth surfaces

The 3DT technology can also be used to form ongrowth surfaces. These are less complex, providing an aggressive initial fixation texture, but without a porous ingrowth medium. A close-up of an ongrowth surface created using this technology is shown in Fig. 10.

Conventionally ongrowth surfaces are applied in a coating process and consequently delamination is always a concern. Because the 3DT surface is co-formed with the implant body, no boundary exists between the surface and the body, precluding delamination as a failure mode. Also, because the process is not a line-of-sight process, these aggressive textures can be applied to undercuts, inner diameter and other difficult to reach areas. Furthermore, the process is highly repeatable and tailorable.

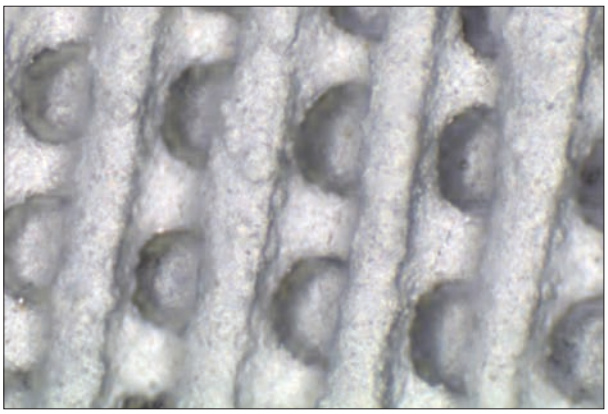

Fig. 11 Polymer anchoring surface with interlocking loops

Fig. 12 Cross-section of moulded UHMWPE bonded with an anchoring layer

Providing a bond between metal and polymer surfaces

Monoblock or unitised devices present another unique challenge that the 3DT technology addresses. These devices need a surface that UHMWPE can be moulded against to create a strong bond between the metal substrate and the polymer surface. Polymer anchoring layers can be used to create these bonds and there are several designs that have been demonstrated to provide excellent bonding between the polymer and the metal substrate. Two examples of these types of designs are interlocking loops and undercuts.

Interlocking loop design

A close-up of an interlocking loop design can be seen in Fig. 11. This approach provides for many loops that polymer can flow around to anchor to the surface. The surface shown is approximately 0.5 mm (0.02 in.) in thickness.

A cross-section of an interlocking loop surface that UHMWPE has been moulded against is visible in Fig. 12. The polymer is shown to have connected through the loops in an uninterrupted manner.

Undercut design

Undercuts can also be used to form strong bonds between a polymer bearing surface and a metal substrate. By moulding the plastic material into undercuts on a surface, the plastic becomes mechanically bonded to the surface. Fig. 13 shows an undercut surface that can be used as a polymer anchoring layer. This surface is approximately 0.35 mm (0.014 in.) in thickness.

Both interlocking loops and undercuts create very strong bonds with moulded UHMWPE. Test coupons tested in tension using a modified ASTM F1147 protocol demonstrated tensile bond strengths in excess of 14 MPa (2000 psi) for either approach. The polymer anchoring layers also have the same advantages of eliminating delamination concerns and great flexibility with regard to where they can be applied.

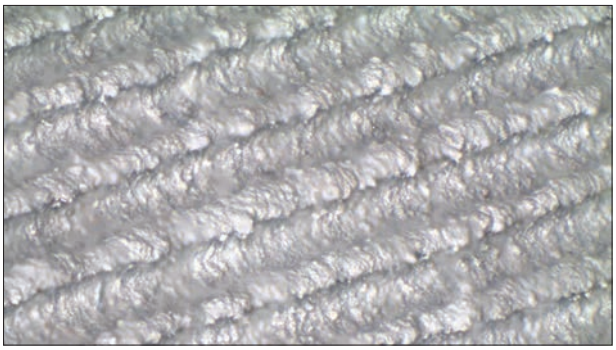

Fig. 13 Undercut surface for use as a polymer anchoring layer

Conclusion

In conclusion, implantable grade Ti-MIM has moved from an academic undertaking to a production capable process. In a validated process, robust capability has been demonstrated with both interstitial contamination and mechanical properties. Further, the technology has been augmented to provide fatigue strengths suitable for load bearing implants and economical, net shape integration surfaces.

Author

Joe Grohowski
President
Praxis Technology
604 Queensbury Ave.
Queensbury, NY 12804, USA
Tel: +1 518 812 0112 x105
Fax: +1 518 812 0115
Email: joegr@praxisti.com

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