Int'l Journal of Research in Chemical, Metallurgical and Civil Engg. (IJRCMCE) Vol. 1, Issue 1(2014) ISSN 2349-1442 EISSN 2349-1450

Metal Injection Moulding as a Possible Processing Route for Porous Prostheses

Shepherd Bhero

***Abstract*—**Metal injection moulding (MIM) can be used to make close to full density parts. The superiority of MIM over conventional casting and machining lies in lower cost and shorter processing time. Of paramount importance is the fact that powder processing circumvents the typical casting problems that include macro-segregation and discontinuities. Thus powder processing is a possible route for manufacturing full-density prostheses devoid of casting imperfections. However, the full density metallic part is much higher than that of bone tissue and would exert skewed weight on the skeletal configuration. This paper explores the feasibility of producing high strength porous prostheses in a bid to lower the overall implant density and to a lesser extent, stiffness, so as to match the bone properties as closely as possible. Powder characteristics and sintering parameters are critical in achieving the desired end.

***Keywords*—**Density, metal injection moulding, porous prostheses, sintering parameters

I. Introduction

THIS paper reviews the merits of metal injection moulding (MIM) in the manufacture of metallic parts for bio-medical application. Conventional casting and machining are time-consuming and also very expensive. Furthermore the parts lack the metallurgical soundness required for certain critical applications. Some of the imperfections include long-range segregation, shrinkage cavities and non-metallic inclusions. The merits of MIM processing include *inter alia* ability to produce near-net shapes that are intricate and complex. Such parts are extremely difficult to produce by conventional processes.

MIM can be used for either full density or porous sintered parts by following critical processing steps that include; selection of powder with specific characteristics, thorough mixing of powder (and binder), and use of appropriate sintering parameters. Full-density sintered parts are preferable for high mechanical strength. However, compact metallic prostheses would have much higher density than bone resulting in the loss of symmetry in the gait of the skeleton. It is therefore necessary to consider reducing the density of the metallic prostheses by introducing uniformly distributed porosity without compromising on mechanical strength. The size, shape and distribution of pores determine overall strength of sintered part.

Shepherd Bhero is with the University of Johannesburg, South Africa (corresponding author's phone: +27 79 604 8080 ; e-mail: bshepherd@uj.ac.za).

II. Literature Review

A. Powder Characteristics

Unlike conventional powder pressing which generally accommodates coarse powders, the mean/median particle size of powders for metal injection moulding is below 40μm [1]-[2]. Fine powders are generally more expensive, but the cost is justifiable if the parts being produced are value-added. Several researches have suggested use of powders of less than 20μm [3]-[4]-[5]-[6]. However, too fine powders require increased binder addition due to higher specific surface area so as to allow smooth flow of mixture during moulding.

The powder particles should be spherical as shown in Fig.1. Such powder geometry is achieved by gas "atomisation" rather than water "atomisation".

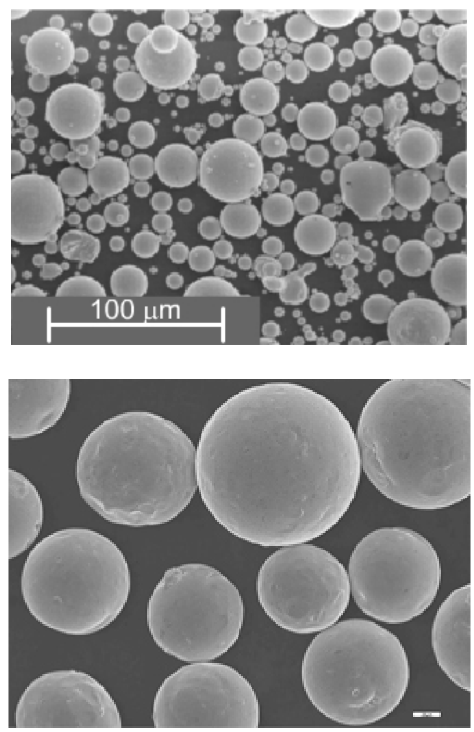

Fig.1 SEM micrographs of gas atomised powder particles

Where high final density is required, wide particle size distribution and fine powders that pack densely are recommended. Powders for porous parts would not pack densely but the pores have to be uniformly distributed.

Angular or irregular particles tend to interfere with particle mobility during compaction since particles tend to interlock and cannot easily glide over each other. The poor packing also results in coarse irregular pores that will not fully heal during sintering.

B. Mixing of powder with binder

Mixing of binder and powder is critical and determines the final density and quality of the sintered parts. Binder must be uniformly spread to from a thin layer around each powder particle. Too much binder will reduce metal powder loading resulting in low sintered densities. On the other hand too little binder may not adequately "plasticise" the powder, thus affecting the rheological properties of the mixture. Thorough mixing is necessary to avoid uneven distribution or segregation of the binder. Binder segregation causes irregular flow patterns of mix, distortion during sintering, non-uniform shrinkage, formation of gross cavities and density variation in the sinters [7]. Use of a high shear rate mechanised device similar to the Brabender torque rheometer plastograph mixer uniform disperses the binder uniformly in the powder [8].

C. Sintering

Clean oxide-free powders have better sintering behaviour. Optimum sintering parameters i.e. temperature and time have to be determined in order to attain high densification during sintering. Excessive temperatures and prolonged sintering times are a needless waste. Prolonged sintering leads to degradation of metallurgical properties due to coarsening of microstructure [9].

D. Difference in properties of bone and metal

Bone mater is different from metals in density and stiffness (Young's modulus). The differences are likely to cause imbalances in the skeletal frame. Table I shows the differences in density between bone tissues materials used to make prostheses. There is no perfect match in densities of metal and bone matter. While Ti-6Al-4V has the lowest density of 4.42g/cm^3, it is still denser than the compact bone 1.90g/cm^3. Stainless steel and Co-Cr-Mo alloy are considerably denser (i.e. 8.0 and 8.3 g/cm^3).

TABLE I
DENSITIES OF BONE VERSUS BIOMETALS

Material	Density (g/cm^3)	Source
Ti-6Al-4V	4.42	[10]
Nitrinol	6.45	[11]
Co-Cr-Mo	8.30	[12]
Stainless steel	8.00	[13]
Compact bone	1.90	[14]
Cancellous bone	0.01 -3.00	[15]

III. EXPERIMENTAL WORK

The metal injection moulding process was investigated using metal powders of different properties. A two-binder system developed at the University of Sheffield was used. The binder consisted of 65% poly-ethyl glycol (PEG) and 15% polyethylene methacrylate (PMMA). The binder was mixed with each metal powder up to 70% metal loading. Compacts were moulded and PEG was leached in water at 60^{o}C leaving the insoluble component PMMA as backbone holding the compacts together the compacts. The compacts were then dried at 40^{o}C prior to sintering

Differential thermal analysis (DTA) was used to determine phase changes taking place in the metals as temperature was raised. The points of inflexion on the DTA traces were taken as indicators of sintering temperatures.

Brown compacts were placed in the sintering furnace and temperature was ramped at 3^{o}C/ minute to 350^{o}C where PMMA was thermally broken down for a period of 30 minutes. The temperature was then raised at 5^{o}C/minute to a predetermined sintering temperature and held to sinter for a duration giving maximum density for each alloy.

IV. RESULTS

A. Characteristics of powders used

The Co-Cr-Mo powder supplied for investigation can be deduced in Fig. 1.

Fig. 2 Coarse Co-Cr-Mo powder with inferior surface finish

The Co-Cr-Mo powder particle size up to 106µm was too coarse and way out of range for particle size recommended for MIM [3]. Some of the particles lacked in sphericity and would cause poor powder packing.

TABLE II
CHEMICAL COMPOSITION OF CO-CR-MO POWDER

Co	Cr	Mo	Ni	Fe	C	Si	Mn	O
63.5	28.5	6	0.18	0.20	0.20	0.47	0.49	0.03

Table II gives the chemical composition of the Co-Cr-Mo alloy, which is a trademark bio-medical material for surgical implants commonly known as Vitallium. Titanium based prosthesis have been found to be more tissue-friendly than Co-Cr-Mo and less corroded by physiological fluids. The chemical compositions of martensitic stainless steel powders used in the investigation are given in Table III.

TABLE II
CHEMICAL COMPOSITION OF STAINLESS STEEL POWDERS

Type	C	Mn	Si	Cr	Mo	Ni	V
440C	1.08	1.0	1.0	16.4	0.43	-	-
440K	1.24	0.74	0.75	18.3	1.25	0.13	0.12

The median particle sizes for type 440C and type 440K powder were 10.2μm and 5.5μm respectively. Both alloys had low sulphur and phosphorus contents.

B. Distribution of binder in the metal powder

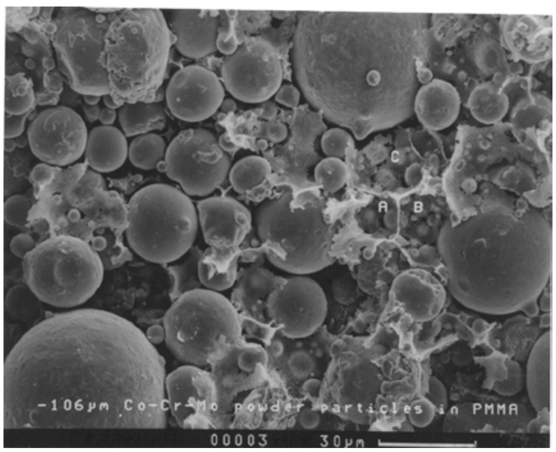

Fig. 3 Residual PMMA after leaching PEG

Uniform distribution of binder is necessary. Any segregation of binder would result in gross pores after sintering. Fig 3 shows the distribution of binder around particles that had been removed from points A, B and C. A concentration of PMMA (white flakes) in some areas indicates non-uniform distribution of binder as a result poor mixing.

C. Sintering response

All alloys showed an initial increase in density to a maximum. With prolonged sintering, the density began to decrease. Figure 4 shows the variation of density of 440K compacts with sintering time. Figure 5 shows the variation of density of 440C compacts with sintering temperature. Maximum densities were obtained at optimum temperature and time. Type 440K stainless steel showed the best sintering response resulting in final density of 99.2% of full density. The maximum densities for type 44C stainless steel and Co-Cr-Mo alloy were 98.3% and 97.4% of full densities respectively. It is important to note that the density for sintered Co-Cr-Mo powder (97.4%) was quite high considering the coarseness and poor surface finish of powder. More satisfactory results would be expected for better powder properties.

At sub-optimal parameters such as too low or too high sintering temperature (Fig. 5) and too short or too long sintering time (Fig. 4), the sintered densities were lower than those obtained at optimum parameters.

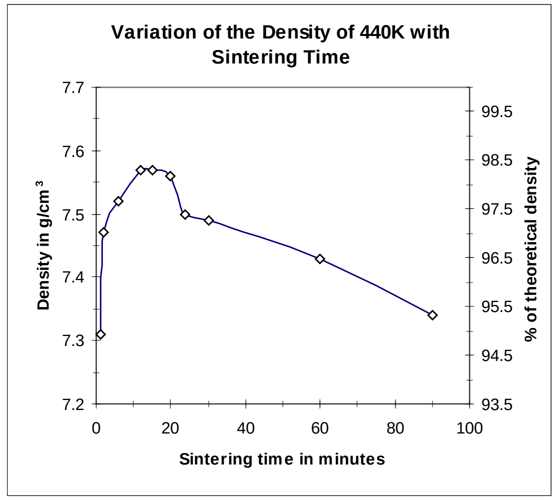

Fig. 4 Maximum density at alloy sintering temperature

Therefore to achieve high density, the specific sintering time for an alloy needs to be determined by a series of sintering trials using preliminary DTA temperatures.

Thus there is no benefit derived from sintering outside optimum parameters if maximum density is the primary objective.

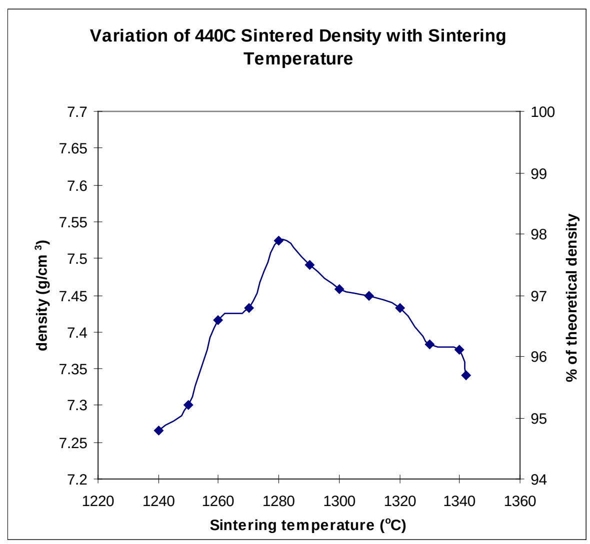

Fig. 5 Maximum density at sintering temperature 1245°C

Sub-optimal sintering parameters can be employed when porous compacts are required. However, the shape of pores depends on the sintering temperature and time chosen. Fig. 6 shows a schematic illustration of formation of porosity.

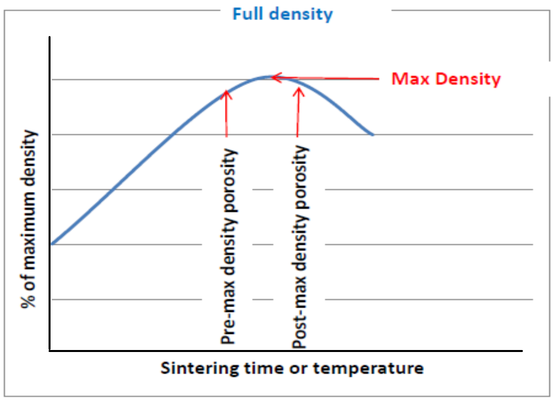

Fig. 6 shows a schematic illustration of formation of porosity.

The porosity occurring prior to full density is irregular and non-spherical because of incomplete sintering. Sintered parts with such porosity have impact strength and would not be satisfactory in the case of prostheses. Parts with spherical porosity are less prone to failure from shock loading and such porosity can be tailor-made by prolonged sintering, where pre-existing pores progressively assume a spherical shape. Pores reduce the overall density of the component. Further decrease in density will occur during extended sintering as a result of coarsening of pores. A careful configuration of porosity in the prosthetic devices will reduce the overall density of the component closer to the density of the bone. Metal injection moulding is a possible process that can be applied to introduce uniform porosity into the prosthesis without significantly affecting the mechanical properties.

V. Conclusion

1) Appropriate particle size and particle size distribution of gas atomised powders can be selected for full density or porous components.. Fine powders and wide particles size distribution being favourable for full densification.
2) Narrow particle size distribution and even mono-size being more ideal for wide and uniform distribution of pores in sintered products. While uniformly dispersed porosity can be employed to reduce density of components, gross cavities caused by binder segregation and poor sintering are detrimental to the integrity of sinters.
3) Mono-size particles pack in a manner that introduces dispersed porosity. Prolonged sinter changes the pore shape to a spherical morphology.
4) Pre-maximum sintering parameters yield angular pores that will reduce the impact strength of the sintered parts.
5) Porosity will decrease the density of metal compact to make the prostheses more structurally compatible with the framework of bones.

Acknowledgment

Special thanks to the University of Johannesburg for sponsorship to attend the conference.

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Shepherd W. Bhero (Pro Eng. ECSA) This author holds a BSc honours degree in metallurgical engineering from the University of Zimbabwe in Harare in 1991, Master of Metallurgy degree from Sheffield, UK in 1994, MSc total quality management from Sheffield Hallam University UK in 1998, PhD in metallurgy from Sheffield University UK in 1999 and MCom Business management from the University of Johannesburg in 2012. He has over 30 years industrial experience starting off as a Metallurgical Cadet then; Metallurgical Technician, Quality Control Shift Supervisor, Plant Metallurgist, Research Scientist, Director of Institute and University Lecturer. Currently he is Senior Lecturer in metallurgy at the University of Johannesburg with special research interest in pyro-metallurgy.

Dr. Bhero is a professional engineer registered with the Engineering Council of South Africa.