LASER POWDER BED FUSION PROCESS DEFECTS AND MECHANICAL PROPERTIES OF TI6AL4V ELI MANDIBLE IMPLANTS

J.A Wessels[1*], A. du Plessis [2], J. Els[1], I. Yadroitsava [1] and I. Yadroitsev[1]

ABSTRACT

Ensuring additive manufactured metal based components are free of major defects is crucial to fulfil medical requirements for clinical applications. Random porosity, high surface roughness and deformation during processing are the main current drawbacks in laser powder bed fusion parts. The prediction of defective samples' mechanical properties with numerical simulations is highly important to understand the effect of these defects. A step-by-step systematic approach of determining defects in laser powder bed fusion (LPBF) and their influence on mechanical properties will be used for the current research. This paper presents the first successful steps in this project.

[1] Department of Department of Mechanical and Mechatronics Engineering, Central University of Technology, Free State, Bloemfontein, South Africa
[2] University of Stellenbosch, CT Scanner Facility, Stellenbosch, South Africa

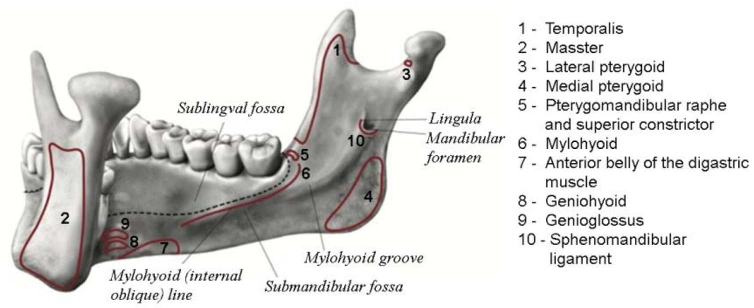
1. INTRODUCTION

The need for new implant, treatments and prosthesis as well as prolonging the life span of current implants has increased, and has brought a robust change in people's quality of life [Khan, 2014]. Additive manufacturing of metals is growing steadily, and offers exciting possibilities for future development - producing parts with higher complexity with many applications including medical and aerospace [du Plessis, 2016a; Dunbar, 2016]. The endoprosthesis replacement of the lower jaw with an artificial implant is typically performed in the case of traumatic gunshot injuries, or when a large section of bone was destroyed by a chronic infection, when malignant neoplasms developed, and also in connection with osteomyelitis lesions. The clinical goal for mandibular implants is to serve as a replacement or anchor for muscle and mastication loading, which recreates the skeleton's original stress-strain trajectories. The mandible is the only movable stress bearing bone in the face and reconstruction of mandibular defects should restore the anatomical height and contour of the resected mandible. Ti6Al4V alloy is one of the most suitable materials of choice for such implants, due to its incredible strength, low weight ratio, outstanding corrosion resistance and biocompatibility.

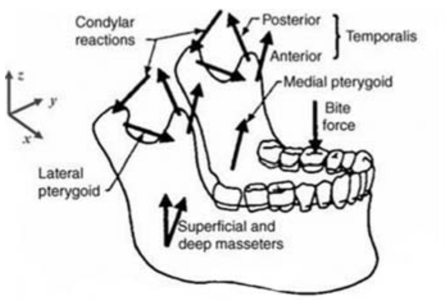
Quality control in metal based additive manufacturing is extremely important for effective control of dimensional inaccuracy, porosity and other defects that arise during the process [du Plessis, 2016b]. Grain morphology and texture of any part produced in laser powder bed fusion (LPBF) are strongly affected by the solidification rate, rapid cooling, and cyclic re-heating and cooling from adjacent tracks and subsequent layers. The inconsistency of the thermal processing throughout the part arises imprecisions from a variety of factors, including scan strategies, processing conditions, and geometric effects such as the reduced ability to conduct heat away from the melt pool when creating overhangs [Dunbar, 2016; Zhao, 2017;Yadroitsev, 2018]. Special microstructure and high residual stress are peculiarities of LPBF material that influence its mechanical properties. Also consecutive layer by layer delivering of powder particles sometimes tends to clump together causing inhomogenous powder layers, which result in unmelted areas and pores in the final product. Inappropriate selection of LPBF process parameters, scanning and manufacturing strategies of complex objects can lead to fractures, warps, and pores. CT scanning has previously been used for defect analysis of LPBF parts and it is clear that imperfections such as pores, etc, are prevalent in this process. [du Plessis, 2018; du Plessis, 2016b].

1.1 Biomechanics of mandible

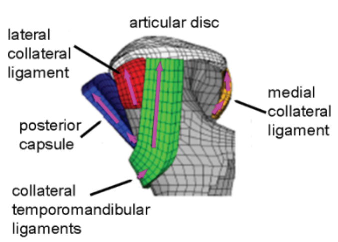
During mastication the following muscles serve in the movement of the mandible: masseter muscle, temporal muscle, medial pterygoid muscle, lateral pterygoid muscle, and buccinator muscle (Fig.1) [Emin, 2018; Kober, 2004; Ingawalé, 2012]. The mandible is the only movable stress bearing bone of the face, and disregarding the forces acting on the mandible can lead to reconstruction failure [Wong, 2011]. Loadings and bone properties are factors that have to be taken into account for modelling of mastication. For numerical simulations, during a mastication cycle the directions of the forces exerted by the jaw closing muscle can be assumed as uniform, due to the fibres running approximately parallel close to the insertion to the mandible; constant through the cycle, according to the small amplitude motion of their insertion points (Fig. 1) [Commisson, 2015].



(a)



(b)



(c)

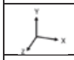
Fig. 1: Mandibular bone with respective muscle attachment (Posterolateral view) (a); typical loading forces, boundary conditions, and force vectors during mastication [Faulkne, r1987] (b) and forces working on the temporomandibular joint during mastication [Commisson, 2015] (c).

For numerical simulations, cortical bone material properties can be considered orthotropic in different anatomic regions of the mandibular bone, cancellous bone and reconstruction plates and screw can be defined as isentropic independent of directions. Loads can be applied to the five principal muscles (Table 1). Material properties of different bones found in the human mandible are shown in Table 2. Thus, typical mechanical properties of different areas of mandible and response of human mandible (strains and loadings) can be found in the literature.

Table 1: Muscular load action [Ramos, 2011].

Muscle action	Load (N)		
	X	Y	Z
Deep masseter	7.776	127.23	22.68
Superficial masseter	12.873	183.5	12.11
Medial pterygoid	140.38	237.8	-77.3
Temporalis	0.064	0.37	-0.13
Medial temporal	0.97	5.68	-7.44

Table 2: Material properties of different bone found in the human mandible [Vajgel, 2013].

Material property	Cortical bone			Cancellous bone
	Symphysis	Body	Angle	
Elastic Modulus, E_x (MPa)	20,492	21,728	24,607	1,500
Elastic Modulus, E_y (MPa)	12,092	12,700	12,971	1,500
Elastic Modulus E_z(MPa)	16,350	17,828	18,357	1,500
Poisson's ratio, P_{xy}	0.43	0.45	0.38	0.3
Poisson's ratio, P_{yz}	0.22	0.2	0.23	0.3
Poisson's ratio, P_{xz}	0.34	0.34	0.28	0.3

1.2 Mandible reconstruction by LPBF

Additive Manufacturing (AM) makes the manufacturing of any given design possible regardless the geometric complexity and allows the production of integrated components. This is extremely important towards the design of revolutionary shapes and lighter parts without the need to consider manufacturing constrains related to machining, moulding, etc. The production in AM does not need any special tooling, thus, making it easy and immediate to manufacture complex objects with various changes in geometry such as customised implants (Fig. 2).

AM mandible reconstructions begins from a CT scan of the patient of the diagnosed area; the CT scan is then converted to a 3D virtual and physical model (Fig. 2a, c and e). The result of this 3D model is used as an input to plan the resection planes and design of a cutting guide that allows surgeons to precisely cut around the tumor/ affected area without fault. A titanium alloy implant is designed to replace the affected area of the mandible (Fig. 2 b, d and f) to fit the required geometry in order to restore facial symmetry and to allow quick recovery.

In some author's expert opinion, a single screw in the anterior segment (Fig. 1a) may not prevent rotation, leading to implant failure (Mommaerts, 2016). Loading and contact boundary conditions are described by three main muscular forces during chewing operation; masseter, medial pterygoid, and temporalis. Each person's physique and bone structure differ from others which means the chewing load can vary in both magnitude and direction. Parameters to evaluate the design is the flexibility of the reconstructive implant, that is, the capability to absorb the chewing load, and the stress and strain distribution, ensuring the maximum stresses developed are lower when compared to the yield strength of Ti6Al4V ELI (Al-Ahmari, A, et al. 2015). Thus, at present implants are designed to withstand forces of 700N with a safety factor of two.

A scaffolding is useful where there is bony contact, the friction provides primary stability as with any screw-fixed plate (Cordey, J, et al. 2000). Scaffolding (Fig. 2b, d, f) increases the overall elasticity to more closely approximate that of bone (Lin, C.Y, et al. 2004) and therefore reduce stress shielding and permanent loosening of an implant fixed to the weight-bearing mandible, and allows an easy method for weight reduction (Mommaerts, M.Y. 2016). In the present cases, implant thickness was 2 mm.

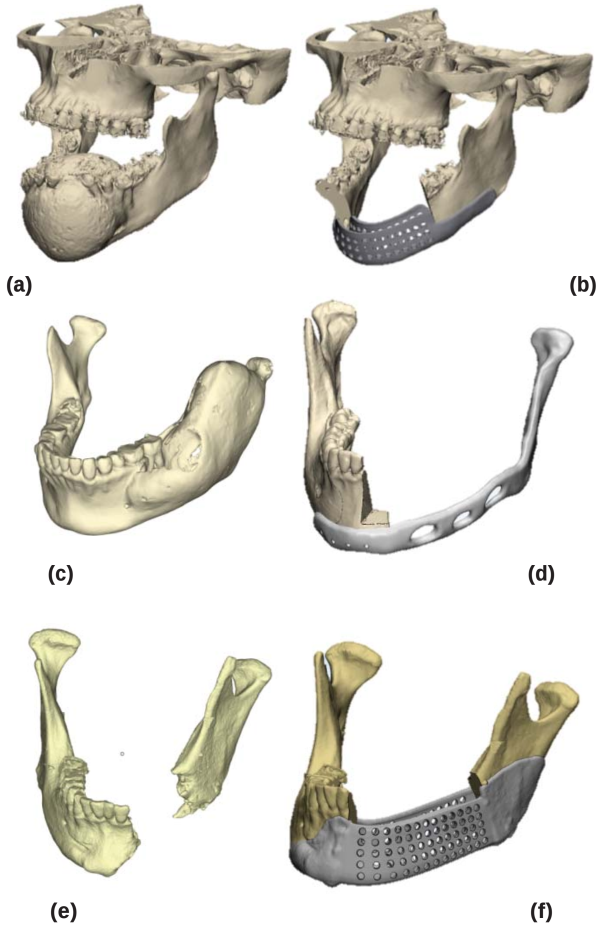


Fig.2: Design steps followed for reconstructing mandibular implants: 3D reconstructed CT scan of mandibular effected area (a,c and e); 3D rendering of designed titanium implant, (b, d and f).

2. METHODOLOGY AND RESULTS

2.1 Methodology

Spatial distribution, size and shape of the defects of AM parts can be done by microCT to predict the effect of the pores on the mechanical properties of the part (du Plessis et al., 2017; Weiler et al., 2005). Vanderesse et al. (2011) and Nicoletto et al. (2010) used microCT in combination with FEA to highlight stress regions and correlate fatigue cracks with pores and their stress regions in aluminium castings. du Plessis et al. (2017) simulated static loads directly on CT scans voxel data. It was shown that combination of spatial data of pore's geometry and their locations is very suitable to predict the effect of the pores on the performance of the part. Failure location prediction by FEA for LPBF mandible implant could serve as a tool for optimizing the design of mandible implants as well as for quality control of produced implants having some porosity or other differences from the CAD model.

Main stages of the present study have to be mentioned:

- investigation of biomechanics and numerical simulations on human mandible functioning;
- analysis of stresses and critical points for different types of mandibular reconstruction;
- analysis of porosity in LPBF parts on the basis of CRPM's experience and literature sources;
- production and testing AM Ti6AL4V ELI samples with prescribed sizes, loadings, directions and porosity;
- numerical simulations and validation data compiled from numerical simulation with mechanical testing of LPBF samples after heat treatment.

Thus, test samples with defects and without defects will be manufactured by LPBF. The test samples will have geometrical characteristics similar to the mandible and mechanical testing will be done according mastication loadings.

2.2 Porosity in LPBF parts

In [Yadroitsev, 2018] LPBF Ti6Al4V ELI tensile samples were analysed by CT scans in as-built condition. It was found that maximum pore size was 132 µm for horizontal sample and 96 µm for vertical one (Fig. 3). The pores were randomly distributed throughout the volume. Although a statistically significant difference was found in the pore sizes for these samples, it can be stated that the porosity of the objects was insignificant: the estimated porosities were 0.0004% for vertical samples and 0.0018% for horizontal ones. The mechanical properties of the horizontal and vertical samples did not differ significantly. It should be mentioned how these samples were produced and analysed: 1) rectangular bars 10×10×60 mm were manufactured in horizontal and vertical directions; 2) round specimens with threaded ends were machined from bars accordingly to the geometry recommended by ASTM E8M standard (gauge length four times the diameter); 3) microCT scans were made for gages 4 mm in diameter and 20 mm in length.

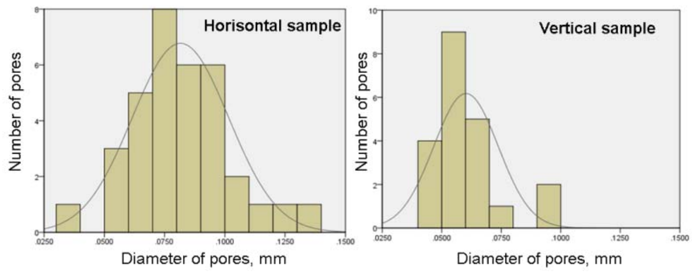
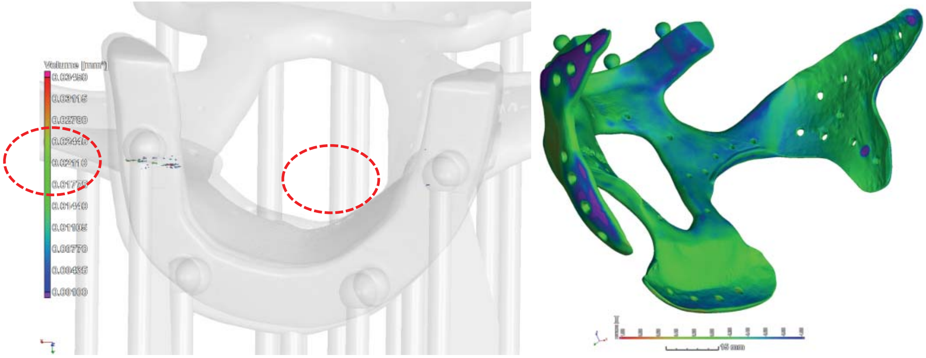


Fig. 3: Diameter of the pores measured by microCT scans in tensile samples.

In the presented study, for Ti6Al4V (ELI) samples density measured by microCT scans, was more than 99.99% (for pores >30 µm). It was found that the biggest pores were predominantly elongated in shape and can be considered as interlayer pores. Defects in LPBF are caused due to melt pool discontinuities and lack of melt pool overlapping. If the laser output is stable with prescribed scanning speed and power, with prescribed optimal scanning strategy, all voids are likely resulting from inhomogeneity in the powder layer. This inhomogeneity can be caused, in turn, by poor powder flowability, violation of loose powder layers during processing and deformation of the part during processing. Thus by LPBF, high density can be achieved at optimal process-parameters and scanning strategy.

Fig. 4 shows the graphic steps followed to produce a complete approved Ti6Al4V metal facial implant of an adult human produced in CRPM, at Central University of Technology with an EOSINT M280. Biomaterials are used to engineer functional restoration of different tissues to improve human health and the quality of life. A key issue in the designing of a new implant of any given product through AM is the prediction of the mechanical properties of the material. Several experimental results show AM-based products are often affected by widespread porosity, low density regions within their volume and anisotropy. These factors are due to manufacturing process, despite efforts of improving the process parameters. AM offers a product development for rapid iteration between designs, assembly and functional tests, (Fig. 4c) bringing about a remarkable decrease in both time and product development costs.

(a) (b) (c)



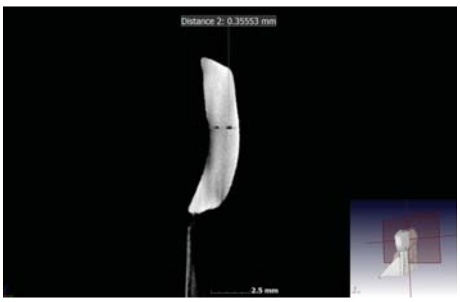
(d) (e)

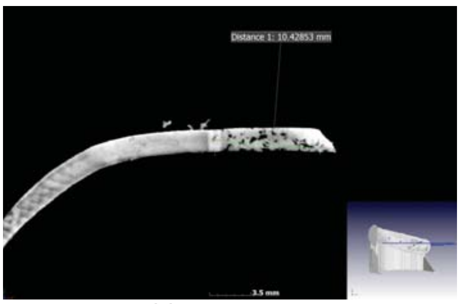
Fig. 4: Facial titanium implant step preview (a) CT scan of human skull defected area (b) implant attachment after the use of a cutting guide, and (c) full 3D assembly, showing placement of implant with lower jaw to ensure no implant error; CT scans reconstruction of the facial Ti6Al4V ELI implant: pores (d) and the deviation of the LPBF part from prescribed CAD sizes (e).

Maxillofacial Ti6Al4V ELI implants produced by LPBF were analysed in terms of porosity by CT scans by General Electric VTomex L 240kV CT as described by du Plessis *et al.* in [du Plessis, 2016a-b]. First scans were done at 100 μm resolution and second time, when main porosity was identified, with 50 μm resolution to provide porosity details in the region of interest. In the first implant, two places were identified with serious defects, which run across the entire width of the sample (Fig. 4d). Under higher resolution, these type of porosity was recognised as layered defects. Analysis of the sample shows that layer pores were arisen due to problems with the powder delivering. When the synthesis of the overhanging parts of the implant began, high residual stress led to the deformation of this part. That, in turn, caused the contact of the recoater with the deformed part and its vibration. The deformation confirmed by the high deviation of this part of the implant from prescribed sizes (Fig. 4e). After vibration, loose powder was compacted and settled, which led to a non-uniform delivering of the next powder layer and, subsequently, to the porosity of the part. For quality control of layered defects du Plessis et al. (2018) proposed to use witness specimens to ensure lack of layered or other unwanted types of defects, as this allows higher resolution to ensure these defects are not missed in a large-part scan [du Plessis, 2018].

In the other sample, the pores also were in a layer and form a porous plane through the samples tip which is up to 17.6 mm in length and the maximum pore's width in that region was 355 μm (Fig. 5). The recommendation was that the pores located in a load bearing section of the implants can influence on mechanical response of the implant and it was decided that this implant could not be suitable for the implant procedure. New implants with additional supports were produced and no internal defects more than 300 μm were identified by CT scans.

(a)





(b) (c)

Fig. 5: Total view of Ti6Al4V mandible implant; CT scans cross-section of mandibular implant: parallel (b) and perpendicular (c) to the build direction.

The new part with supports removed was then scanned and found defect free as mentioned above. This new part was subjected to microCT-based load simulation using the structural mechanics simulation module of VGStudioMax 3.2. This is an immersed-boundary finite element method which allows calculation of displacements and stresses in voxel data using solid mesh elements, but not requiring a conformal mesh as in typical finite element softwares. This is computationally efficient and allows to evaluate the effect of real part geometries including defects, warping, internal porosity, etc. In this simulation, the inputs were linear isotropic material properties for Ti6Al4V: elastic modulus 115 GPa and Poisson's ratio 0.3, with loading direction selected as shown in Figure 6, with load 10 N. The stress distribution shows two major stress areas as shown where the stress reaches 100 MPa. This value is significantly lower than the yield stress of the material and hence, in this case, sufficiently safe for typical loads. However, much work remains in determining a suitable safety factor, validating this workflow, and assessing also the effect of such defects on fatigue life.

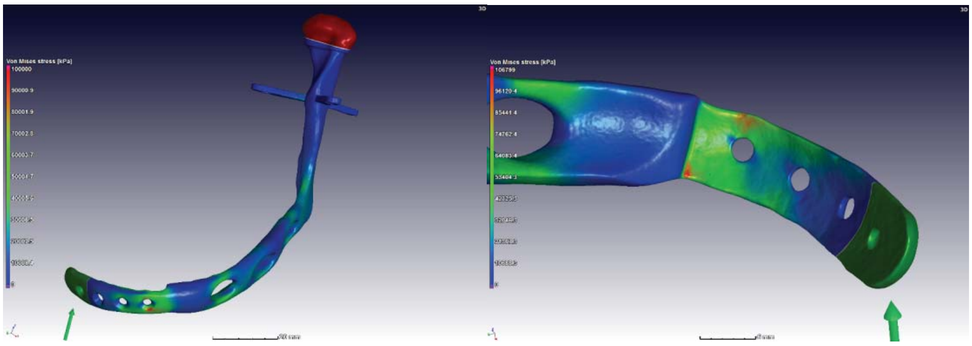


Figure 6: Mechanical simulation highlighting high stress areas in real part, using microCT data.

Quality control for complex shape objects such medical implants are of great interest; non-destructive quality control has to be performed on parts before implantation. ASTM had 21 standards for Ti alloys concerned medical device standards and implant standards to inspect and assess such instruments to ensure proper quality and workmanship [ASTM international]. But additionally, standard similar to "Standard practice for Computed Tomographic (CT) examination of castings", "Standard practice for radiographic examination of advanced aero and turbine materials and components", "New guide for non-destructive testing of Additive Manufactured metal parts used in aerospace applications" and "Standard guide for micro-computed tomography of tissue engineered scaffolds" should be developed for systematic CT scans non-destructive testing of AM Ti implants. It is very important that an implant does not contain critical pores. Understanding of critical porosity in general and maximum critical size of the pores permissible for different type of AM Ti implants are vital task for the new AM technology.

Detailed analysis of manufacturing strategy, build orientation and supports and its influence on porosity can be performed based on X-ray micro-computed tomography. Revealing of the typical shape of the pores and its size permit to produced samples with artificial defects and without it by LPBF. Analysis of porosity and mechanical properties of samples with defects will be compared against standard specimen samples and numerical simulations. Numerical simulation of various loading on test parts with defects and its evaluation by experimental data on loading permit to establish mechanical consequences of defects on manufactured customised mandible implants.

3. CONCLUSION AND FURTHER RESEARCH

A conclusion in the early stages of this work can be done on the basic expected outcomes of the research project:

- When conducting mechanical testing for the purpose of validation of the numerical simulations, all joint and forces during mastication must be tracked and included in both experiments and simulations, to assume ideal conditions for both cases boundary conditions has to be taken into consideration.
- Defects are included in the design of mandibular Ti6Al4V test samples with different design geometries, thus to indicate to which size and direction of porosity will cause any significant defect in the geometry of a human mandible. An example of different test subjects will be subjected to porosity during LPBF with the aid of microCT-based simulations to validate experimental and simulated data.
- We believe that with the aid of mechanical testing in collaboration with numerical simulation and data analysis of different loaded porous implants could open a new chapter in the development of reliable AM medical implants.

ACKNOWLEDGMENTS

This work is based on the research supported by the South African Research Chairs Initiative of the Department of Science and Technology and National Research Foundation of South Africa (Grant №97994) and the Collaborative Program in Additive Manufacturing (Contract №CSIR-NLC-CPAM-15-MOA-CUT-01). The authors would like to acknowledge the contributions made by the CRPM Team from the Central University of Technology, Free State as well as the Team from the CT Scanner Facility at the University of Stellenbosch.

REFERENCES

[1] Al-Ahmari, A., Nasr, E.A., Moiduddin, K., Anwar, S., Al Kindi, M., Kamrani, A., 2015. A comparative study on the customized design of mandibular reconstruction plates using finite element method. Advances in Mechanical Engineering, 7(7), pp. 1-11

[2] ASTM International, 2018. Medical Device Standards and Implant Standards. Available from: https://www.astm.org/Standards/medical-device-and-implant-standards.html.

[3] Commisson, M. S., Martınez-Reina, J., Ojeda, J., Mayo, J., 2015. Finite element analysis of the human mastication cycle. Journal of the mechanical behaviour of biomedical materials, University of Seville, Spain, vol. 41, pp. 23-35.

[4] Cordey, J., Borgeaud, M., Perren, S.M., 2000. Force transfer between the plate and the bone: relative importance of the bending stiffness of the screws friction between plate and bone. Injury, 31, pp. 21-28, 79, 84, 88, 92.

[5] du Plessis, A., le Roux, A.G., Booysen, G., Els, J. 2016b. Quality Control of a Laser Additive Manufactured Medical Implant by X-Ray Tomography, 3D Printing and Additive Manufacturing. 3(3), pp. 175-182.

[6] du Plessis, A., le Roux, S.G., Els, J., Booysen, G. 2016a. Directionality of cavities and porosity formation in powder-bed laser additive manufacturing of metal components investigated using X-ray tomography, 3D Printing and Additive Manufacturing. 3(1), pp. 48–55.

[7] Du Plessis, A., Sperling, P., Beerlink, A., le Roux, S.G., 2018. Standard method for microCT-based additive manufacturing quality control 5: witness specimen analysis. MethodX, in press.

[8] du Plessis, A., Yadroitsava, I., le Roux, S.G., Yadroitsev, I., Fieres, J., Reinhart, Ch., Rossouw, P. 2017. Prediction of mechanical performance of Ti6Al4V cast alloy based on microCT-based load simulation. Journal of Alloys and Compounds, 724, pp. 267-274.

[9] Du Plessis A., Yadroitsava I., Yadroitsev I., le Roux S., X-ray micro computed tomography in additive manufacturing: a review of the current technology and applications. 3D printing and Additive Manufacturing, 2018 *5*(3), pp.227-247.

[10] Dunbar, A. J., 2016. Analysis of the laser powder bed fusion additive manufacturing process through experimental measurement and finite element modelling. PhD thesis, The Pennsylvania State University, USA, pp. 158.

[11] Emin, A., 2017. The Mandible - Structure - Attachments – Fractures. TeachMeAnatomy, Available at: http://teachmeanatomy.info/head/osteology/mandible/, pp. 9 [Accessed March 20, 2018]

[12] Faulkner et al., 1987. A three dimensional investigation of temporomandibular joint loading. Journal of Biomechanics, 20, pp. 997-1002

[13] Ingawalé, S.M. and Goswami, T., 2012. Biomechanics of the Temporomandibular Joint, in: Human Musculoskeletal Biomechanics, Ed. T. Goswami, InTech, Available from: http://www.intechopen.com/books/human-musculoskeletal-biomechanics/biomechanics-of-thetemporomandibular-joint.

[14] Khan, W., Muntimadugu, E., Jaffe, M., Domalb, A. J., 2014. Chapter 2 Implantable Medical Devices, Focal controlled drug delivery, pp. 33-59.

[15] Kober, C., Erdmann, B., Lang, J., Sader, R., Zeilhofer, H. F., 2004. Adaptive finite element simulation of the human mandible using a new physiological model of the masticatory muscles. In: Proc. Appl. Math. Mech. 4, pp.332–333.

[16] Lin, C.Y., Kikuchi, N., Hollister, S.J,. 2004. A novel method for biomaterial scaffold internal architecture design to match bone elastic properties with desired porosity. Journal of Biomechanics, 37, pp. 623-36.

[17] Mommaerts, M.Y., 2016. Guidelines for patient-specific jawline definition with titanium implants in esthetic, deformity, and malformation surgery,Annals of maxillofacial surgery, 6(2), pp. 287-291

[18] Nicoletto, G., Anzelotti, G., & Konečná, R., 2010. X-ray computed tomography vs. metallography for pore sizing and fatigue of cast Al-alloys. Procedia Engineering, 2(1), pp. 547-554.

[19] Ramos, A., Ballu, A., Mesnard, M., Talaia, P., Simões, J. J., 2011. Numerical and Experimental Models of the Mandible, Experimental Mechanics vol. 10, pp. 1007.

[20] Vagjel et al., 2013. Comparative Finite Element Analysis of the Biomechanical Stability of 2.0 Fixation Plates in Atrophic Mandibular Fractures. Journal of Oral and Maxillofacial Surgery. 71 (2), pp. 335-342.

[21] Vanderesse, N., Maire, E., Chabod, A., & Buffière, J. Y., 2011. Microtomographic study and finite element analysis of the porosity harmfulness in a cast aluminium alloy. International Journal of Fatigue, 33(12), pp. 1514-1525.

[22] Weiler, J. P., Wood, J. T., Klassen, R. J., Maire, E., Berkmortel, R., & Wang, G., 2005. Relationship between internal porosity and fracture strength of die-cast magnesium AM60B 333 alloy. Materials Science and Engineering: A, 395(1), pp. 315-322.

[23] Whitaker, L.A., 1991. 1Aesthetic augmentation of the posterior mandible. Plastic and Reconstructive Surgery, 87, pp. 268-75.

[24] Wong, R. C. W., Tideman, H., Merkx, M. A. W., Jansen, J., Goh, S. M., Liao, K., 2011. Review of biomechanical models used in studying the biomechanics of reconstructed mandibles. Int. J. Oral Maxillofacial Surgeons, vol. 40, pp. 393-400.

[25] Yadroitsev, I., Krakhmalev, P., Yadroitsava I., du Plessis, A., 2018. Qualification of Ti6Al4V ELI alloy produced by laser powder bed fusion for biomedical applications. The Journal of The Minerals, Metals & Materials Society (TMS), 70 (3), pp. 372–377.

[26] Zhao, C., Fezzaa, K., Cunningham, R.W., Wen, H., De Carlo, F., Chen, L., Rollett, A.D., Sun, T., 2017. Real-time monitoring of laser powder bed fusion process using high-speed X-ray imaging and diffraction, Scientific Reports, 7, pp. 61-230.