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A review on artificial bone modelling: Materials and manufacturing techniques

Neha Khasnis [a,*], Pankaj Dhatrak [b], Alekh Kurup [a]

[a] Department of Mechanical Engineering, MAEER'S MIT, Pune 38, India
[b] School of Mechanical Engineering, Dr. Vishwanath Karad MIT-WPU, (Formerly MAEER'S MIT), Pune 38, India

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ABSTRACT

Artificial bone modelling has been done in recent times in order to replicate the mechanical properties and microstructure of the human cancellous and cortical bone. The applications of artificial bone models include bone grafting and testing of orthopaedic instruments. The recent developments in this field make the use of natural polymers and bio-compatible metals to manufacture porous structures replicating bone properties. The aim of this review is to consolidate the research efforts done in the field of making artificial bone models of the human trabecular bone by studying the material and manufacturing aspects of the process. The article reviews the bone models under the categories of polymers and composites and metal scaffolds. A combination of natural polymers along with a bio-compatible synthetic polymer and bio-ceramic reinforcement has proven to be a suitable candidate in replicating the mechanical properties of the bone while ensuring bio-compatibility. Three manufacturing techniques were reviewed with respect to the ability to replicate the bone microstructure. SLM is a potential candidate in terms of design flexibility and manufacturing accuracy. A direction to future research in extending the application of artificial bone models to the testing of dental implants has also been provided.

1. Introduction

1.1. Dental implant Stability

A dental implant is a surgical component that is used as a replacement for the tooth. A dental implant is usually made of titanium and its alloys owing to its high strength and bio-compatibility. The threads on the dental implant fuse with the bone to form strong support. The success and utility of a dental implant depend on various factors, majorly on its stability and osseointegration. The stability of an implant depends on various interactions at the interface of the bone and the implant that include bone density, the surface finish of the implant, micromovements, and stress concentration. The stability of an implant is divided into two phases: primary stability, secondary stability. Primary stability involves restricting the micromovements of the implant to 150 μm and depends on implant design parameters and human jawbone quality. Jawbone quality is assessed by the quantity and density of cancellous and cortical bone. Quantitative properties like the strength of the bone provide a more objective way of studying the impact of bone quality on the stability of the implant [1]. Secondary stability involves osseointegration and bone formation [2].

1.2. Implant failure

Dental implants failures occur due to a wide range of factors that include implant cracking, occlusal overloading, infection, improper implantgeometry, and bone quality at the implant site. It was observed that 17% of the implants failed due to occlusal overloading and 1.7% of the implant failures were due to implant fracture which involved a mechanical failure of the implant [3]. The failure timeline is divided into early implant failure and late implant failure where early implant failure is classified as implant failure within the first 12 months after surgery. Late implant failure is caused due to improper implant design which leads to mechanical failure of the implant under overloading. This causes higher bone loss as compared to early implant failure, thus creating problems for reimplantation. Hence it is important to prevent late

* Corresponding author.
E-mail address: nehakhasnis24@gmail.com (N. Khasnis).

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implant failure. In order to prevent late implant failure, it is important to test the design of the implant under occlusal loading for various scenarios in vitro before implantation. Since implant design and its behaviour under loading are important factors under consideration in implant design, in vitro testing of the implants under all these conditions can help modify the geometrical parameters of the implant before the final in vivo application.

1.3. In- vitro testing of dental implant

In order to test the impact of implant architecture on the survival rate on the dental implants, it is necessary to have modifications in implant parameters like thread type, thread angle, implant diameter, and implant length. Finite Element Method (FEM) has proved to be an important tool in modifying and testing different implant designs. FEM is used to study the stress patterns around the dental implant under occlusal loading and the design is modified according to requirements at an analytical level. O Kayabasi used a probabilistic model in order to optimize the design of a dental implant using FEM analysis. This design was further tested after manufacturing on a test rig and the stress values obtained in the case of the analytical method and actual testing were close to each other [4]. FEM is the tool to design the implant and test various design iterations using calculations, but in the case of actual occlusal loading in vivo, biomechanical properties of the jawbone and conditions at the bone-implant interface play a major role in stress distribution analysis. Hence testing the implant design in vitro on test rigs by using materials that mimic the actual mechanical properties and microstructure of the human jawbone is necessary in order to effectively optimize the implant design. Mechanical properties of a large number of materials have been studied to select a suitable material that can serve as a replacement for the human bone. The material selected is expected to have mechanical properties closely aligning with the human bone and the specimens are manufactured to have a microstructure that resembles the human cancellous and cortical bone. Kayla Calvert et al used rigid polyurethane blocks of varying densities to simulate the mechanical properties of the human trabecular bone for in vitro implant testing [5]. Yamaguchi et al used resins as artificial bone models. Polyphenylene sulphide was used to replicate the cancellous bone, while polypropylene replicated the cortical bone. The models were analysed using FEM and compared with experimental data. The conclusion was that FEM analysis results can help in improving in vitro testing methods [6]. Costa Borges et al designed a semiautomatic device for bone implantation and measured the drill force and temperature observed. The device was tested on a polyurethane synthetic block and also on a natural bovine bone. The synthetic bone was used to test various configurations of the machine before its use on the actual bone sample [7]. P. Streckbein et al studied the surface damage of the dental implants during surgery using rigid solid polyurethane foams as artificial bone [8]. Dhatrak et al performed a FEM analysis on a bone-dental implant model by modelling the jawbone as both orthotropic as well as isotropic in order to compare the results obtained in both cases. It was concluded that the orthotropic model closely replicated the jawbone properties as compared to the isotropic model [9,10]. An in vitro test was also conducted by Dhatrak et al using the photoelastic technique by drilling the implants into a resin. The same model was also simulated using FEM analysis and the results were found to be in agreement with each other [11]. Hence in vitro tests are necessary in the design procedure of various surgical techniques and implants.

1.4. Biomechanical properties of the human jawbone

Initially, the properties of the cortical and cancellous bones were studied by using different techniques like actual specimen testing, micro-computed tomography etc. The results obtained over the course of all these research efforts have been summarised to get a range of values for different mechanical properties of cortical and cancellous bone. Carl Misch obtained samples of the human mandible and tested them to obtain a range of values for the mechanical properties of the cancellous bone [12]. The density of the cancellous bone was found to be in the range of 0.85–1.53 g/cm^3 with an average value of 1.14 g/cm^3. The young's modulus varied from 3.5 MPa to 125.6 MPa with an average of 56 MPa. The ultimate compressive strength varied from 0.22 MPa to 10.44 MPa and had an average value of 3.9 MPa. Carl Misch studied the properties of the cortical bone in the human jawbone in compressive, tensile and shear loading scenarios [13]. He found the cortical bone to have a value of 193 MPa in compression, 133 MPa in tension in the longitudinal direction and 68 MPa in torsional shear. The properties of cortical and cancellous bones are considered to be isotropic or orthotropic based on their density values [9–11]. Bone quality is classified into four different categories D1, D2, D3 and D4 based on the density of the cortical and cancellous bones present in the jawbone in a particular case [14]. The porosity of the cancellous bone varies from 50 to 90% and for the cortical bone it is in the range of approximately 10% [15].

2. Material selection for artificial bone

Artificial bone modelling explores the avenue of using synthetic material and advanced manufacturing techniques to replicate the mechanical properties of the human trabecular and cortical bone. The applications of artificial bone modelling include bone grafting and in vitro testing of orthopaedic instruments. In the case of bone grafting, the mechanical properties of the artificial bone have to be closely aligned with that of the bone to prevent stress shielding and a study of bioactivity is also necessary. In case of in vitro testing, mechanical properties and microstructure of the artificial bone are important. This article reviews the research done in the field of artificial bone modelling of the human trabecular bone and compares mechanical properties and microstructure. The manufacturing techniques used in these cases have also been compared. The materials that have been recently used for artificial bone modelling include bio-ceramics, carbon nanotubes, metal scaffolds, polymers and composites [16,17]. The scope of this review is limited to two categories: Polymers and Composites and Metal scaffolds. Every category has a variation in material (Table 1), composition and manufacturing techniques that have been compared and summarized. A comparison with respect to mechanical properties and microstructure has been done in the further sections.

2.1. Polymers and composites

The increase in the use of polymers as a substitute to the human cancellous bone can be attributed to their suitable material properties, accessibility and manufacturing ease. The materials are selected based on their similarity to the human bone in terms of properties such as cellular structure and having a similar mechanical strength to stiffness ratio that closely matches the human cancellous bone. Natural polymers like gelatin, chitosan and sodium alginate are selected in manufacturing scaffolds majorly used for bone grafting since these polymers are bio-compatible and stimulate the growth of bone tissue. A drawback of natural polymers is the poor mechanical properties exhibited by the scaffolds manufactured using only these polymers. Hence recently the use of polymers like poly (lactic acid), polyglycolic acid (PGA) and poly(ε-caprolactone) (PCL) has picked up pace owing to their higher mechanical strength. Recent efforts in artificial bone modelling involve using the properties of both these groups of polymers together in the form of composites. In certain cases, bio-ceramics like bio-active glass are introduced into the polymer structures for added strength without compromising on bio-activity.

Polymer foam – Herex C70.5 was studied by V. Palissery et al to use it as an alternative material to simulate the properties of human cancellous bone. This material was selected since it possesses a similar cellular structure and modulus to strength ratio in compression to that of cancellous bone [18]. P. Bicudo et al tested the use of rigid polyurethane foams known as sawbones to be used as an alternative to trabecular bone. The Sawbones specimens had varying porosity that simulated the different densities of the mandible and were selected after conducting an image analysis evaluation [1]. Kayla L Calvert et al tested seven different densities of rigid polyurethane foam according to the ASTM F 1839-97 standard. The samples were selected after measuring cell parameters using a SEM and stereology [5]. A 13-93 bioactive glass scaffold was selected as an alternative to the human trabecular bone by Qiang Fu et al. Bioactive glass with 13-93 composition (weight %) ($6Na_2O$ $5MgO$ $53SiO_2$ $12K_2O$ $4P_2O_5$ CaO) was selected as its favoured bone formation by osseo-induction and osseo-conduction. This material also possesses higher mechanical strength than polymers like Polylactic Acid (PLA) and Polyglycolic Acid (PGA). Owing to its viscous flow behaviour and a lower crystallization tendency 13-93 glass was chosen over other silicate-based bio-active glass such as 45S5 [19]. Rogina et al manufactured a scaffold of poly (lactic acid) (PLA) and introduced a composite of hydroxyapatite and chitosan gel into the porous structure. PLA was used to strengthen the scaffold since natural polymers fail to provide the required strength to the scaffold. PLA was selected due to its bio-compatibility considering in vivo applications. The chitosan-hydroxyapatite solution was infused into the PLA scaffold to improve bio-activity and osseo-induction as well as osseo-conduction [20]. Mao et al used the combination of a composite of a synthetic polymer and a reinforcing ceramic with a composite of a natural polymer. A scaffold was prepared using a composite of poly(ε-caprolactone) (PCL) and bioactive glass in order to give the artificial bone model its desired mechanical strength. A composite of sodium alginate and gelatin microspheres were also infused into the lattice for improved bio-activity. PCL was selected as it exhibits good bio-degradability and ease of manufacturing associated with its use [21]. An integrated approach was employed by Tcacencu et al to manufacture bone models that replicated the mechanical properties of both the cancellous and cortical bone. A porous structure of PLA was used to replicate the cancellous bone and an apatite-wollastonite structure was used for the cortical bone replication [22]. Cao et al used a composite of zirconia and hydroxyapatite and prepared in a scaffold where zirconia was selected to impart mechanical strength to the artificial bone model while hydroxyapatite was used to facilitate bone tissue formation [23].

2.2. Metal scaffolds

Metals like Ti and Mg have often been used as implant material owing to their bio-compatibility. Metals also have high recoverable strain that more closely aligns with that of the human trabecular bone. The mechanical properties of metals are on the higher side as compared to the bone properties hence porous metallic structures are manufactured to reduce the strength of the metal sample and also to replicate the microstructure of the bone. The pores also facilitate the growth of bone tissue in the case of bone grafting applications. The flexibility provided by metals with respect to manufacturing techniques makes it easier to vary various factors in the porous scaffold that include porosity, pore distribution, trabecular thickness and pore size. Hence metallic porous structures are being used to replace the bone models.

L Hao et al manufactured composite using stainless steel (SS) and hydroxyapatite. Stainless steel 316L was used for its high mechanical strength and hydroxyapatite was introduced in order to provide tissue regrowth abilities [24]. Alireza Vahid et al developed a porous Mg alloy using Mg, Nb and Ta powders with varying compositions. Mg was selected as the primary material as it possesses mechanical properties similar to the human bone as well as its bio suitability. However, due to the lower strength of the Mg porous structure, Nb and Ta were added as reinforcing materials [25]. A TiNbZr based scaffold was selected by Xi Rao et al. A Ti-based alloy foam was developed and Nb and Zr were added to reduce the elastic modulus of the whole scaffold to avoid stress shielding [26]. Mirzaali et al used aluminium to manufacture a porous structure replicating the closed cell structure of the human trabecular bone [27]. Xi Rao et al manufactured a porous scaffold using pure Ti with an average particle size of 30 μm. Titanium was selected owing to its high bio-compatibility and corrosion resistance considering bone grafting approaches. The scaffold was also covered with titania to improve bioactivity [28]. A silicon scaffold was combined with wollastonite, a bio-ceramic material by Kamboj et al. Silicon was used owing to its bio-compatibility and its positive effects on bone mineralization, while wollastonite was used owing to good osteoconductive properties [29]. Li et al used a combination of Ti-Zr-Nb-Sn to manufacture scaffolds using alloy fibres in the solid state. The scaffold was prepared without Ni as a constituent material in order to get a greater recoverable strain as compared to other scaffolds as the natural bone has a higher elastic property [30]. Cockerill et al manufactured porous zinc scaffolds in order to simulate the trabecular bone with tissue regeneration properties. Zinc was selected as its rate of corrosion is equal to the tissue generation rate in the human body and it is also previously present in the human body. The selection of material, in this case was done primarily to facilitate bone regeneration [31].

3. Manufacturing methods

The materials used in the manufacturing of specimens replicating the properties of the human trabecular bone were processed using different manufacturing techniques and each method with different operating parameters lead to a change in the microstructure of the obtained scaffold. Three techniques: metal foaming process, polymeric foam replication and selective laser melting (SLM) are the most common manufacturing techniques (Table 2) being used for the manufacturing of porous polymeric and metal scaffolds. Metal foaming and Polymeric foam replication methods have majorly been used to manufacture metal scaffolds. SLM has been used equally for both polymeric as well as metal porous structures.

3.1. Metal foaming process

Metal Foaming processes are used to create a porous metallic structure for applications such as light weight construction, vibration damping, thermal insulation and chemical filtration. These materials are selected as they have high stiffness values combined with low specific weight and energy absorption capabilities [32,33]. There are two metal foaming methods: Direct Foaming and Indirect foaming. In the direct foaming method, gas bubbles are introduced into molten metal to create a foam structure. In the indirect foaming method, metal powders are mixed with foaming agents (metal hydrides) to create a porous metallic structure. The indirect foaming process involves uniformly mixing the metal powders with foaming agents and then the mixture is compacted to form a dense precursor (billets, plates and rods) without any open porosity. The next step is to heat this material to the melting point of the metal powder. At these temperatures, the foaming

Table 1
Comparison of material properties for artificial bone models.

Polymers & Composites Ref no	Material used	Material Properties	Metal Scaffolds Ref no	Material used	Material Properties
[18]	Herex C70.5 PVC Foam	Similar cellular structure and modulus to strength ratio as cancellous bone	[24]	Stainless steel/ hydroxyapatite	Higher mechanical strength/ improved bone tissue formation
[19]	13–93 bioactive glass	Favoured osseo-induction/conduction. Higher mechanical strength	[25]	Mg/Nb-Ta	Mechanical properties similar to human bone and bio-suitability/ reinforcing materials
[20]	Poly (lactic acid) PLA and chitosan-hydroxyapatite	Bio-compatibility and improved bio-activity	[26]	Ti/Nb-Zr	Bioactive/ Reduction of overall strength of scaffold
[21]	Poly (ε-caprolactone) and bioactive glass composite / sodium alginate and gelatin	Improved mechanical strength/ Improved bone tissue generation	[29]	Silicon/ Wollastonite	Biocompatibility and improves bone mineralization/ Osteo-conductive properties
[22]	PLA/ apatite-wollastonite	Replicated properties of cancellous bone/ Similar properties to cortical bone	[30]	Ti-Zr-Nb-Sn	Greater recoverable strain without Ni as a constituent
[23]	Zirconia/ Hydroxyapatite	Higher mechanical strength/ improved bone tissue formation	[31]	Zinc	Tissue regenerative properties and rate of corrosion equal to tissue generation rate.

agents decompose, forming a porous structure. The foaming parameters can be varied by changing the amount of foaming agent, temperature and rate of heating.

Porous Mg composites were prepared by using indirect foaming method. Mg, Nb and Ta powders that are commercially available were chosen as the base material. Since ammonium hydrogen carbonate (AHC) has a lower decomposition temperature it was used as a foaming agent so as to avoid fusing with Mg powder. The size was AHC particles was limited to a range of 300–500 μm and was mixed with the metal powders in varying proportions. The mixture was then uniaxially compacted at high pressure and temperature to obtain the porous composite. It was observed that micro-porosities were formed in the scaffold due to shrinkage of Mg powder during sintering along with macro-porosities formed by the space holder. This led to uneven cell edge thickness and in some places thin cell edges. The formation of magnesium oxide during the sintering process hindered the fusion of Mg particles thus increasing the porosity of the structure and the size of the pores [25]. A porous structure was manufactured using an alloy of NiTi by Wei et al using the metal foaming process. In order to eliminate the variation in the macropores in space holder sintered porous structures a layer by layer mixing of the NiTi powder and NaCl, which was the space holder was adopted. Ni and Ti powders were used with an average size of 26–95 μm and the NaCl particles had a size of 403 μm. The layer by layer mixing was achieved by the agitation of a part of the crucible while pouring the NiTi powder into the NaCl powder for uniform distribution in batches. The layered powder mixture was then heated to remove the space holder and further sintered at high temperatures to ensure uniform diffusion of Ni and Ti. Reduced micropores were observed on the sample and uniform distribution of macropores was obtained. The porosity of the sample was 19% higher than the calculated value due to the space in the NaCl powder [34].

3.2. Polymeric foam replication

Polymeric replication method is used to manufacture metal foams that replicate the structure of the original foam precisely. The advantage of this method is the ease of availability of the pattern to be replicated in the form of polyurethane foams. The metals powder is converted into a slurry by using a suitable medium. The polymeric foam is then dipped into this slurry and id dried in a centrifuge to remove the excess slurry. This process is repeated until the desired thickness of the slurry is obtained. The slurry covered metal scaffold is then heated to burn out the polymeric foam, followed by sintering of the final metallic foam [35,36].

A powder of 13–93 glass having standard composition was prepared in a crucible and then milled to get the required size of 5–10 μm. The glass particles were then suspended in an aqueous medium. A polyurethane foam of structure resembling that of the trabecular bone was selected as the pattern. This foam was dipped in the aqueous glass suspension and the excess was removed until the desired thickness was obtained. The structure was then dried at room temperature, followed by heating at higher temperatures to decompose the foam. The pore size observed was in the range of 100–500 μm [19]. TiNbZr scaffolds were produced by dipping the polymeric foam in a Ti20Nb15Zr slurry. A polymeric foam of pore size in the range of 0.2–1 mm was selected. These samples were heat treated followed by sintering to add strength to the Ti walls by densifying the metal. The polymeric foam was prepared to match the morphology of the human cancellous bone. The sample replicated the structure of the polymer foam and had a pore size varying from 300 to 1000 μm. A range of pores varying in size was also observed due to the use of the polymeric foam [26]. A Ta-Nb alloy scaffold was manufactured by replicating a polyurethane foam of pore size 500–800 μm. The pore size of the sample was observed to be 400 μm. The porous structure was found to be continuously connected and no gaps were observed. A few micropores were observed due to capillary action in the polymer foam [37].

3.3. Selective laser melting (SLM)

Selective Laser Melting (SLM) is an additive manufacturing process in which a high intensity laser is used to melt and fuse metal powder. The first step in this process is the generation of a CAD file of the required component. The manufacturing process starts with a layer of metal powder on the building platform. The laser is then used to melt and fuse the required regions of the structure. This process is repeated until the final structure is made. The manufacturing parameters that affect the properties of the final sample produced involve scanning speed, power of the laser, layer thickness and hatching space [38].

316L Stainless steel powder having a size range of 30–80 μm was selected as the base material. Hydroxyapatite particles with a size range of 3–8 μm were mixed into the metal powder and 5% HA was added to get a biocompatible composite. The scaffolds were manufactured by experimenting on the samples by varying the manufacturing parameters of the SLM process. It was observed that the strategy of scanning twice was helpful in reaching the amount of energy required to melt the metal powders and in giving the scaffold its mechanical strength. The balling and furrowing phenomena were avoided by reducing overheating by keeping

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the laser power low and having a higher scanning speed. The hatching space also when reduced was helpful in improving the surface finish of the scaffolds. The optimum parameters for SS/HA scaffolds were determined to be: 200 W laser power, scanning speed of 0.13 m/s and a hatching distance of 80 μm [24]. AISI 316L powder was used as the base material with a particle size of 15–50 μm. The powder was melted with a YAG laser with a maximum power of 200 W and a spot diameter of 200 μm with the following conditions: layer thickness 30 μm and scanning rate 0.2 m/s. Balling phenomenon was observed on the strut surface in the structure, this was attributed to improper parameters of manufacturing [39]. A comparison was done between the powder sintering process and SLM technique by Li et al. Porous scaffolds were manufactured by the SLM method with a laser power of 200 W, scanning speed of 0.13 m/s and hatch distance of 80 μm. The obtained samples had a uniform distribution of pore size and pore shape in the case of the SLM technique while the powder sintering method had varied pore shape and sizes [40].

4. Testing of bone models

4.1. Mechanical properties

It is important that the artificial bone models closely replicate the properties of the human jawbone. The properties of the jawbone models must be tested before their application in bone grafting because higher mechanical strength will lead to stress shielding. The artificial bone models manufactured in different studies were tested for their mechanical properties. The testing was done for Ultimate Compressive Strength (UCS) and Young's Modulus values (E). The Ultimate Compressive Strength (UCS) of the cancellous bone lies in the range 0.22–10.44 MPa whereas for cortical bone it lies between 165 and 193 MPa in the human jawbone [13]. It was observed that the bone models with polymers and composite material had properties closer to the human cancellous bone as compared to the metal scaffolds. The mechanical properties for Polymer, composites and metal scaffolds are shown in Table 3.

Table 2
Comparative analysis of the manufacturing techniques.

Manufacturing Technique	Advantages	Disadvantages
Metal foaming	• Economical and fast process • Lower complexity of process • Flexibility in process	• Manufacturing of intricate lattices is not possible • Limited control over the distribution of pore sizes and shape • Porosity of manufactured scaffold is higher due to micro porosities and incomplete linking of struts
Polymeric foam replication	• Better control on pore size and shape distribution • Flexibility in design of scaffold characteristics • Reduced micro porosities leading to dense, connected structures	• Porosity of scaffold depends on polymer structure, hence cannot be controlled • Capillary action leads to micro porosities and uneven densification of scaffold struts • Complete compliance to required pore structure is not obtained
Selective Laser Melting (SLM)	• Higher control on internal (pores) and external (lattice) structures • High manufacturing accuracy • High design flexibility and high scaffold surface finish	• High operating and maintenance cost with lower efficiency • Final product characteristics are highly sensitive to changes in manufacturing parameters • Base material properties also affect product finish and process is slow

4.2. Microstructure studies

The microstructure of the human trabecular bone has been studied previously and various observations about its cell structure and pore size and porosity were documented. The human trabecular bone is observed to be a highly porous structure with a porosity range of 30–95%. The microstructure of the trabecular bone is documented to be a cellular microstructure of interconnecting plates and rods [42]. The morphometric parameters of the bone help define and quantify the bone microstructure. Bone volume fraction (BV/TV), trabecular number (TbN), trabecular thickness (TbTh) and trabecular separation (TbSp) are the morphometric parameters under consideration [43]. The trabecular bone was further studied to document the value of the sizes of the interconnecting rods and plates [42].

A Ti-6Al-4 V structure was manufactured using selective electron beam melting by Peter Heinl et al and the microstructure of this specimen was subjected to μCT measurements and SEM examination. The observed values showed that the structure had a porosity of 61.29% and 80.50% in the hatched structure and the diamond structure respectively. The average pore size ranged from 0.451 mm to 1.22 mm [44]. The bioactive glass scaffolds manufactured by Qiang Fu et al had a porosity of 85% and the microstructure observed bore a very close resemblance to the microstructure of the human trabecular bone [19]. Deepak Pattanayak et al prepared a Ti scaffold sample with porosities in the range of 75–55% and having a pore size of 500 μm to 2 mm [45]. Mirzaali et al prepared an aluminum scaffold and the morphometric properties of the manufactured structure were calculated and compared with that of a bovine bone sample. Trabecular thickness was in the range of 345 μm which was obtained using a μCT analysis [27]. The TiNbZr scaffold manufactured by Xi Rao et al using polymeric foam replication showed a porosity of 75% and the pore size was in the range of 100–500 μm [26]. As seen in Fig. 1 the microstructure of the human cancellous bone has a porous cellular structure. The SEM images of the artificial bone models show that these models were able to replicate the structure of the bone and the size of the pores and interconnected rods varied according to the material and manufacturing process of the specific bone model.

5. Summary and discussion

The recent advancements in artificial bone modelling revolve around bio-compatible scaffolds and their applications in bone grafting. The applications in bone grafting require bio-compatible materials and hence a blend of bio-ceramics and natural polymers is preferred. Hydroxyapatite is a promising bio-active material included in most artificial bone models as it is a constituent of the human bone and stimulates the formation of osteoblasts. In terms of mechanical properties of the scaffolds polymers and composites closely replicate the trabecular bone properties in terms of compressive strength. Metal scaffolds tend to have higher compressive strength values and elastic modulus due to the inherent metallic properties. Hence in order to reduce this strength to prevent stress shielding, appropriate materials are used as alloying materials. In terms of the manufacturing techniques reviewed metal foaming offered little control over the distribution of pore size and shape. Although the layer by layer method in metal foaming eliminated this limitation the micro porosities lead to scaffolds with lower mechanical properties and thin cell walls. The limita-

Table 3
Mechanical Properties of artificial bone models.

Polymers & Composites				Metal Scaffolds			
Ref no	Material	UCS (MPa)	E (GPa)	Ref no	Material	UCS (MPa)	E (GPa)
[18]	Herex C70.5 PVC Foam	0.63	0.039	[25]	Mg + Nb + Ta	10–60	–
[19]	13–93 Bioactive glass	11	–	[26]	Ti + Nb + Zr	15	0.2
[5]	Polyurethane foam	4.7	0.115	[37]	Ta-Nb	83.43	2.54
[20]	Poly (lactic acid)/chitosan	5–8	0.016–0.027	[29]	Silicon/Wollastonite	90–110	–
[41]	PLA/ethyl cellulose/HA	1.57	0.035	[34]	Ni-Ti	30.8	1.1
[21]	PCLA/BG + sodium alginate/gelatin	1.44	–	[30]	Ni free Ti scaffold	5–16.7	0.33–1.05
[23]	Zirconia/HA	15–50	2.5–4.5	[31]	Zinc	6–11	0.1–0.2

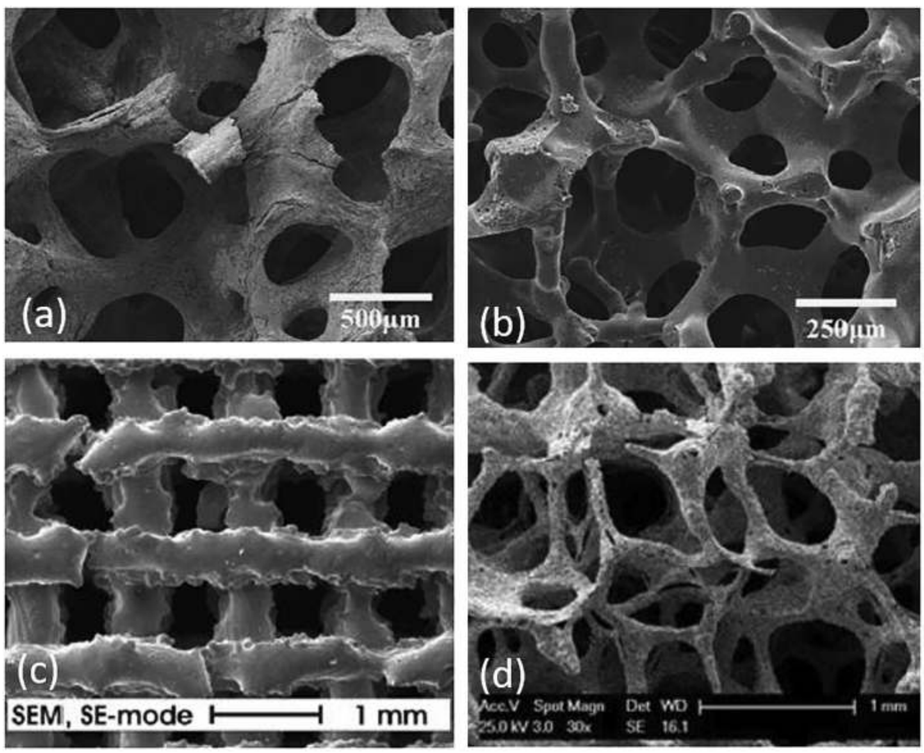

Fig. 1. SEM images of (a) human trabecular bone [19] (b)13-93 bioactive glass scaffold [19] (c)Ti-6Al-4 V structure [44] (d)TiNbZr scaffold [26].

tions of metal foaming were solved to some extent by polymeric foam replication which offered control over the pore distribution but micro porosities were still present in the scaffolds. SLM proved to be the most convenient method as it offered control over the pore distribution and porosity which are important biological parameters for bone grafting applications. SLM offered advantages in the form of control over the final product surface finish by varying the parameters of the process.

The above observations suggest that human bone properties can be replicated by selecting appropriate material and proper manufacturing techniques. The microstructures of the specimens closely aligned with that of the human bone and further improvements can be done in order to get the two structures to replicate each other completely. This can be done by changing the parameters of the manufacturing process in the case of SLM or Polymeric foam replication. SLM parameters for different combinations of material can be studied and tested to obtain standard values for manufacturing parameters for a group of scaffold material. In the case of metal foaming process, parameters like space holder dimensions and proportions can be varied for desired results. The current studies focus on using these bone models as bone grafts and hence are generally aimed at ensuring bio-compatibility and prevention of stress shielding A study focusing on the use of these artificial bone models in the field on dental prosthetics in order to study the stability of a dental implant in-vitro can be considered to be a future scope application.

CRediT author contribution statement

Neha Khasnis: Conceptualization, Data curation, Formal analysis, Investigation, Methodology, Project administration, Software, Validation, Visualization, Writing - original draft. **Pankaj Dhatrak:** Data curation, Formal analysis, Investigation, Project administration, Software, Supervision, Validation, Visualization, Writing - review & editing. **Alekh Kurup:** Data curation, Formal analysis, Investigation, Methodology, Validation, Visualization.

Declaration of Competing Interest

The authors declare that they have no known competing financial interests or personal relationships that could have appeared to influence the work reported in this paper.

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