



Article

Mechanical and Biological Properties of Fe-P Scaffolds Fabricated by Powder Metallurgy Method for Bone Tissue Engineering Applications

Zahra Bostaki, Taghi Isfahani *  and Mohammad Khodaei 

Materials Engineering Group, Golpayegan College of Engineering, Isfahan University of Technology, Golpayegan 87717-67498, Iran

* Correspondence: t.isfahani@iut.ac.ir

Abstract

In this research, Fe-P scaffolds were successfully fabricated by the powder metallurgy method for the first time, using NaCl as the space holder for bone tissue engineering applications, with apparent porosities of approximately 70%. The Fe₃P powder was successfully synthesized by the mechanochemical method under an argon atmosphere using an initial mixture of Fe and P powders. The XRD patterns show that Fe₃P was obtained after sintering the milled powders at 1000 °C. Fe, Fe₃P, and Fe-50 wt% Fe₃P composite scaffolds and bulk pellets were prepared by sintering the milled powder at 1000 °C. Furthermore, the mechanical properties (compression strength) and bioactivity of the Fe-P scaffolds were determined. According to the compression test results, the composite scaffold showed higher compressive strength, lower fracture strain, and higher elastic modulus than the Fe and Fe₃P scaffolds, indicating that adding Fe₃P to Fe improves the mechanical properties. Moreover, among the scaffolds prepared by sintering at 1000 °C, the Fe scaffold exhibited the highest corrosion rate compared to the Fe₃P and composite samples, while the corrosion resistance of the composite sample was 3 times higher than that of the Fe sample. The ICP analysis showed that the amount of Fe released from the bulk pellets during soaking in PBS solution after four weeks was 3220 µg/dL, 4003 µg/dL, and 4774 µg/dL for the composite, Fe₃P, and Fe samples, respectively. The composite sample showed the highest cell viability, while the Fe sample had the lowest. The compressive strength (12.62 MPa) and fracture strain (5.98%) of the porous sintered composite scaffold at 1000 °C were within the range of trabecular bone, while the compressive strength of the composite sample was 17 times higher than that of the Fe sample. Furthermore, the MTS test showed that all the samples had good viability, while the composite sample had the best cell viability. The scaffolds were not cytotoxic. It can be concluded that the mechanical and biological properties of the composite sample were superior to those of the Fe and Fe₃P samples and that it may be a promising candidate for bone tissue engineering applications, especially for trabecular bone replacement.



Academic Editor: Steven Y. Liang

Received: 14 November 2025

Revised: 29 January 2026

Accepted: 3 February 2026

Published: 14 February 2026

Copyright: © 2026 by the authors.

Licensee MDPI, Basel, Switzerland.

This article is an open access article distributed under the terms and conditions of the [Creative Commons Attribution \(CC BY\)](https://creativecommons.org/licenses/by/4.0/) license.

Keywords: Fe-P scaffold; powder metallurgy; mechanical properties; bioactivity

1. Introduction

Nowadays, thousands of surgeries are done every day to substitute or repair tissues that are harmed and damaged by disease or trauma. The developing field of tissue engineering (TE) aims to reconstruct damaged tissues by combining body cells with highly

porous bioscaffolds. The scaffolds serve as templates for tissue regeneration and as guides for tissue growth [1]. Disease, injury, and trauma can cause damage and degeneration of human body tissue, which requires treatment to facilitate repair, replacement, or regeneration. The treatment usually involves transplanting tissue from one area to another in the same patient (autografting) or from one person to another (transplant or allograft). Although these treatments are fundamental and life-saving, both techniques have major problems. In addition to the high cost of harvesting autografts, the procedure is painful, imposes anatomical limitations, and is associated with donor morbidity due to infection and hematoma. Similarly, allografts and transplants have severe limitations due to difficulties in accessing adequate tissues for all patients who need them, the risk of rejection by the immune system, and the possibility of infection or disease transmission from donor to patient. In response, the field of tissue engineering (a term that can be used instead of regenerative medicine) has been introduced and aims to avoid replacement, aiming to regenerate damaged tissues using biological substitutes that restore, maintain, or improve tissue function. It should be noted that tissue engineering is a relatively new field, although the concept of replacing tissues dates back to the 16th century [1].

To address the problems of tissue degradation and regeneration, the multidisciplinary field of tissue engineering combines engineering tools and health science. Cells, scaffolds, and biological factors are the three necessary parts of this field. Scaffolds play an important role as physical matrices that maintain integrity and/or restore damaged tissue, while also serving as sites for the deposition of biological entities. For this reason, the selection of the biomaterial of the scaffold should be based on the type of tissue that needs treatment [2]. The efficiency of a biomaterial implanted in the body depends greatly on its role and nature within the body. The characteristic that distinctly separates a biomaterial from other materials is its ability to contact the relevant human body tissues without causing an undesirable degree of damage to those tissues or their constituents. Tissue engineering products, such as biocompatible scaffolds, must function as substrates that support proper cellular activities, including promoting mechanical and molecular signaling systems to enhance the efficiency of tissue regeneration, without triggering inappropriate local or systemic reactions in the host [3].

Biomaterials are classified into four groups: ceramics, polymers, metals, and composites. Metallic scaffolds with highly porous structures have recently become a new class of material suitable for a wide range of applications. Their high structural similarity to spongy bone and their large surface area have recommended their use as bone substitutes with suitable mechanical properties, as well as for reinforcing monolithic bone [4]. Biodegradable or absorbable metals that corrode in the *in vivo* environment are ideal materials for temporary medical implants that bear load. They provide the necessary mechanical support during the healing process of the treated tissue and are gradually degraded without leaving any implant residue, thus avoiding the need for additional surgery. Magnesium, iron, zinc, and their alloys are considered in this new class of metallic biomaterial [5].

Scaffolds are considered biomaterial replacements for bone defects, which accelerate the recovery of damaged bone tissue. In recent years, the major challenge in biomaterials has been producing highly porous materials with interconnected pores and physical [6] and mechanical properties [7] similar to those of bone tissue.

Scaffolds have several requirements, including biocompatibility, biodegradability, mechanical properties, and structural characteristics.

In the study by Yusop et al., biodegradable metals were used as materials for hard tissue scaffolding in situations where polymers could not be used because of their poor mechanical properties. Scaffolds need to facilitate the formation and maturation of new tissue or organs, and their degradation, cell growth, and temporary mechanical support

must be balanced. Biodegradable metals, such as magnesium and its alloys, have mechanical properties similar to those of natural human bone, and when used as tissue scaffolds, they show promising results. Therefore, their supportive physical properties have made them suitable for orthopedic implants. Porous iron can also be considered a potential scaffold material, but there are very few reports, especially regarding its application in bone tissue. Among the several promising techniques for producing metallic scaffolds, a few processes such as the CO–CO₂ gas foaming powder metallurgy process, the solid–gas eutectic solidification process, and powder metallurgy using a polymer foaming agent have been used. However, due to the very high melting temperature, the solid free-form method cannot be used for Fe, as the NaCl template is destroyed by the excessive heat. The solid free-form method is the most promising approach for producing biodegradable metallic scaffolds with optimal pore morphology for cellular growth and cell proliferation [8].

In the research by Hrubovcakova et al., open-pore Fe-P scaffolds were produced by the replica method, based on the powder metallurgy method, for use as temporary biodegradable materials for bone replacement. The phosphor-alloyed iron scaffold was used to control the corrosion rate and improve mechanical properties. The results showed that the produced scaffolds had an open, interconnected microstructure with a high porosity of up to 89%, similar to cancellous bone. The presence of phosphorus improved the mechanical properties of the scaffolds and reduced the corrosion rate compared to the pure iron scaffold. Therefore, the iron–phosphorous alloyed scaffold appears to be a suitable candidate for biodegradable bone grafts capable of bearing high loads. However, more research is needed to regulate the biocorrosion rate and to experimentally study the in vitro biocompatibility of the material [9].

In the study by Sharma et al. [10], the effect of porosity on the corrosion behavior of porous iron scaffolds manufactured for biodegradable implant applications was investigated. A newly developed process combining 3D printing and pressureless microwave sintering was used to produce two types of iron scaffolds. The random porous iron scaffolds (RPIs) with random microporous structures and topologically ordered porous iron scaffolds with designed interconnected microporous structures were studied. Small isolated micropores, large interconnected micropores, and designed interconnected macropores were obtained. The results indicated that the corrosion of the porous iron scaffold was mainly governed by the diffusion process. It was shown that an increase in random microporosity and a reduction in designed microporosity increased the corrosion rate of the iron scaffold [10]. Ulum et al. [11] stated that biodegradable metals, such as magnesium, iron, and their alloys, are potential materials for medical scaffolds. However, much research on biodegradable metals has focused on optimizing mechanical properties and degradation behavior, without emphasizing improvements in their bioactivity. Therefore, the possibility of improving the biodegradation rate of iron and its bioactivity was studied using different bioactive bioceramics. Iron-based bioceramic composites with reinforcements of hydroxyapatite, biphasic calcium phosphate, and tricalcium phosphate were prepared by mechanical mixing and a sintering process. The degradation studies showed that compared to pure iron, the bioceramic addition reduced the corrosion potential of the composites and slightly increased the corrosion rate [11].

In the study by Wang et al. [12], a composite with an iron matrix and calcium silicate (CS) as reinforcement was produced using the powder metallurgy process. The microstructures, mechanical properties, apatite deposition, and biodegradation behavior of the Fe-CS composites, as well as cell attachment and proliferation on their surfaces, were obtained. The results showed that incorporating calcium silicate particles into iron can be an effective approach for developing iron-based biodegradable bone implants with improved biomedical performance [12].

In the study by Wassila et al. [13], the thermal stability of Fe-8 wt% P milled powders was investigated. Rietveld analysis of the XRD patterns showed that during annealing of the powders up to 210 °C, three phases, including Fe₃P, α-Fe(P) solid solution, and FeP phosphites, could be observed. The results showed that the thermal stability and microstructure of the powders were significantly affected by the milling procedure. The different heat treatments showed that the 21 h milled Fe-8 wt% P powder was unstable. Heat treatment resulted in homogenization, an increase in the α-Fe(P) solid solution fraction, and the formation of Fe-P phosphide phases. The heat treatment up to 450 °C led to the formation of iron oxides, Fe_xP (1 < x < 2) phases, Fe₃P, and α-Fe(P) phases. In the heat-treated powders, the relative amount of the phases changed during the annealing. These results showed that milling is an effective method for controlling the material's thermal stability. However, thermally stable iron phosphide powder was obtained after the heat treatment at 210 °C [13].

Although many efforts have been made in recent years by several researchers to provide scaffolds suitable for tissue engineering, there is a lack of information on the mechanical and biological properties of Fe-Fe₃P scaffolds. Furthermore, limited research is focused on Fe-based scaffolds and phosphor-alloyed iron scaffolds. These scaffolds have better corrosion properties than pure Fe scaffolds, but their poor mechanical properties limit their use. Therefore, it seems that a hard ceramic reinforcement, such as Fe₃P, together with the ductile Fe constituent, can impair the weak mechanical properties of the phosphor-alloyed Fe scaffolds while still retaining the improved corrosion properties provided by the presence of phosphor.

Therefore, this research aims to fabricate Fe-P scaffolds using the powder metallurgy method, and to study the mechanical properties (compressive strength) and bioactivity of the Fe-P scaffolds for the first time. This study also introduces a Fe-P scaffold suitable for long-term bone tissue engineering applications.

2. Materials and Method

In this research, Fe (Sigma-Aldrich (Burlington, MA, USA), purity ≥ 99/9%, particle size < 10 µm) and phosphorous red (Sigma-Aldrich (Burlington, MA, USA), purity ≥ 99/99%, particle size < 10 µm) powders were used to synthesize Fe₃P powder through the solid-state mechanochemical method by mechanical milling and heat treatment of the powders using stoichiometric amounts according to Equation (1):



Appropriate amounts of Fe and P powders were placed inside a NARYA-MPM-2 × 250 H high-energy planetary ball mill (Tehran, Iran) using a steel vial purged with protective Ar gas. Stainless steel balls with diameters of 8, 10, 12, and 15 mm (10 balls of each diameter) were used. Mechanochemical synthesis was carried out at a ball-to-powder ratio of 30:1, a rotation speed of 300 rpm, and a milling time of 12 h. For the heat treatment, a tube furnace with an argon atmosphere was used, with heating rates of 5 and 10 °C/min. The heat treatments of the milled powders were done at 100, 200, and 1000 °C, as presented in Figure 1.

After synthesizing the Fe₃P powder, bulk and porous scaffold samples of Fe, Fe₃P, and the Fe-50 wt% Fe₃P composite were fabricated. The samples for the corrosion, compression, and porosity tests, as well as the microstructural investigation, were prepared in the same way by pressing the powder with a hydraulic pressure of 200 MPa in a cylindrical steel tube with a diameter of 1 cm. The height of the pressed sample was 0.5 ± 0.05 cm. Bulk samples were prepared for the corrosion tests. To manufacture the bulk samples, a 5% PVA (polyvinyl alcohol) solution was used as a binder. Scaffolds with 70% porosity were obtained

using NaCl (purity $\geq 99\%$) as a space holder [14]. For the bulk samples, the powders were mixed with one drop of 5% PVA solution and pressed hydraulically at 200 MPa. The samples obtained were cylindrical, with a diameter of 1 cm and a height of 0.5 cm. Moreover, for the scaffold samples, the amounts of Fe, Fe₃P, and NaCl were obtained by Equations (2)–(5), where V and wt% were the volume fraction and weight percent, respectively. The Fe and NaCl densities were considered as 7.874 and 2.16 g/cm³, respectively.

$$wt\%_{Fe} = \frac{\rho_{Fe} V_{Fe}}{\rho_{Fe} V_{Fe} + \rho_{NaCl} V_{NaCl}} \times 100 \quad (2)$$

$$wt\%_{Fe_3P} = \frac{\rho_{Fe_3P} V_{Fe_3P}}{\rho_{Fe_3P} V_{Fe_3P} + \rho_{NaCl} V_{NaCl}} \times 100 \quad (3)$$

$$\rho_{composite} = \frac{\rho_{Fe} + \rho_{Fe_3P}}{2} \quad (4)$$

$$wt\%_{Fe-50wt\%Fe_3P \text{ composite}} = \frac{\rho_{composite} V_{composite}}{\rho_{composite} V_{composite} + \rho_{NaCl} V_{NaCl}} \times 100 \quad (5)$$

After the pellets (pre-scaffold samples) were prepared, they were placed in an alumina crucible (Base Co, Iran) and heated in an oven at 60 °C, then placed in a tube furnace (Atra furnaces, Tehran, Iran) under an argon atmosphere. The heat treatment was carried out using a three-step heating procedure (Figure 1) to prevent cracks and fractures in the scaffolds during the evaporation of residual water from the NaCl dissolution process, and to avoid thermal shock at the high sintering temperature. This stepped heat treatment cycle was therefore used. This heat treatment cycle provided a slow heating rate at critical temperatures. The samples were first heated to 100 °C at 5 °C/min and held for 30 min. The temperature was then increased to 200 °C at the same rate, with a holding time of 1 h. Finally, the temperature was raised to 1000 °C at a heating rate of 10 °C/min. After reaching this temperature, it was held for 1 h, then cooled to room temperature inside the furnace under argon gas flow.

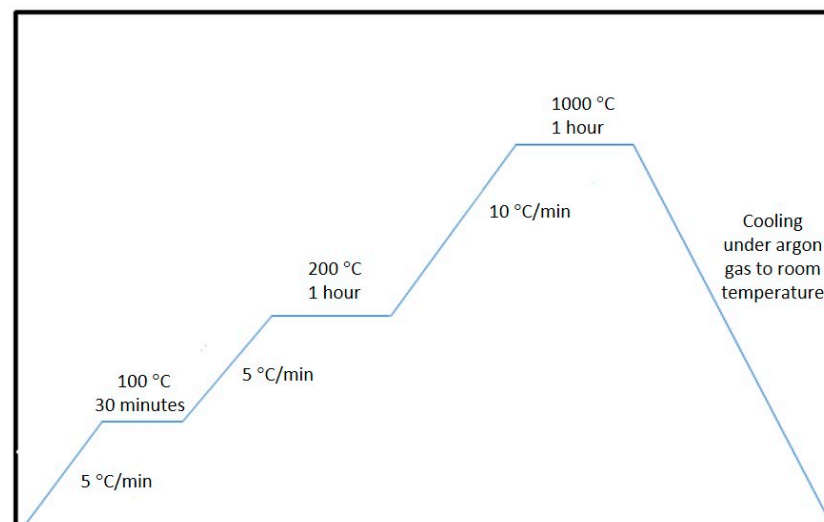


Figure 1. The heat treatment cycle used for scaffold fabrication involved using a tube furnace with an argon gas flow.

To obtain the compression strength, fracture strain, and Young's modulus of the scaffolds, a compression test was applied using the Hounsfield machine (Redhill, UK) with a ramp speed of 0.5 mm/mm. The compressive modulus of the samples was calculated according to the ISO 844-2014 standard [15]. For the determination of the Young's modulus, the slope of the initial linear elastic region was used. For this reason, the linear least squares

fit was applied to the data in the linear region. All the reported values had $R^2 > 0.999$. The compressive strength was considered as the maximum stress sustained before fracture. The fracture strain is the strain at the fracture point. The tests were repeated three times to ensure consistent results.

A Philips (AW-XDM300) XRD machine (Philips, Mississauga, ON, Canada) was used to perform the X-ray diffraction analysis at 50 kV and 40 mA. The diffraction patterns were obtained using Cu K α radiation with $\lambda = 1.542 \text{ \AA}$ in the range of 20–90 degrees, with a step size of 0.05. X'Pert High Score Plus 3.0.5 software in combination with the PDF-2 database (ICCD, 2007) was used to identify the phases. A quantitative phase analysis of the measured powder diffraction patterns was performed using the Topas4-2 program for Rietveld refinement. A scanning electron microscope (SEM, Philips, XI30, Eindhoven, The Netherlands) was used to characterize the microstructure, morphology, and size of the synthesized powders and scaffolds.

In this research, L929 human fibroblast cells were used to study the biocompatibility of the cells on the surface of the sintered samples. The cells were cultured using flasks of 25 cm^2 . After the flask was filled and passaged twice, the cells were detached from the bottom of the container and counted. The samples were sterilized using UV radiation for 30 min and then placed in the cell culture container. Finally, an appropriate number of L929 cells was seeded on the surface of the sterilized sample.

2.1. Study on Cell Viability

To assess cell activity and viability in the samples, an MTS assay was performed. The measurement of biological activity of the cells was based on the fact that tetrazolium salt is converted into a colored salt soluble in cell culture medium by the enzyme present in the mitochondria of living cells, which ultimately determines the absorption intensity of the color created by this substance, indicating the presence of live and active cells [16].

To perform the MTS test, the supernatant medium was first replaced with a serum-free medium containing 20% MTS, and the samples were incubated for 4 h. The colored medium was then transferred to a 96-well plate, and the amount of dye absorbed by the mitochondria in living cells was measured at a wavelength of 492 nm. During the transfer of the medium to the 96-well culture vessel, the supernatant medium from each group of L929 cells was distributed into three wells to reduce the error rate (100 microliters per well). The reduction in the error rate is because the test was repeated three times, with the average value reported. The optical densities were analyzed using Excel, and a corresponding graph was created and examined. The MTS test was performed over one week at four time points: 4 h, 2, 4, and 6 days after cell culture. Each sample was tested three times.

The porosimetry test on the scaffolds was conducted using the Archimedes method based on flotation in liquid. For each sample, the test was repeated three times for the Fe, Fe₃P, and Fe-50 wt% Fe₃P scaffolds, and the mean value was reported. For this reason, the dry scaffolds were first weighed (W_d), then the weight of the scaffolds during their placement in water (W_s) was measured, and finally, the weight of the scaffold immediately removed from the water (W_w) was recorded. The apparent porosity and real porosity were measured according to ASTM B962 standard [17] by Equations (6)–(8), respectively.

$$\text{apparent porosity} = \frac{W_w - W_d}{W_w - W_s} \times 100 \quad (6)$$

$$\text{real density} = \frac{\rho - B}{\rho} \times 100 \quad (7)$$

$$B = \frac{W_d}{W_w - W_s} \times 100 \quad (8)$$

where B is the bulk density, and ρ is the theoretical density.

The degradation test was conducted for one month, and inductively coupled plasma spectroscopy (ICP) was used on each sample to compare the amount of Fe released by the different samples. The tests were applied every week for a one-month period.

2.2. Statistical Analysis

All data from the quantitative tests are reported as the mean of the samples with the standard deviation, and $p < 0.05$ is considered significant.

3. Results and Discussion

3.1. X-Ray Diffraction Analysis

As mentioned in Section 2, to synthesize the Fe_3P powder, Fe and P powders were milled for 12 h under an argon atmosphere and then heat-treated at 1100 °C under argon flow. Figure 2 shows the whole pattern Rietveld analysis of the X-ray diffraction pattern of Fe_3P powder synthesized by the mechanochemical method and heat-treated at 1000 °C. The whole-pattern Rietveld analysis indicates that the synthesized Fe_3P (CIF number 1561394) was pure, with no unwanted phases present in the XRD pattern.

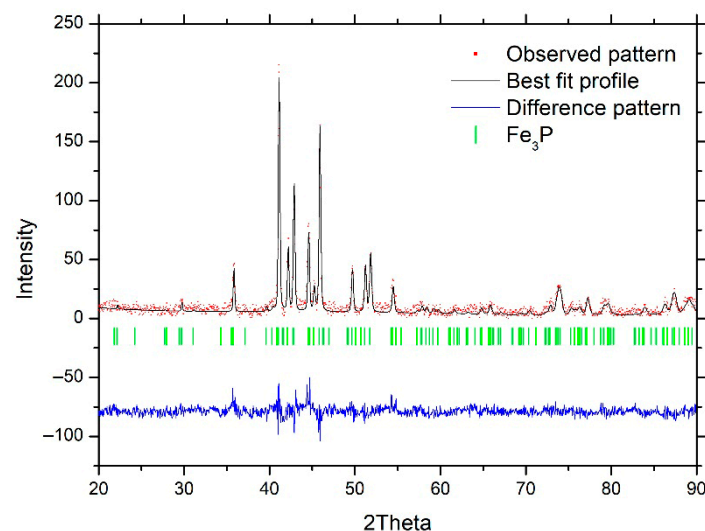


Figure 2. The XRD pattern of the Fe_3P powder synthesized by the mechanochemical method.

3.2. Scanning Electron Microscope

As shown in Figure 3, the prepared porous Fe, Fe_3P , and Fe-50 wt% Fe_3P composite scaffold samples, heat-treated at 1000 °C, were examined by SEM to observe the scaffold morphology. The SEM images show that the macropores had sizes of around 500 microns, which is in good agreement with the size of the used salt particles as space holders. The composite matrix (Figure 3c) also contained micropores smaller than 10 microns, which can aid cell nutrition. These macropores, formed by removing NaCl space holders during the washing process, were interconnected within the scaffolds, facilitating cellular diffusion and cell nutrition. Porosity is considered a biological characteristic of cellular scaffolds because it enables the delivery of nutrients to cells and the removal of waste produced by cellular interactions. It is also suitable for the growth and cellular reproduction of pores and holes for cell proliferation.

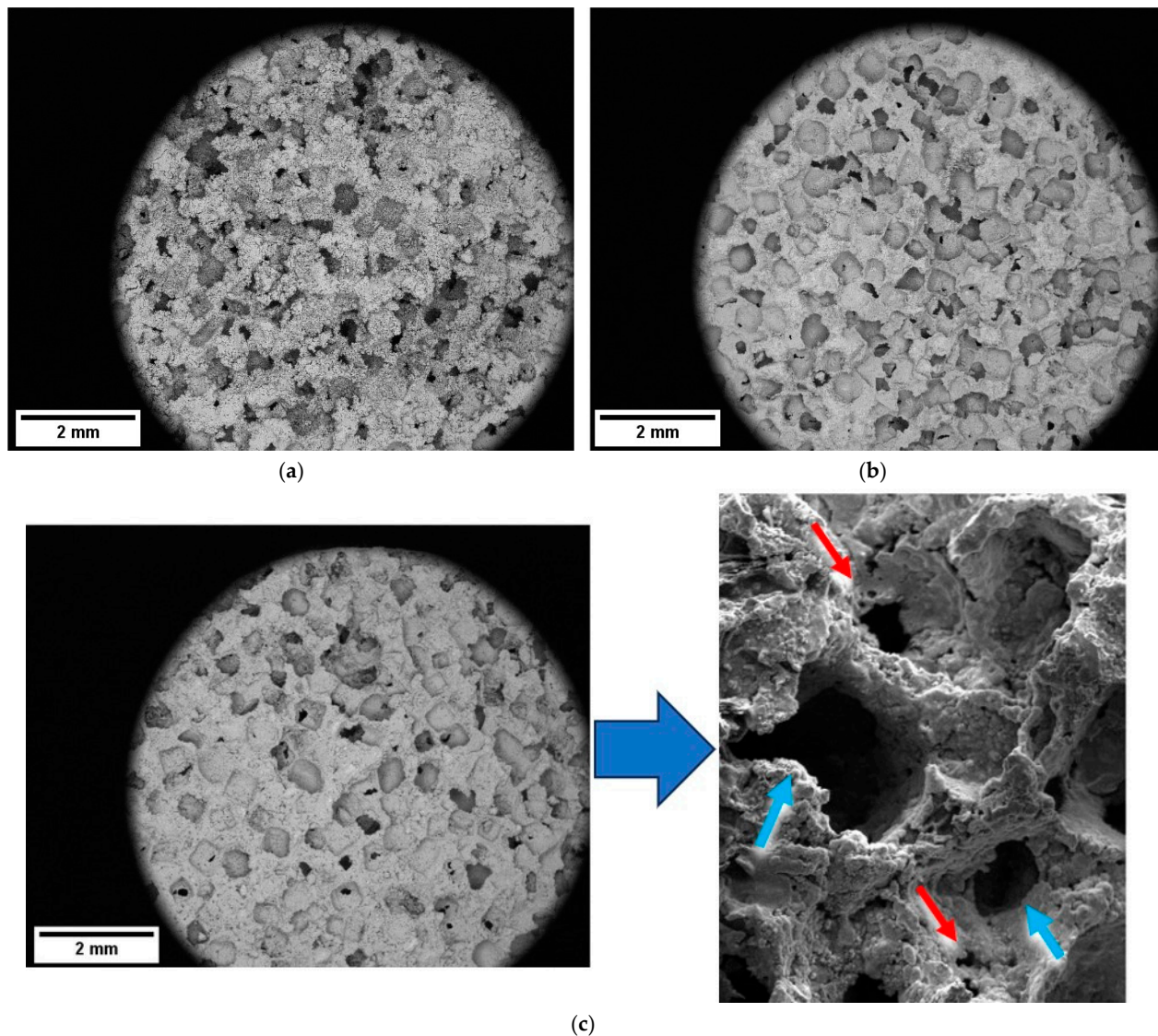


Figure 3. SEM images of the porous scaffolds: (a) Fe, (b) Fe_3P , and (c) Fe-50 wt% Fe_3P , and a higher magnified image (blue arrows: microporosity; red arrows: microporosity). Samples sintered at 1000 °C.

3.3. Porosimetry Test

As mentioned, the Archimedes method was used to obtain the porosity value according to Equations (6)–(8), and the values are shown in Table 1.

Table 1. Results of the Archimedes porosimetry test of porous Fe, Fe_3P , and composite samples.

	Apparent Porosity (%)	Real Porosity (%)	Closed Porosity (%)
Fe	71.7 ± 1.7	75.8 ± 0.13	4.1
Fe_3P	72.5 ± 0.6	74.4 ± 0.7	1.9
Composite	70.3 ± 0.7	72.25 ± 0.87	1.98

As the porosity value directly affects the mechanical properties, it was crucial to ensure that the NaCl space holder was completely removed during the washing procedure and that interconnected pores were obtained. As presented in Table 1, the results indicate that the desired porosity and interconnectivity of the pores were achieved. The Fe, Fe_3P , and composite samples showed apparent porosities of 71.7, 72.5, and 70.3, respectively.

3.4. Mechanical Properties

The stress–strain curves of the scaffold samples are presented in Figure 4, and the mechanical properties are presented in Table 2. The fracture strains of the Fe, Fe₃P, and Fe-50 wt% Fe₃P composite samples were 27.33%, 12.41%, and 5.98%, respectively. The pure Fe sample exhibited the highest fracture strain among the samples, due to the ductile nature of metallic Fe with a BCC structure and the absence of the brittle intermetallic Fe₃P phase in this sample. In the Fe-50 wt% Fe₃P composite scaffold sample, the presence of Fe₃P particles in the Fe matrix provided reinforcement and acted as a barrier to dislocation mobility, thereby increasing the strength. Furthermore, the elastic moduli of pure iron, Fe₃P, and the composite sample were obtained as 0.05 MPa, 0.27 MPa, and 0.76 MPa, respectively. Therefore, the composite had the highest elastic modulus, while the pure Fe scaffold sample had the lowest value. The compression strength values of the pure Fe, Fe₃P, and composite samples were 0.74, 1.31, and 12.62 MPa, respectively. Therefore, the composite scaffold had the highest compression value and thus had the best effect on the mechanical properties. This may be due to the reinforcing effect of the composite. Fe is soft and tough, while Fe₃P is hard and brittle. A combination of these two materials can provide a high-strength composite.

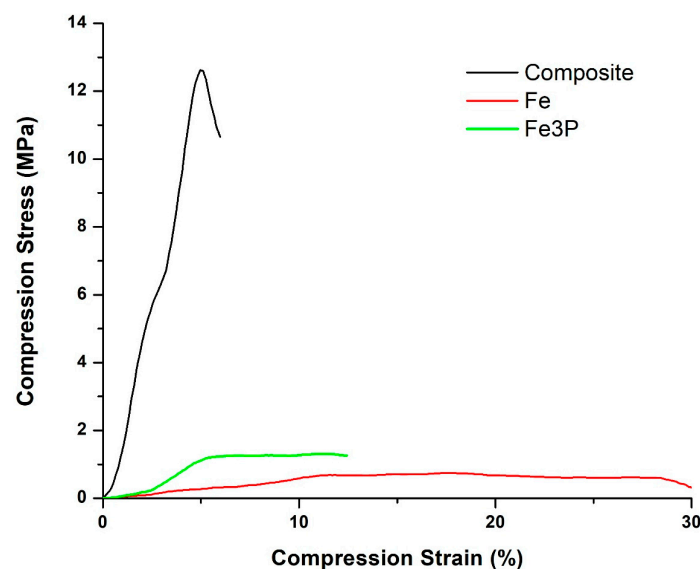


Figure 4. Compression stress and strain curves of porous Fe, Fe₃P, and Fe-50 wt% Fe₃P composite scaffolds sintered at 1000 °C.

Table 2. Compression test results of porous scaffolds sintered at 1000 °C.

	Compression Strength (MPa)	Fracture Strain (%)	Elastic Modulus (MPa)
Fe	0.74 ± 0.11	27.33 ± 2.01	0.05 ± 0.012
Fe ₃ P	1.31 ± 0.34	12.41 ± 0.17	0.27 ± 0.008
Fe-50 wt% Fe ₃ P	12.62 ± 2.7	5.98 ± 0.3	0.76 ± 0.043

The compression strength of all the porous scaffolds sintered at 1000 °C was in the range of trabecular bone (0.1–16 MPa) [18], while the fracture strain of the composite scaffold was the only sample in the range of trabecular bone (5–7%) [19,20]. Furthermore, the compressive elastic modulus of trabecular bone should be in the range of 0.12–1.1 MPa. These results indicate that the composite sample is a suitable candidate for the replacement or repair of trabecular bone from a mechanical perspective. In the research by Hrubovcakova et al. [9], it was shown that iron scaffolds alloyed with phosphorus using the replica

method usually exhibit increased yield strength compared to pure iron scaffolds, and it was demonstrated that phosphorus is a suitable alloying element for scaffold production. The improved results were due to its beneficial effect on compression sintering, which significantly increased the mechanical strength [9].

It should be noted that in bulk material, the elastic modulus of the composite is always between those of the matrix and the reinforcement and is never more than the constituent with the higher value. Moreover, the modulus of the bulk material is determined by the volume fraction, reinforcement geometry, orientation, and the interface bonding. In the case of porous materials, especially multicomponent materials, it is a very complicated issue. Many researchers, such as Phani and Niyogi [21], have noted that predicting and describing the relationship between the composition, mechanical properties, and structure of porous materials is not straightforward [22]. Until now, understanding of the relationship between macroscopic mechanical behavior and microstructural properties (such as porosity) has been far from satisfactory [23].

In the case of porous material, the total volume and type of porosity (open or closed), the shape of porosity, randomness, uniformity of distribution, geometrical arrangement of the elements (angle of intersection), and the wall shape (e.g., curvature, cross-sectional shape, and uniformity) determine the elastic modulus value. Furthermore, at a larger scale, the geometrical arrangement of pores is crucial and severely affects the elastic modulus [24,25].

It has been shown that the elastic properties of porous composite materials can be determined from the elastic properties of their components in several ways: statistically using scattering theory, through variational principles, or by assuming specific geometries for the material under consideration. It is noteworthy that specific geometry models must be considered if useful results are to be obtained [26].

In the present research, the higher elastic modulus of the composite sample compared to its components can likely be attributed to the porosity characteristics, such as the shape, geometric arrangement, and wall shape of the pores formed after sintering the Fe and Fe₃P particles together at 1000 °C. Furthermore, it should be mentioned that in this research, the tensile tests were repeated three times for all samples. In Table 2, the compression strength, fracture strain, and elastic modulus are presented, along with their error ranges. The presented results reveal the reproducibility and consistency of the data.

3.5. Corrosion Properties

It is well known that Fe is an important element with vital roles in the human body, such as storage and reduction of RNA and DNA, cell growth, and oxygen transport [27]. At the start of the twenty-first century, Peuster et al. [28] investigated Fe as a potential absorbable metal, indicating that it does not produce toxic corrosion products when used to replace biological tissues. However, in addition to having high porosity and pore interconnectivity, the porous scaffolds must maintain structural integrity. Therefore, their corrosion rate should match the new bone regeneration and healing time to provide a suitable transition of the load from the scaffold to the bone [29]. The main drawback of using Fe in bioresorbable applications is its excessively slow corrosion rate in physiological environments [8,30–32]. In other words, corrosion or degradation behavior is one of the most critical criteria for evaluating metallic materials for clinical applications. Therefore, much research has focused on modifying the corrosion rate of Fe to overcome this obstacle and provide successful clinical applications [33–36]. The published results indicate that pure unmodified porous Fe scaffolds have higher corrosion rates (0.03–2.25 mm year^{−1}) compared to those of reported non-porous bulk pure Fe counterparts (0.008–0.242 mm year^{−1}) [37–44]. This demonstrates the advantage of a porous structure in increasing the otherwise too-slow corrosion kinetics of Fe. Grain refinement achieved through additive manufacturing (AM),

severe plastic deformation (SPD), and mechanical milling techniques may also contribute to the increase, due to the greater number of grain boundaries [45,46] and higher surface reactivity to corrosion [47].

Tafel curves obtained from the corrosion test using SBF solution are presented in Figure 5a. The pH of the SBF solution was 7.4, and its formulation is shown in Table 3. As shown in Figure 5a, the Fe sample had the highest current density (highest degradation rate) of 0.1382. Furthermore, according to the Tafel curves (Figure 5a), the corrosion current density (i_{corr}) (Figure 5b), and the corrosion rate of the samples (Table 4) obtained by the corrosion test in SBF solution indicates that the lowest corrosion rate belonged to the composite sample. However, as expected, the corrosion rates of the samples were not the same, but all had acceptable values. Therefore, the different samples can be used depending on the implantation location, where varying corrosion rates are required. Also, according to the patient's age, a different corrosion rate of the implant is required. For example, in the case of older men, tissue regeneration is too slow, so a low-corrosion-rate implant is needed. This indicates that all the samples can be useful when corrosion properties are needed. Furthermore, several studies have shown that the use of porous scaffolds increases the corrosion and degradation rates compared to the bulk state [8,48].

Table 3. The concentration of ions in the SBF solution.

Ion	Concentration (mM)
Na ⁺	142.0
K ⁺	5.0
Mg ²⁺	1.5
Ca ²⁺	2.5
Cl [−]	147.8
HCO ₃ [−]	4.2
HPO ₄ ^{2−}	1.0
SO ₄ ^{2−}	0.5

Table 4. Corrosion results of the bulk Fe, Fe₃P, and Fe-50 wt% Fe₃P composite samples sintered at 1000 °C, tested using SBF solution.

	β_c (mv)	β_a (mv)	E_{cor} (mv)	i_{corr} (mA/cm ²)	Corrosion Rate (mm/Y)
Fe	−216 ± 50.7	345.4 ± 42.1	−485.7 ± 22.7	0.1382 ± 0.051	2.3154 ± 0.17
Fe ₃ P	−371.3 ± 53.3	309.6 ± 57.4	−535.1 ± 20.1	0.0786 ± 0.0058	1.3164 ± 0.02
Fe-50 wt% Fe ₃ P	−190.2 ± 41.2	125 ± 33.7	−418 ± 31.3	0.0467 ± 0.004	0.78195 ± 0.08

According to the research of Daud et al. [49], the pure Fe sample had the highest current density (most degradation) compared to HA-Fe and HA/PCl-Fe samples [49]. In our research, it was also found that the maximum current density was highest for the Fe sample, followed by Fe₃P, and lowest for the composite sample. Based on the research by Ulum [11], the corrosion potential of Fe-based composites was lower than that of pure Fe samples, while the corrosion rate increased slightly. Moreover, research by Sharma and Pandey [10] showed that porosity value and morphology significantly affected corrosion resistance.

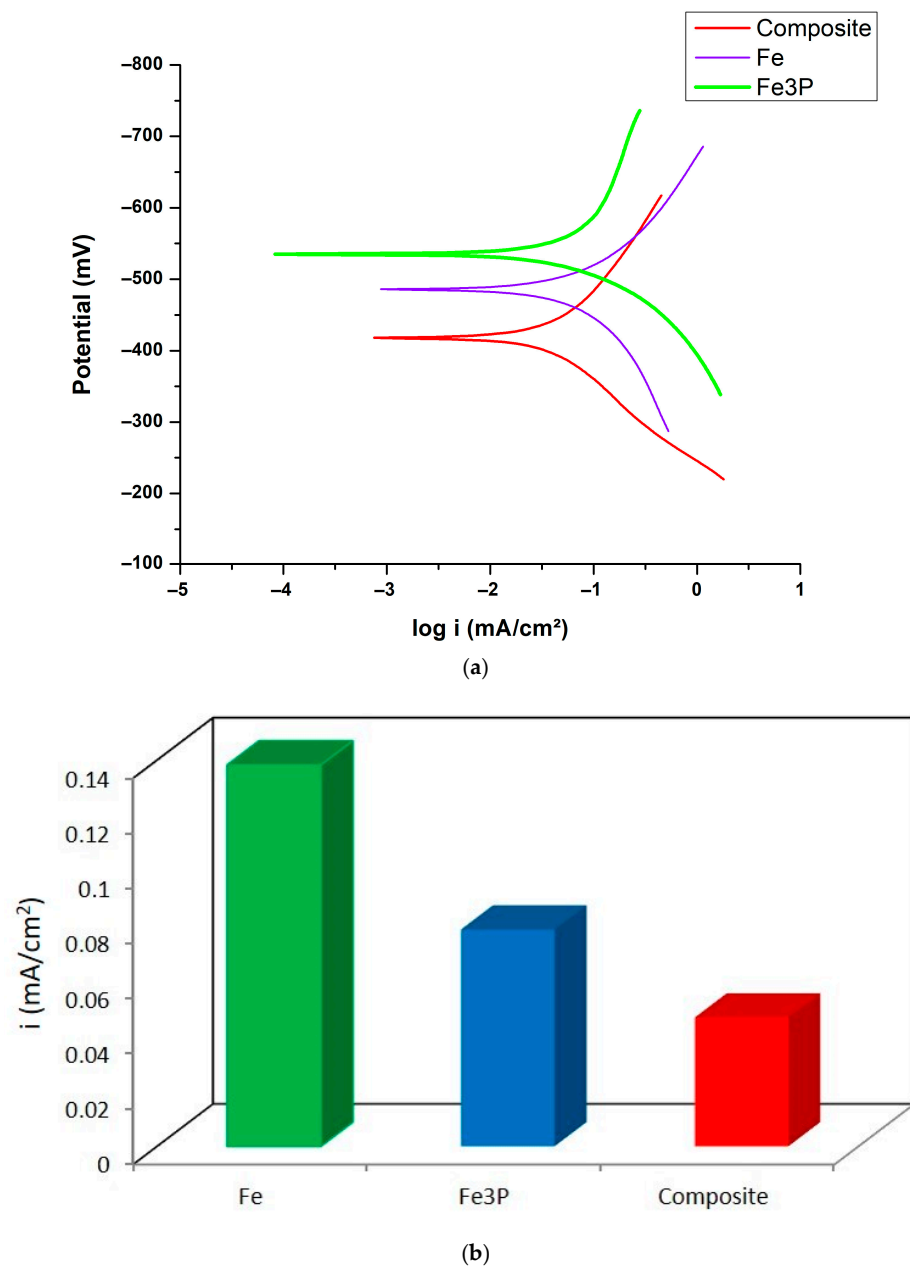


Figure 5. (a) Tafel polarization curves of bulk Fe, Fe₃P, and Fe-50 wt% Fe₃P composite samples sintered at 1000 °C tested using SBF solution. (b) Bar plots showing the corrosion current density of the bulk Fe, Fe₃P, and composite samples sintered at 1000 °C.

In Figure 5, SEM images of the compact Fe, Fe₃P, and composite samples before and after the corrosion tests are presented. As seen by comparing the Fe samples before (Figure 6a) and after the corrosion test (Figure 6b), large amounts of corrosion products with elongated morphology appeared alongside the round spherical Fe particles after the corrosion test. This agrees with the Tafel test, which indicated a high corrosion rate for Fe. In Figure 6c,d, SEM images of the Fe₃P samples are presented. The Fe₃P samples had large and sharp edges. It can be seen that after the corrosion test, the corrosion products differed from those of the Fe samples. The Fe₃P appeared to have thin, adherent surface layers, likely due to the presence of phosphorus. Figure 6e,f corresponds to the composite sample before and after the corrosion test, respectively. As seen in the composite sample, the corrosion differed from that of the Fe and Fe₃P samples. In these samples, corrosion had flattened and smoothed sharp edges, resulting in non-uniform coverage of the composite.

It seems that a galvanic couple formed in the composite sample. In an electrolyte, the two phases have different electrochemical potentials. Fe is more anodic, meaning it is active at a lower potential and serves as the sacrificial anode, whereas Fe_3P is more cathodic (noble, with a higher potential).

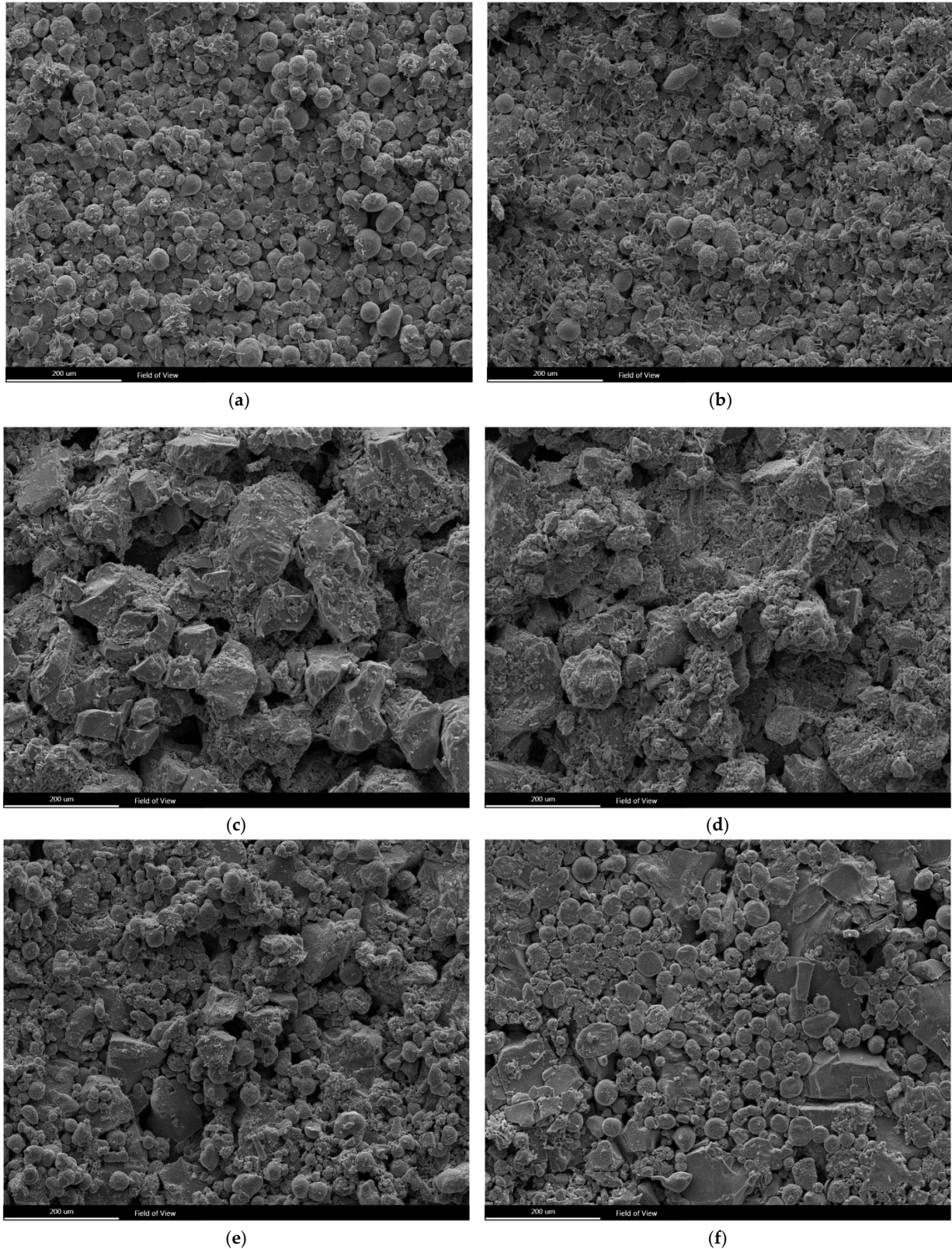


Figure 6. SEM images of Fe sample (a) before and (b) after corrosion test, Fe_3P sample (c) before and (d) after corrosion test, and composite sample (e) before and (f) after corrosion test.

It should be noted that Fe_3P (iron phosphide) is an intermetallic compound, and its corrosion behavior is undoubtedly different from that of pure Fe. Fe_3P is cathodic and more noble compared to pure Fe. If the Fe_3P powder contains minor amounts of Fe due to incomplete synthesis or if Fe_3P has inhomogeneous regions, the Fe_3P phase will act as a cathode, resulting in the dissolution of iron-rich (more anodic) regions. This accelerates the dissolution of the iron-rich regions. However, the corrosion of Fe_3P causes the oxidation of both Fe and P, with Fe likely forming iron oxides and oxyhydroxides, such as Fe_2O_3 , Fe_3O_4 , and FeOOH . The phosphorus (P) can probably be oxidized to phosphate ions (PO_4^{3-}). Depending on the pH, phosphate can remain soluble or form precipitates such as iron phosphate (FePO_4).

Moreover, according to Figure 6, the Fe, Fe_3P , and composite samples do not have the same porosity. It is well known that compact sintered material is not fully dense and has inherent porosity. The shape, size, and amount of porosity largely depend on the particle size, particle size distribution, and particle morphology. Higher porosity will almost certainly result in a significantly higher corrosion rate. In fact, for powder compacts, the porosity is often the single most dominant microstructural factor controlling the corrosion rate. This is even more than the composition in many cases. For example, two compacts with the same composition but different porosity will behave like two completely different materials. The porosity has a much larger internal surface. This increased surface area provides more sites for electrochemical reactions, resulting in higher measured weight loss and faster degradation. Moreover, each pore can act as a natural crevice. This can also create a localized galvanic cell. The localized corrosion rates inside the pores can be orders of magnitude higher than those on a free surface. Therefore, when comparing the corrosion results of the compact samples, it must be kept in mind that the porosity present in the samples is different.

3.6. Degradation Test

The degradation test was conducted for one month, and the results were recorded weekly, as shown in Table 5 and Figure 7. SBF solution was used to perform the degradation test. The tests were performed at 37 °C using a bain-marie for one month. The Fe released from the samples was determined by ICP analysis each week (Table 4).

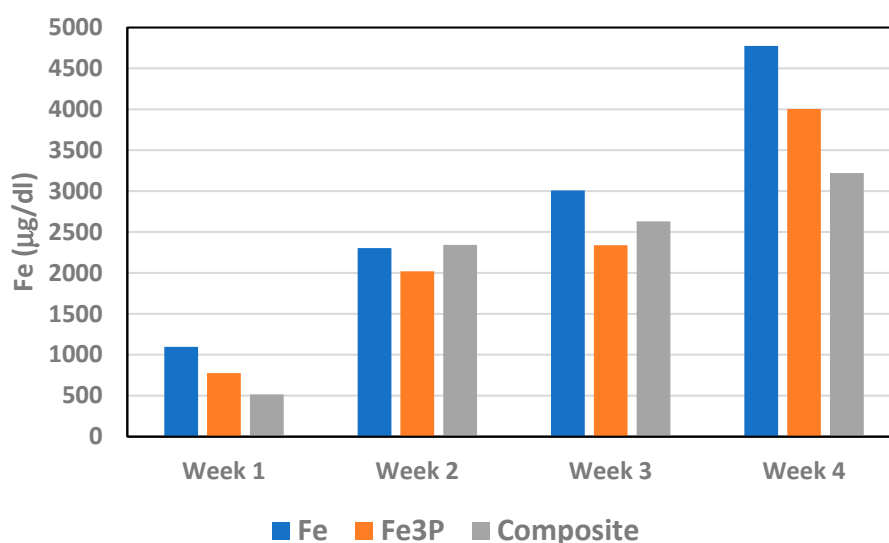


Figure 7. Cumulative curves of the Fe element amount over four weeks in bulk samples of Fe, Fe_3P , and composite samples sintered at 1000 °C.

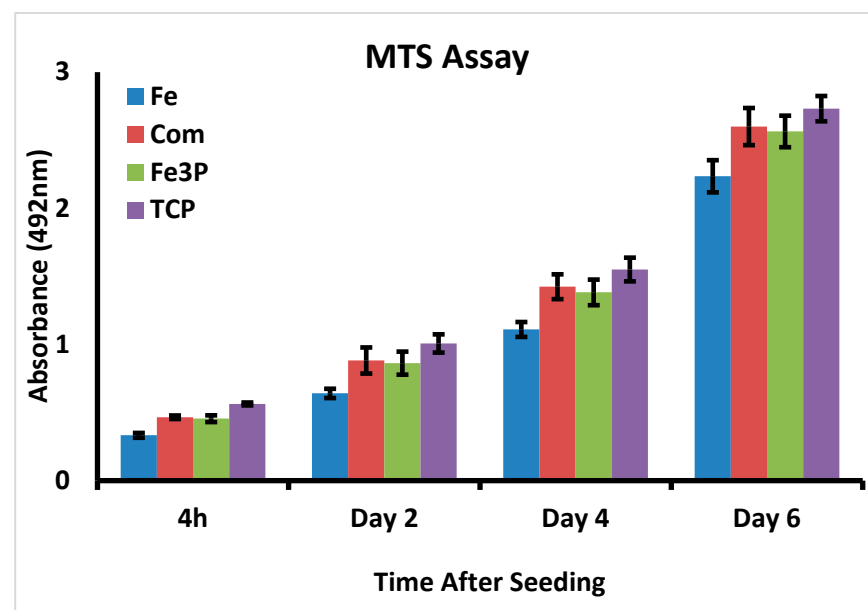
Table 5. The Fe element amounts in the Fe, Fe₃P, and composite samples heat-treated at 1000 °C, as determined by ICP analysis for each of the 4 weeks.

Amount of Fe	Fe	Fe ₃ P	Composite
First week	1096	775	516
Second week	2303	2020	2342
Third week	3007	2338	2630
Fourth week	4774	4003	3220

Figure 7 presents bar plots showing the Fe element values in the solution over 4 weeks. According to this plot, it is obvious that in the first week, the amount of Fe element released from the Fe sample was higher than in the other two samples, while in the second week, the amount of the Fe element released due to the degradation of the composite sample was higher than in the other two samples. In the third and fourth weeks, the amount of the Fe element in the Fe sample was higher than in the other two samples.

3.7. MTS Test

In Figure 8, the results of the cell viability of L929 human fibroblast cells (MTS test) are presented. According to the bar plots in Figure 8, the vertical axis shows the number of absorbed live cells. As the height of the graph increases, the number of live cells increases. The test was performed at four different times (4 h, 2 days, 4 days, and 6 days), and cell viability was studied at each of these time points. The samples were compared with a standard positive control of tri-calcium phosphate TCP, which is a bioactive bioceramic. Cold-pressed TCP powder with a particle size smaller than 80 micrometer was subjected to pressure up to 300 MPa for this comparison. The porosity of bulk TCP was measured to be smaller than 1%. It can be seen that tri-calcium phosphate compared to the other samples had better viability across all days. The composite sample had the highest viability, Fe₃P showed good viability, and pure Fe had the lowest viability. Therefore, it can be concluded that all samples showed good viability and the scaffolds were not cytotoxic.

**Figure 8.** The results of the cell viability test of L929 human fibroblast cells (MTS test).

4. Conclusions

1. Fe₃P powder was successfully prepared using the mechanochemical method by milling Fe and P powders for 12 h at 300 rpm with a ball-to-powder ratio of 30:1, followed by sintering at 1000 °C.
2. The compression test of the samples showed that the Fe-50 wt% Fe₃P composite sample sintered at 1000 °C met the requirements for trabecular bone. Its compressive strength, fracture strain, and elastic modulus were 12.62 MPa, 5.98%, and 0.76 MPa, respectively.
3. The corrosion test results obtained using simulated body fluid solution show that the corrosion rate of Fe was higher than that of the composite and Fe₃P samples sintered at 1000 °C. The corrosion rate of Fe was 3 times higher than that of the composite samples.
4. ICP analysis of the SBF solution showed that after the 4th week of immersing scaffolds in SBF solution, the amount of Fe element was 3220 µg/dL, 4003 µg/dL, and 4774 µg/dL in the composite, Fe₃P, and Fe samples, respectively.
5. The MTS test results show that the composite samples had the highest cell viability, Fe₃P had moderate cell viability, and pure Fe had the lowest. The scaffolds were not cytotoxic.

Author Contributions: Conceptualization, T.I.; methodology, T.I. and M.K.; formal analysis, T.I., Z.B. and M.K.; investigation, T.I. and Z.B.; resources, T.I. and Z.B.; data curation, T.I., Z.B. and M.K.; writing—original draft preparation, T.I., Z.B. and M.K.; writing—review and editing, T.I., Z.B. and M.K.; supervision, T.I.; project administration, T.I. and Z.B. All authors have read and agreed to the published version of the manuscript.

Funding: This research received no external funding.

Data Availability Statement: The data and materials that support the findings of this study are available from the corresponding author upon reasonable request.

Conflicts of Interest: The authors declare no conflicts of interest.

References

1. O'Brien, F.J. Biomaterials & scaffolds for tissue engineering. *Mater. Today* **2011**, *14*, 88–95. [\[CrossRef\]](#)
2. Mata, N.A.; Arango-Ospina, M.; Velasquez, P.; Murciano, A.; De Aza, P.N.; Boccaccini, A.R. Effect of Sr, Mg and Fe substitution on the physico-chemical and biological properties of SiCaP multilayer scaffold. *Boletín Soc. Española Cerámica Vidr.* **2021**, *62*, 145–159. [\[CrossRef\]](#)
3. Saqaei, M.; Fathi, M.; Edris, H.; Mortazavi, V. Preparation and biocompatibility evaluation of bioactive glass–forsterite nanocomposite powder for oral bone defects treatment applications. *Mater. Sci. Eng. C* **2015**, *56*, 409–416. [\[CrossRef\]](#) [\[PubMed\]](#)
4. Glorius, S.; Nies, B.; Farack, J.; Quadbeck, P.; Hauser, R.; Standke, G.; Rößler, S.; Scharnweber, D.; Stephani, G. Metal Foam–Bone cement Composites: Mechanical and biological properties and perspectives for bone implant design. *Adv. Eng. Mater.* **2011**, *13*, 1019–1023. [\[CrossRef\]](#)
5. Bagha, P.S.; Khakbiz, M.; Sheibani, S.; Hermawan, H. Design and characterization of nano and bimodal structured biodegradable Fe-Mn-Ag alloy with accelerated corrosion rate. *J. Alloys Compd.* **2018**, *767*, 955–965. [\[CrossRef\]](#)
6. Md Yusop, A.H.; Ulum, M.F.; Al Sakka, A.; Nur, H. Current Status and Outlook of Porous Zn-based Scaffolds for Bone Applications: A Review. *J. Bionic Eng.* **2022**, *19*, 737–751. [\[CrossRef\]](#)
7. Sadeghzade, S.; Emadi, R. Improving the mechanical and bioactivity of hydroxyapatite porous scaffold ceramic with diopside/forsterite ceramic coating. *Nanomed. J.* **2019**, *6*, 50–54.
8. Yusop, A.H.; Bakir, A.A.; Shaharom, N.A.; Abdul Kadir, M.R.; Hermawan, H. Porous biodegradable metals for hard tissue scaffolds: A review. *Int. J. Biomater.* **2012**, *2012*, 641430. [\[CrossRef\]](#)
9. Hrubovčáková, M.; Kupková, M.; Džupon, M. Fe and Fe-P foam for biodegradable bone replacement material: Morphology, corrosion behavior, and mechanical properties. *Adv. Mater. Sci. Eng.* **2016**, *2016*, 6257368. [\[CrossRef\]](#)
10. Sharma, P.; Pandey, P.M. Corrosion behavior of the porous iron scaffold in simulated body fluid for biodegradable implant application. *Mater. Sci. Eng. C* **2019**, *99*, 838–852. [\[CrossRef\]](#)
11. Ulum, M.; Arafat, A.; Noviana, D.; Yusop, A.; Nasution, A.; Kadir, M.A.; Hermawan, H. In vitro and in vivo degradation evaluation of novel iron-bioceramic composites for bone implant applications. *Mater. Sci. Eng. C* **2014**, *36*, 336–344. [\[CrossRef\]](#)

12. Wang, S.; Xu, Y.; Zhou, J.; Li, H.; Chang, J.; Huan, Z. In vitro degradation and surface bioactivity of iron-matrix composites containing silicate-based bioceramic. *Bioact. Mater.* **2017**, *2*, 10–18. [[CrossRef](#)]
13. Tebib, W.; Alleg, S.; Grenèche, J.-M.; Suñol, J.J. Thermal stability of the nanocrystalline Fe-8P (wt.%) powder produced by ball milling. *Phosphorus Sulfur Silicon Relat. Elem.* **2018**, *193*, 500–506. [[CrossRef](#)]
14. Khodaei, M.; Meratian, M.; Savabi, O. Effect of spacer type and cold compaction pressure on structural and mechanical properties of porous titanium scaffold. *Powder Metall.* **2015**, *58*, 152–160. [[CrossRef](#)]
15. ISO 844-2014; Rigid Cellular Plastics-Determination of Compression Properties. ISO: Geneva, Switzerland, 2014.
16. Aslantürk, Ö.S. In Vitro Cytotoxicity and Cell Viability Assays: Principles, Advantages, and Disadvantages. In *Genotoxicity: A Predictable Risk to Our Actual World*; IntechOpen: London, UK, 2018; Volume 3, pp. 1–19.
17. ASTM-B962-08b; Standard Test Method for Density of Compacted or Sintered Powder Metallurgy (PM) Products Using Archimede's Principle. American Society for Testing and Materials International: West Conshohocken, PA, USA, 2008.
18. Gerhardt, L.-C.; Boccaccini, A.R. Bioactive glass and glass-ceramic scaffolds for bone tissue engineering. *Mater. Sci. Eng. C* **2010**, *3*, 3867–3910. [[CrossRef](#)]
19. Alijani, S.; Anvari, A. Characterization of Implantation's Biomaterials Based on the Patient and Doctor Expectations. *Res. Med. Eng. Sci.* **2018**, *4*, 315–316.
20. Peron, M.; Torgersen, J.; Berto, F. Mg and its alloys for biomedical applications: Exploring corrosion and its interplay with mechanical failure. *Metals* **2017**, *7*, 252. [[CrossRef](#)]
21. Phani, K.K.; Niyogi, S.K. Young's modulus of porous brittle solids. *J. Mater. Sci.* **1987**, *22*, 257–263. [[CrossRef](#)]
22. Černý, M.; Petruš, J.; Chamradová, I. The Influence of Porosity on Mechanical Properties of PUR-Based Composites: Experimentally Derived Mathematical Approach. *Polymers* **2023**, *15*, 1960. [[CrossRef](#)] [[PubMed](#)]
23. Pal, R. Porosity-dependence of Effective Mechanical Properties of Pore-solid Composite Materials. *J. Compos. Mater.* **2005**, *39*, 1147–1158. [[CrossRef](#)]
24. Rubshtein, A.; Makarova, E.; Rinkevich, A.; Medvedeva, D.; Yakovenkova, L.; Vladimirov, A. Elastic properties of a porous titanium–bone tissue composite. *Mater. Sci. Eng. C* **2015**, *52*, 54–60. [[CrossRef](#)]
25. Roberts, A.P.; Garboczi, E.J. Elastic properties of model random three-dimensional open-cell solids. *J. Mech. Phys. Solids* **2002**, *50*, 33–55. [[CrossRef](#)]
26. Watt, J.P.; Davies, G.F.; O'Connell, R.J.O. The Elastic Properties of Composite Materials. *Rev. Geophys.* **1976**, *14*, 541–563. [[CrossRef](#)]
27. Papanikolaou, G.; Pantopoulos, K. Iron metabolism and toxicity. *Toxicol. Appl. Pharmacol.* **2005**, *202*, 199–211. [[CrossRef](#)] [[PubMed](#)]
28. Peuster, M.; Wohlsein, P.; Brüggemann, M.; Ehlerding, M.; Seidler, K.; Fink, C.; Brauer, H.; Fischer, A.; Hausdorf, G. A novel approach to temporary stenting: Degradable cardiovascular stents produced from corrodible metal-results 6–18 months after implantation into New Zealand white rabbits. *Heart* **2001**, *86*, 563–569. [[CrossRef](#)]
29. Yusop, A.H.M.; Alsakkaf, A.; Kadir, M.R.A.; Sukmana, I.; Nur, H. Corrosion of porous Mg and Fe scaffolds: A review of mechanical and biocompatibility responses. *Corros. Eng. Sci. Technol.* **2021**, *56*, 310–326. [[CrossRef](#)]
30. Heiden, M.; Walker, E.; Stanciu, L. Magnesium, iron and zinc alloys, the trifecta of bioresorbable orthopaedic and vas-cular implantation—A review. *J. Biotechnol. Biomater.* **2015**, *5*, 2–9.
31. Sharma, P.; Pandey, P.M. Corrosion rate modelling of biodegradable porous iron scaffold considering the effect of porosity and pore morphology. *Mater. Sci. Eng. C* **2019**, *103*, 109776. [[CrossRef](#)]
32. Han, H.S.; Loffredo, S.; Jun, I.; Edwards, J.; Kim, Y.C.; Seok, H.K.; Witte, F.; Mantovani, D.; Glyn-Jones, S. Current status and outlook on the clinical translation of biodegradable metals. *Mater. Today* **2019**, *23*, 57–71. [[CrossRef](#)]
33. Lee, M.-K.; Lee, H.; Park, C.; Kang, I.-G.; Kim, J.; Kim, H.-E.; Jung, H.-D.; Jang, T.-S. Accelerated biodegradation of iron-based implants via tantalum-implanted surface nanostructures. *Bioact. Mater.* **2022**, *9*, 239–250. [[CrossRef](#)]
34. Wang, H.; Zheng, Y.; Li, Y.; Jiang, C. Improvement of in vitro corrosion and cytocompatibility of biodegradable Fe surface modified by Zn ion implantation. *Appl. Surf. Sci.* **2017**, *403*, 168–176. [[CrossRef](#)]
35. Qi, Y.; Li, X.; He, Y.; Zhang, D.; Ding, J. Mechanism of acceleration of iron corrosion by a polylactide coating. *ACS Appl. Mater. Interfaces* **2019**, *11*, 202–218. [[CrossRef](#)]
36. Lin, W.; Qin, L.; Qi, H.; Zhang, D.; Zhang, G.; Gao, R.; Qiu, H.; Xia, Y.; Cao, P.; Wang, X.; et al. Long-term in vivo corrosion behavior, biocompatibility and bioresorption mechanism of a bioresorbable nitrided iron scaffold. *Acta Biomater.* **2017**, *54*, 454–468. [[CrossRef](#)]
37. Hermawan, H. Updates on the research and development of absorbable metals for biomedical applications. *Prog. Biomater.* **2018**, *7*, 93–110. [[CrossRef](#)]
38. Wang, H.; Zheng, Y.; Liu, J.; Jiang, C.; Li, Y. In vitro corrosion properties and cytocompatibility of Fe-Ga alloys as potential biodegradable metallic materials. *Mater. Sci. Eng. C* **2017**, *71*, 60–66. [[CrossRef](#)]
39. Chen, Y.; Zhang, W.; Maitz, M.F.; Chen, M.; Zhang, H.; Mao, J.; Zhao, Y.; Huang, N.; Wan, G. Comparative corrosion behavior of Zn with Fe and Mg in the course of immersion degradation in phosphate buffered saline. *Corros. Sci.* **2016**, *111*, 541–555. [[CrossRef](#)]

40. Huang, T.; Cheng, J.; Bian, D.; Zheng, Y. Fe–Au and Fe–Ag composites as candidates for biodegradable stent materials. *J. Biomed. Mater. Res. Part B Appl. Biomater.* **2016**, *104*, 225–240. [[CrossRef](#)]
41. Moravej, M.; Prima, F.; Fiset, M.; Mantovani, D. Electroformed iron as new biomaterial for degradable stents: Development process and structure–properties relationship. *Acta Biomater.* **2010**, *6*, 1726–1735. [[CrossRef](#)]
42. Cheng, J.; Huang, T.; Zheng, Y.F. Relatively uniform and accelerated degradation of pure iron coated with micro-patterned Au disc arrays. *Mater. Sci. Eng. C* **2015**, *48*, 679–687. [[CrossRef](#)]
43. Huang, T.; Zheng, Y. Uniform and accelerated degradation of pure iron patterned by Pt disc arrays. *Sci. Rep.* **2016**, *6*, 23627. [[CrossRef](#)]
44. Obayi, C.S.; Tolouei, R.; Paternoster, C.; Turgeon, S.; Okorie, B.A.; Obikwelu, D.O.; Cassar, G.; Buhagiar, J.; Mantovani, D. Influence of cross-rolling on the micro-texture and biodegradation of pure iron as biodegradable material for medical implants. *Acta Biomater.* **2015**, *17*, 68–77. [[CrossRef](#)]
45. Li, Y.; Jahr, H.; Lietaert, K.; Pavanram, P.; Yilmaz, A.; Fockaert, L.I.; Leeftang, M.A.; Pouran, B.; Gonzalez-Garcia, Y.; Weinans, H.; et al. Additively manufactured biodegradable porous iron. *Acta Biomater.* **2018**, *77*, 380–393. [[CrossRef](#)]
46. Zadpoor, A.A. Mechanical performance of additively manufactured meta-biomaterials. *Acta Biomater.* **2019**, *85*, 41–59. [[CrossRef](#)]
47. Qin, Y.; Wen, P.; Guo, H.; Xia, D.; Zheng, Y.; Jauer, L.; Poprawe, R.; Voshage, M.; Schleifenbaum, J.H. Additive manufacturing of biodegradable metals: Current research status and future perspectives. *Acta Biomater.* **2019**, *98*, 3–22. [[CrossRef](#)]
48. Fu, J.; Su, Y.; Qin, Y.-X.; Zheng, Y.; Wang, Y.; Zhu, D. Evolution of metallic cardiovascular stent materials: A comparative study among stainless steel, magnesium and zinc. *Biomaterials* **2020**, *230*, 119641. [[CrossRef](#)]
49. Daud, N.M.; Sing, N.B.; Yusop, A.H.; Majid, F.A.A.; Hermawan, H. Degradation and in vitro cell-material interaction studies on hydroxyapatite-coated biodegradable porous iron for hard tissue scaffolds. *J. Orthop. Transl.* **2014**, *2*, 177–184.

Disclaimer/Publisher’s Note: The statements, opinions and data contained in all publications are solely those of the individual author(s) and contributor(s) and not of MDPI and/or the editor(s). MDPI and/or the editor(s) disclaim responsibility for any injury to people or property resulting from any ideas, methods, instructions or products referred to in the content.