Reconditioning by Welding of Prosthesis Obtained through Additive Manufacturing

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Abstract: Biocompatible titanium alloys are increasingly being used to make custom medical implants using additive manufacturing processes. This paper considered the welding reconditioning of a titanium-alloy customized additive manufactured hip implant with several manufacturing defects. The personalized implants are made starting from a Computer-Aided Design (CAD) model as a direct result from the medical imaging investigations of the areas of interest. Then the customized implant is fabricated using an additive manufacturing process (in this case Powder Bed Fusion—Direct Metal Laser Sintering—DMLS). The analysis of the chemical composition values as well as the values of the mechanical properties of the samples obtained via DMLS additive manufacturing process, revealed that such a manufacturing process can be successfully used to make customized surgical implants. The mechanical properties values of the DMLS samples are approximately equal to those specified by the manufacturer of the titanium powder used for sintering. On average, the tensile strength was found to be 24.75% higher, while yield strength 22.7% higher than the values provided in the standard for surgical implants applications. In case the additive manufacturing process produces products with defects one might want to try and recover the implant due to costs and time constraints. The Tungsten Inert Gas (TIG) welding reconditioning process with ERTi-5 Ti64 rod for welding titanium alloys with a content of 6% aluminum and 4% vanadium filler material was used to restore the geometric characteristics as well as the functional properties of a custom hip medical prosthesis. After welding depositing successive layers of materials, the surfaces of the prosthesis were machined to restore the functional properties according to the characteristics of the original 3D model. A 3D scan was used to compare the geometrical characteristics between the original part and reconditioned one. Deviations were less than 1 mm and were acceptable from the medical point of view.

Keywords: titanium alloy; powder bed fusion; hip implants; TIG welding reconditioning

1. Introduction

Medical implants and prostheses that replace different parts of the human body are made of biocompatible materials, which have mechanical and physical-chemical properties specific to these applications [1–4]. The materials used in medical implants manufacturing are, at best, inert to the bone tissue having the possibility to become bioactive by controlling the morphology and chemical composition at surface. The elements that must be considered when choosing the material used for medical implants such as compatibility, the material nature and degree of alloying, process and manufacturing method, mechanical properties, processing conditions significantly influence the interaction between the material and bone.
The long-term stability of the implant depends largely on all of the aspects presented above but also on its ability to integrate into the adjacent bone tissue [5].

Considering the possible problems caused by the incompatibility between the materials used to make the prosthesis on the bone tissue, scientists collaborate to create new biocompatible materials with a much higher acceptance rate [6,7].

Currently, a wide variety of biocompatible materials are used to make medical implants, such as metals and alloys (stainless steel, Ti-Al-V, Co-Cr-Mo) ceramics and glasses (alumina, zirconia), polymers (polyethylene, polyamides), composites (PMMA-glass fillers) [8,9].

In present, most medical prosthesis are made of titanium alloys, alloys that offer high mechanical properties (tensile strength, yield strength), low Young’s module, low specific weight (half the specific weight of stainless steel). In addition, such materials are not toxic (in comparison, stainless steel can cause several allergies), they are not ferromagnetic (the possibility of performing magnetic resonance imaging investigations) and have a good corrosion resistance in specific working environments [10,11].

The Ti6Al4V titanium alloy, is the most common titanium alloy used for making medical implants, being an \( \alpha + \beta \) type of alloy initially used in the aerospace industry [12–15]. As in the case of stainless steels, the corrosion resistance of titanium alloys is given by the formation of an oxide layer on the surface of the material, in this case being the TiO\(_2\) layer [16]. The metallic materials used in the manufacture of medical implants are inert to human bone tissue [17,18]. To obtain customized implants having shapes and sizes specific to their use in case of surgery in areas with continuous growing tumors has led to the need of using new manufacturing methods. The classical methods of obtaining implants and medical instruments (forging, stamping) do not offer a high dimensional accuracy of the product [19]. To meet the dimensional and quality requirements for customized implants, the new trend is the use of additive manufacturing processes (3D printing), using the information obtained from the imaging examinations to which the patient was subjected [20,21]. Popov et al. [22] present the possibility of healthcare digitalization in the Industry 4.0 revolution using 3D printing which allows on-site printing of freeform shapes, which are potentially useful to develop custom-sized implants or prostheses.

Direct metal laser sintering (DMLS) is an additive manufacturing process used for printing metal products, which is based on using the energy of a laser beam to melt layers of metal powder (sintering) which, following solidification, finally forms the 3D model of the desired product. The size of the metal powder particles is between 20 and 40 \( \mu m \). The size of the particles, the surfaces of the products, as well as their geometric configuration influence the final printing resolution [23–27].

In certain situations, when implants are made by classical technological processes, there are disturbing factors that appear and influence the result of the process. In such cases, the resulting product has a series of dimensional or shape (geometric) imperfections that may lead to the decision to classify the resulting product as scrap. In these situations, the prosthesis material must be remelted and subjected to a new processing process, which leads to increased costs related to obtaining implants in the final form [28,29]. In the case of additive manufacturing processes, the reuse of scrap is no longer possible, so the financial, time or resources loss is significant.

In the case of recoverable scrap, there is the possibility of using welding reconditioning to restore the geometric configuration and the functional properties of the implants.

Through the advantages offered by the welding processes, it is possible to reduce the costs related to the production of implants, but also those related to the long-term use of surgical instruments or cosmetic implants. The costs related to welding reconditioning of instruments and medical implants, depending on their complexity and the type of defect, are between 15 and 50% of the costs of new products.

In the literature, there is a limited number of studies that refer to welding reconditioning possibilities of different types of products, made of steel, cast iron [30–38] or non-ferrous materials [39]. In terms of welding reconditioning products made of titanium

Following the desktop research, it was found that there is no research conducted on the possibility of modifying the characteristics of the products obtained with the help of AM processes. In some cases, and especially in the case of medical implants, the rapid evolution of malignant bone tumors can lead to changes in the bone area that must be excised and replaced. In such cases, considering the rapid evolution of the tumor, the time from the analysis of the imaging results to the design and realization of the implant with the help of an AM procedure, makes it necessary to modify the constructive form of the implant, by adding additional elements to compensate the excised bone area or of some additional elements for the attachment of the implant on the healthy bone structure.

The purpose of this paper is to analyze the possibilities of restoring the geometric characteristics by welding of implants made by additive manufacturing processes that would otherwise be unusable and unrecoverable. In this case, the use of the DMLS additive manufacturing process of a custom hip implant, resulted in a geometric configuration that did not meet the requirements of the fabrication specifications. To restore the geometric characteristics of the implant, the tungsten inert gas (TIG) welding deposition process was used.

The novelty of the proposed research consists in the restoration and/or modification of the geometric and functional characteristics of an AM made implant with manufacturing errors with the help of welding reconditioning processes by. Welding reconditioning is presented, in some cases, as a fast and inexpensive alternative for remanufacturing the implant according to the new functional requirements.

2. Materials and Methods

2.1. Materials

The implant demonstrator has been additively manufactured using Ti-6Al-4V powder supplied by EOS Gmbh (Krailling, Germany). For the TIG welding deposition the ERTi-5 Ti64 rod filler material was used.

The ERTi-5 Ti64 rod is used for welding titanium alloys with a content of 6% aluminum and 4% vanadium, providing high fatigue strength, toughness, and ductility of the weld bead material. It has good weldability and can be heat treated to reach a higher strength or toughness.

The ERTi-5 Ti64 rod is currently used in the aircraft and motor-sports competition applications for the manufacture of airframes and chassis structures, turbine engine parts, exhaust systems and ducting, discs, wheels, spacer rings, but also in medical industry [47].

The chemical composition and mechanical properties of the materials used, according to the manufacturer’s quality certificates, are presented in Tables 1 and 2.

Table 1. Chemical composition of the used materials in wt%.

<table>
<thead>
<tr>
<th>Materials</th>
<th>Ti</th>
<th>Al</th>
<th>V</th>
</tr>
</thead>
<tbody>
<tr>
<td>EOS Ti64 powder</td>
<td>Bal.</td>
<td>5.5–6.75</td>
<td>3.5–4.5</td>
</tr>
<tr>
<td>ERTi-5 Ti64 rod</td>
<td>Bal.</td>
<td>5.5–6.75</td>
<td>3.5–4.5</td>
</tr>
</tbody>
</table>

Table 2. Mechanical properties of the used materials.

<table>
<thead>
<tr>
<th>Materials</th>
<th>Tensile Strength (MPa)</th>
<th>Yield Strength (MPa)</th>
<th>Elongation (%)</th>
</tr>
</thead>
<tbody>
<tr>
<td>EOS Ti64 powder</td>
<td>1150</td>
<td>1030</td>
<td>11</td>
</tr>
<tr>
<td>ERTi-5 Ti64 rod</td>
<td>895</td>
<td>828</td>
<td>12</td>
</tr>
</tbody>
</table>
2.2. Mechanical Measurements and Microstructural Characterization Methods

Considering that the parts obtained with the help of the DMLS additive manufacturing process are made by a successive deposition of the sintered layers of metallic powder, the resulting parts may not correspond in terms of mechanical properties. To analyze the values of the mechanical properties of the resulting components, four samples (T1, T2, T3, T4) obtained by DMLS with the help of EOSINT M270 equipment (EOS GmbH, Krailling, Germany) were subjected to tensile testing. The tensile test was performed, using H10KT test machine (TINIUS OLSEN, Salfords, England) with a 2 mm/s (23 N/mm² s) working speed value. The Rockwell, HRC test method, 120° cone angle, 1471N compressive force (206× hardness tester model from Affri System, Varese, Italy) was used to measure the hardness of the sample material obtained by additive manufacturing. The shape of the samples, in accordance with ASTM E8/E8M:2021 [48], is presented in Figure 1. Microscopic analysis of the samples was performed using an optical microscope (OLYMPUS GX51, Tokyo, Japan) as well as an electron microscope (SEM INSPECT S, FEI, Eindhoven, The Netherlands) in high-vacuum working mode for imaging and microanalysis of conductive samples and/or conventional prepared samples (coated) and low-vacuum working mode for imaging and microanalysis of unprepared samples.

![Figure 1](image.jpg)

**Figure 1.** The shape of the samples used to perform the tensile test.

2.3. Methods

2.3.1. Additive Manufacturing

Additive manufacturing processes are a rapid prototyping technology that provides the fastest way from product idea to market launch. Innovative companies around the world use this technology for very fast, flexible and cost-effective parts production, directly from electronic format. The DMLS additive manufacturing process was used to make the custom hip implant. For this, the 3D CAD model is divided into layers using EOS RP Tools software (version 5.2, Cambridge, UK). This conversion is necessary as the additive manufacturing equipment can achieve the desired prototype only by layer-by-layer deposition. Furthermore, these specialized software packages ensure that the layers do not form in the air without having the support of the previous layers or special structures (called supports). Therefore, supports are generated where considered necessary. In the DMLS case, the model and the supports are automatically divided by the software into layers (bottom to top) with a thickness of 0.02 or 0.03 mm depending on the type of material used. The 0.03 mm layer thickness is used only for titanium and titanium alloy Ti6Al4V, for all other materials the 0.02 mm thickness is valid. The 3D model of the implant presented in Figure 2a, can be made based on the information obtained from the imaging investigations of the person for whom it is desired to customize the implant. The equipment used to make the implant was a rapid prototyping equipment by laser sintering, EOSINT M270 type (EOS GmbH, Krailling, Germany), which has the possibility of processing titanium powders, by using argon as a protecting atmosphere. The working parameters used for the manufacture of the hip implant using the DMLS process are presented in Table 3. Following the additive manufacturing process, the implant presented in Figure 2b was obtained.
During the additive manufacturing process of the custom implant, an error occurred (Figure 3) which led to improper geometric characteristics of the final product. From the analysis of the model presented in Figure 3 one can observe that the clamping ear as well as the fixing holes of the implant do not meet the initial specifications of the 3D model. This error led to the impossibility of using the implant for surgery and classify it as scrap.

To restore the geometric characteristics of the implant, a TIG welding deposition process was proposed [49]. The removal of the inappropriate layer was carried out by cutting machining. On the resulted surface, successive layers were deposited to reach the shape required for further cutting machining and restoring the implant to the nominal size of the initial 3D drawing. The welding deposition parameters are presented in Table 4.

### Table 3. Working parameters of the EOS M270.

<table>
<thead>
<tr>
<th>Parameters</th>
<th>Hatching</th>
<th>Contouring</th>
</tr>
</thead>
<tbody>
<tr>
<td>Laser power (W)</td>
<td>170</td>
<td>150</td>
</tr>
<tr>
<td>Scan speed (mm/s)</td>
<td></td>
<td>1250</td>
</tr>
<tr>
<td>Hatching spacing (mm)</td>
<td>0.10</td>
<td>NA</td>
</tr>
<tr>
<td>Stripe width (mm)</td>
<td>5.0</td>
<td>MA</td>
</tr>
<tr>
<td>Beam offset (μm)</td>
<td>0.015</td>
<td>0.020</td>
</tr>
<tr>
<td>Scanning pattern</td>
<td>Rotated</td>
<td></td>
</tr>
</tbody>
</table>

### Table 4. The welding parameters used in the TIG welding process.

<table>
<thead>
<tr>
<th>Parameters</th>
<th>Value</th>
</tr>
</thead>
<tbody>
<tr>
<td>Welding current (A)</td>
<td>40–45</td>
</tr>
<tr>
<td>Arc voltage (V)</td>
<td>12–14</td>
</tr>
<tr>
<td>Filler material diameter (mm)</td>
<td>2.00</td>
</tr>
<tr>
<td>Gas flow (L/min)</td>
<td>10</td>
</tr>
<tr>
<td>Travel speed (cm/min)</td>
<td>12</td>
</tr>
<tr>
<td>Type of gas protection</td>
<td>Argon</td>
</tr>
</tbody>
</table>
To restore the geometric shape of the implant, six layers of welding were deposited. The first layer was applied directly to the surface resulted from cutting machining to melt and homogenize all of the micro asperities of the inappropriate surface. The next four layers were deposited on the side surfaces of the clamping ear, and layer six was deposited on the upper surface.

The decision to restore the functional properties by welding reconditioning processes must consider the technological possibilities of production and includes a series of steps specific to the manufacturing processes. The welding reconditioning flowchart for customized implants is presented in Figure 4.

3. Results and Discussion

3.1. Mechanical Properties

The values of the hardnesses as well as of the mechanical properties of the Ti6Al4V samples obtained by DMLS additive manufacturing process are presented in Table 5. Analyzing the values presented in Table 5 one can observe that the samples obtained by DMLS show values within the limits prescribed by the manufacturer of EOS Ti64 powder. The modulus of elasticity varies between 99.41 GPa (T1) and 107.86 GPa (T4), lower values compared to the values of the Ti6Al4V sheets modulus of elasticity obtained by cold-rolling or hot-rolling. The values of the mechanical properties are slightly higher than those obtained by the classical manufacturing processes (tensile strength $R_m = 892$ MPa and $R_p0.2 = Yield strength 828$ MPa [47]). On average, the tensile strength was found to be 24.75% higher, while yield strength 22.7% higher than the values provided in the standard for surgical implants applications [49–54]. The results validated the EOS datasheet specifications, also presented in Table 5.
Figure 4. Welding reconditioning flowchart for customized implants.
Table 5. Mechanical properties of the DMLS Ti6Al4V sample.

<table>
<thead>
<tr>
<th>Properties</th>
<th>T1</th>
<th>T2</th>
<th>T3</th>
<th>T4</th>
<th>EOS</th>
</tr>
</thead>
<tbody>
<tr>
<td>Tensile strength $R_m$ (MPa)</td>
<td>1120.33</td>
<td>1132.69</td>
<td>1091.25</td>
<td>1106.95</td>
<td>1150 ± 60</td>
</tr>
<tr>
<td>Tensile strength $R_m$ difference (%)</td>
<td>25.60</td>
<td>26.98</td>
<td>22.34</td>
<td>24.10</td>
<td>-</td>
</tr>
<tr>
<td>Yield strength $R_p0.2$ (MPa)</td>
<td>1040.99</td>
<td>1018.14</td>
<td>998.50</td>
<td>1006.46</td>
<td>1030 ± 70</td>
</tr>
<tr>
<td>Yield strength $R_p0.2$ difference (%)</td>
<td>25.72</td>
<td>22.96</td>
<td>20.59</td>
<td>21.55</td>
<td>-</td>
</tr>
<tr>
<td>Elongation A (%)</td>
<td>8.98</td>
<td>8.57</td>
<td>8.34</td>
<td>9.01</td>
<td>10 ± 2</td>
</tr>
<tr>
<td>Young Modulus E (GPa)</td>
<td>99.41</td>
<td>101.72</td>
<td>102.66</td>
<td>107.86</td>
<td>110 ± 15</td>
</tr>
<tr>
<td>Hardness (HRC)</td>
<td>31</td>
<td>36</td>
<td>38</td>
<td>34</td>
<td>31–35</td>
</tr>
</tbody>
</table>

3.2. Composition and Microstructure of Ti64 DMLS Samples

Table 6 presents the average chemical composition of the Ti64 powder samples, obtained by DMLS additive manufacturing process, as measured by energy dispersive x-ray (EDX) spectroscopy analysis. Following the analysis, one can notice that the samples are homogeneous, without imperfections and the values of the chemical composition, measured at different points, are approximately equal throughout their mass.

Table 6. Chemical composition of the laser-sintered powder and weld deposit material wt%.

<table>
<thead>
<tr>
<th>Materials</th>
<th>Ti</th>
<th>Al</th>
<th>V</th>
</tr>
</thead>
<tbody>
<tr>
<td>Ti64 implant</td>
<td>Bal.</td>
<td>6.35</td>
<td>4.23</td>
</tr>
<tr>
<td>Weld deposit</td>
<td>Bal.</td>
<td>5.95</td>
<td>3.99</td>
</tr>
</tbody>
</table>

Figure 5 shows the microstructure of the Ti6Al4V samples obtained by the DMLS additive manufacturing process. The microscopical analysis of the images presented in Figure 5 shows a lamellar structure of type ($\alpha + \beta$) combined with a martensitic acicular structure of type $\alpha'$, which could explain the slight increase of the hardness values compared to the specifications of the metal powder manufacturer. The formation of the martensitic structure is caused by the principle of the DMLS process which consists in heating to the melting temperature followed by a rapid cooling process. The microstructure obtained, as well as the values of the mechanical and chemical composition characteristics of the analyzed samples obtained by DMLS, are similar to those of the products fabricated by classical manufacturing processes. Figure 6 presents the results of the EDS analysis for the implant obtained by additive manufacturing process (Figure 6a) and by welding deposition (Figure 6b) made to restore the geometric characteristics of the implant.

Following the analysis of the values of mechanical and chemical composition properties obtained, one can observe that they are similar to those of products obtained by classical manufacturing processes, which may lead to the decision of using additive manufacturing methods to obtain customized medical implants. The implant resulted from the welding deposition process, using the technological parameters presented in Table 4, is presented in Figure 7. The analysis of the sample presented in Figure 3 shows that the width of the deposition area is relatively small ($3.1 ± 0.1$ mm wide). After processing the affected area, by burnishing, the implant clamping edge was reconstructed by welding deposition. Figure 7 shows the layer-by-layer welding deposition process until reaching the dimensions that allow further processing by cutting machining ($6.2 ± 0.1$ mm average width) and restoring the area to the geometric configuration and tolerances prescribed in the fabrication drawing of the 3D model.
After cutting machining on a CNC machine (Figure 8), the surfaces of the “recovered” implant must be brushed to reach the rugosity imposed by its functional role. In the end it is necessary to process the tapped holes to the dimensions of the special fastening screws on the bone structure. The custom implant resulted from post-processing (cutting machining) is presented in Figure 9.

One area of interest is the actual interface between the metallic layer deposited by welding and the main body of the implant (the connection area between the AM part and the welded part). This area was also analyzed under a microscope, as explained in previous section. As a result, Figure 10 shows the structural changes in the reconditioning area. Considering the value of the temperature reached during the welding process, but also the rapid cooling of the material, in the fusion zone (FZ), but also in the thermally influenced zone (HAZ), one can observe the occurrence of the \(\alpha\) acicular phase and the recrystallized \(\beta\) phase. In the area of the cord material, a higher percentage of martensite (\(\alpha'\) phase) can be observed, as well as a lamellar \(\alpha\) structure.
Figure 6. EDS analysis of sintered samples of Ti6Al4V alloy fabricated using DMLS process: (a) sintered samples; (b) weld deposit zone.
Figure 7. Custom implant subjected to the welding deposition process to restore the geometric characteristics prescribed by the 3D model. (a) front view; (b) back view.

After cutting machining on a CNC machine (Figure 8), the surfaces of the “recovered” implant must be brushed to reach the rugosity imposed by its functional role. In the end it is necessary to process the tapped holes to the dimensions of the special fastening screws on the bone structure. The custom implant resulted from post-processing (cutting machining) is presented in Figure 9.

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Figure 8. Cutting machining for the dimensional restoration of the personalized implant: (a) fixing the implant in the device to mill the edges of the implant; (b) drilling holes.
3.3. Geometrical Analysis of the Corrected Holes

To verify the accuracy of the shape and the geometric dimensions of the reconditioned implant, the 3d model of the implant was reverse engineered by using a Hexagon (ROMER) Absolute arm with 3d scanner and touch equipment. A set of surfaces (2394422) were generated over the cloud points by using the Designer 2022.0 R2 software package with reverse capabilities (Figure 11). After a simple re-mesh (automatically surface) a fully 3d triangle mesh surfaces were obtained. This model has high accuracy of the real scanned implant. By generating parametric surfaces with high accuracy of the points cloud, precise measurements are possible to be made. The scanned holes were reversed to cylindrical surfaces and measured by using the software capabilities.
By comparative analysis with the original CAD model some errors were identified as deviations between components and holes shape. A solution was discussed together with an orthopedic surgeon specialist. The measured deviations were accepted (under 1 mm deviations were measured as explained above), and for the holes the next thread bigger diameter was accepted. The technical solution was obtained in two steps: firstly, enlargement of the hole diameter by drilling up to 6.2 mm (shape correction to cylindrical face) and then the generation of the thread by taping to M7. Then a NC program for machining on the Fanuc equipment mentioned above was designed and implemented.

After completing the process of restoring the construction of the medical implant and processing it, using the TIG welding reconditioning process and the post-processing through cutting machining, the samples will be subjected to specific cleaning, decontamination and sterilization treatments specified in the medical quality standards regarding handling and use of medical instruments and equipment.

From the point of view of TIG welding reconditioning possibilities, it is found that there are certain limitations of using the process caused by large dimensional deviations of the obtained surfaces, reduced accessibility to interior areas and the possibility of reconditioning products with complex surfaces. Another possible issue is related to ensuring inert protection in the reconditioning area, protection that influences the weldability of the material but also the properties of the reconditioned product.

4. Conclusions

The present study considered welding reconditioning of a titanium-alloy customized additive manufactured implant with several manufacturing defects. The personalized implants are made starting from a CAD model as a direct result from the imaging investigations of the areas of interest. Then the implant is fabricated using Additive Manufacturing process (in this case, DMLS). The analysis of the chemical composition values as well as the values of the mechanical properties of the samples obtained via DMLS additive manufacturing process, showed that such a manufacturing process can be successfully used to make customized surgical implants. The mechanical properties values of the DMLS samples are approximately equal to those specified by the manufacturer of the titanium powder used for sintering. On average the tensile strength was found to be 24.75% higher, while yield strength 22.7% higher than the values provided in the standard for surgical implants applications [49–54].

If at the end of the AM production process results an inappropriate implant (for example it does not correspond from a functional, geometric, or dimensional point of view),
view), it is considered scrap. The implant cannot be melted and reused, so this study aimed to recover such implants by restoring their geometric and functional characteristics through specific welding reconditioning processes. Several stages are proposed for this reconditioning process. The process starts with depositing successive layers by welding, followed by post-processing using a CNC cutting machining, brushing, or polishing the surfaces to achieve the roughness required by the medical field, processing tapped holes for special screws on the bone structure, followed by specific cleaning decontamination and sterilization treatments. These modifications on the implants can be made to correct certain dimensional and shape deviations resulting from the additive manufacturing process, as well as to adapt according to the continuous evolution of the tumors in the implant area. The mechanical properties of obtained implant, with regards to the welded area, are similar to the original AM obtained implant and hence usable for its original purpose.

There are several limitations to the use of welding technologies for reconditioning of defective parts. If large dimensional deviations occurred, it is difficult, if not impossible, to recondition the part mainly since the large area required for adhesion between the filler material and the main body of the implant creates stress points and the occurrence of large areas that will suffer from heating and rapid cooling processes during welding. Further investigations are underway to solve this aspect by using additive manufactured micro and mini scaffolds to help the welded area maintain its original intended mechanical properties and to increase the adhesion of the filler material to the main body of the implant.

For this study we could not perform destructive testing on the implant, especially in the welded area. Therefore, further work is required to determine the changes in mechanical properties of the deposited layers of the welded materials (as this is affected by heat that can influence the interface between the AM part and the welded-on areas).

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**Informed Consent Statement:** No applicable.

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

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