Additive Manufacturing of a Compliant Multimaterial Heart Model

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Abstract. Additive manufacturing technologies are increasingly taking place in the medical field, enabling the creation of graspable patient-specific anatomical models. Because of their potentiality in improving the understanding of complex anatomies and their shown effectiveness for residents' training, devices testing and planning of innovative surgical interventions, 3D printed models have been incorporated also into cardiac surgery and interventional cardiology. To offer valid and reliable support, however, these printed models are often required to be flexible, with an adequate mechanical response, especially when they aim at replicating soft tissues. The goal of this paper is to provide a high-quality and robust template of a patient-specific whole heart model, obtained starting from a Computed Tomography dataset and exploiting a material jetting printer. Due to the significant shape complexity and the variability in compliance featuring the human heart, the selection of the materials have been diversified, taking into account different model wall thicknesses. Thanks to the capability of the material jetting technology, the 3D model of the heart has been printed with two different material assignments, designed to get highly realistic feedback and reduce the gap between the real heart and the printed ones. Eventually, an accuracy evaluation of the printed model has been performed, by means of a laser 3D scanner. Some further considerations about time and costs required to produce the model are part of the paper, together with a discussion about potential areas of improvement, from materials characterization to the need of speeding up and automating the segmentation procedure.

Keywords: Segmentation, Additive Manufacturing, Material Jetting, Patient-specific Anatomy, Heart
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1 INTRODUCTION

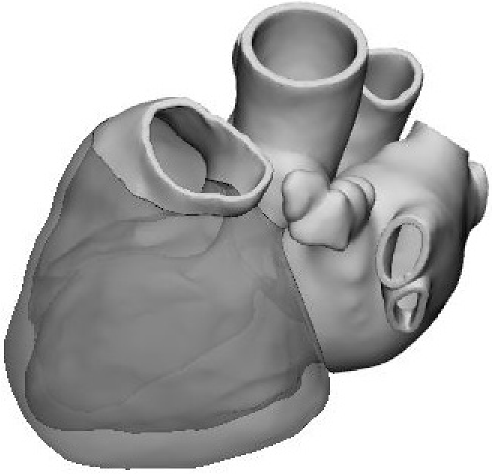
Three-dimensional printing (3DP) is a process of additive manufacturing by which a 3D object is built layer-by-layer. Over the last 10 years, 3DP has become an affordable means for producing bespoke solutions quickly [15], especially when generation of complex 3D geometries is adequately supported by innovative and interactive design tools [4]. In the medical field, this technology has gained popularity for creating customized prosthesis [5], implants, fixtures, surgical tools, as well as reproducing patient-specific 3D anatomical models [9]. The starting point, in this case, is given by a set of 2D images of the interior of the body acquired by medical imaging techniques, such as computer tomography (CT) or magnetic resonance (MR). From them, through a process known as segmentation, anatomical structures of interest are isolated and then exported as faceted geometries, almost ready to be printed.

Even if the technology is not fully mature, there are proved potentialities in exploiting 3D printed models in the cardiovascular field. The key plus is the possibility of customizing geometries, to obtain patient-specific models that can be exploited for personalized cares, so moving from an "average" patient to an "individualized" patient. Physical prints draw their competitive advantage in the haptic perception they guarantee, being therefore complementary to the visual assessment provided by 2D imaging or other 3D visualization techniques, such as virtual reality and augmented reality [15].

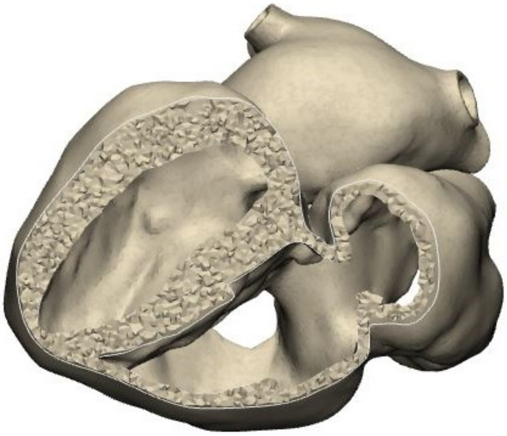
This potentiality turned out to be very useful also in the cardiovascular field: through 3D printing, for example, accurate educational tools able to illustrate complex cardiovascular anatomy and pathology can be created [10]. Compared to 2D images, 3D renderings guarantee a better understanding of the human body, also of fine but fundamental anatomical details [13]. This aspect also turned out to be helpful for improving communication between surgeons and patients [8]. Above all, this technology can be used to create and analyze 3D model before starting actual surgery on the patient [8]. Printed models guarantee the possibility to gather insight into the cardiovascular anatomy, to complement the imaging data with regards to the position and size of the potential heart defects, also allowing to better appreciate relative positions among districts of interest [15]. This may be very important, if we think about the great anatomical complexity, especially in the presence of congenital heart diseases (CHD). In this scenario, the decision-making process, when considered complex and non-routine, can benefit from the availability of physical 3D models, allowing for an effective replication of the surgical procedure, such as dissections, suturing or devices placement, thus reducing operative risks [16]. The employment of adequately distensible resins surely helps to increase the realism and the reliability of these simulations. Moreover, patient-specific implants and custom-made devices could be designed and tested, opening new clinical possibilities [16]. This is the case for example of heart valves, for which a careful choice of the device size is fundamental to avoid complications (e.g. paravalvular leaks). With this regard, recently developing catheter-based procedure can surely benefit from the introduction of these compliant printed models, fundamentals in surgeons' training for devices implant simulations [6].

In this context, the aim of this work is to realize a compliant patient-specific heart model, a sort of high quality template, not found in the literature, endowed with mechanical properties that could get close to those ones of the biological tissue. After a careful segmentation of the anatomy, that includes wall thickness variability, a differentiated materials assignment will be conducted, distinguishing between thin atrial and thicker ventricle walls. This could be considered the underlying idea for potential applications in a clinical environment, as just now discussed: in other words, the paper basically aims at providing medical investigators with useful information on how a patient-specific multi-material heart model should be built from CT or MRI datasets.

Moreover, to add concreteness to the work, we will evaluate the dimensional accuracy of our print. On the basis of a reverse engineering approach, the model will be re-acquired with a laser scanner; this acquisition will be then compared with the original STL files used in the print. In doing so, we will have also an idea of what is the level of agreement between our compliant printed result and the digital model.



(a)



(b)

Figure 1: **(a)** A step in the workflow to obtain the digital heart model. The blood pool is colored in light gray, while the myocardium is gray and semi-transparent. **(b)** Wall thickness variability in the 3D model of the heart.

2 MATERIALS AND METHODS

The starting point for this manufacturing process is segmentation, by which the anatomical structures of interest can be isolated, generating a 3D model. This step has been conducted exploiting Mimics software (Materialise, Belgium, version 21), starting from a stack of high-resolution Computed Tomography (CT) scans. It consisted of 393 slices in the axial plane, with a resolution of 512x512 pixels. The dimension of each pixel was 0.35 mm, while the distance between two subsequent slices was 0.75 mm.

The process was performed manually, to obtain the "blood pool" volume: thresholding algorithm was firstly applied, followed by meticulous fixing operations on the obtained mask, in order to remove over-segmented details in between heart chambers and mitigate the influence of *trabeculae carnee* presence on the surface of the ventricles. Moreover, some shells had to be removed and the mesh fixed. All these operations were performed through 3-matic software (Materialise, Belgium, version 13), in order to be sure of reaching a watertight mesh.

The final model consisted of the two ventricles, the two atria and the first tract of the aorta, while other vessels have been neglected for simplicity. At this point, the blood pool - i.e. the interior volume of the heart filled with blood - is obtained. A strategy to derive the surrounding tissues is to create a shell, starting from the segmented model. Despite the thickness of the shell can be specified as variable, most of the work in the literature assign constant values, thus not reproducing the realistic thickness variation (see for example [18]): this would affect the mechanical properties of the final print, resulting in an unrealistic uniform behavior. To solve the problem, the following strategy has been played out: the myocardium was segmented aside, from the same stack of images, and only in a second step, through boolean operators, it was properly merged with the shell derived from the blood pool. A graphical illustration of a step of the process can be found in Fig. 1a, where the blood pool model has a light gray color, while the myocardium is gray and semi-transparent.

In this way, the wall thickness map has been replicated as faithfully as possible. For example, the atria have an average thickness of about 2.5 mm, as it can be derived from the literature for an adult healthy subject (see, for example [17]). Globally, the wall thickness ranges from 1 mm to more than 10 mm, in correspondence

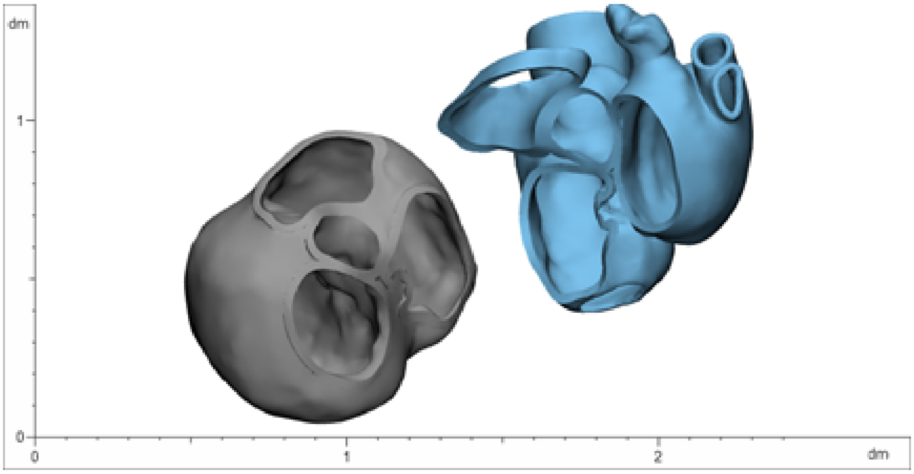


Figure 2: Splitting of the STL model into two parts, to which different printing materials are assigned.

with the left ventricle [7]. A section view, in which the wall thickness variability can be better appreciated, is reported in Fig. 1b. The final STL file consists of 103076 triangles, with a surface area about 829 cm^2 and a volume of about 122 cm^3.

3 3D PRINTING

A Stratasys Connex Object260 (Stratasys, USA) printer was employed. It implements the material jetting technology, allowing to reach a resolution of the building layer up to 16 μm. Differently from other printing technologies, the material jetting can easily create multi-material objects, also realizing polymeric blends whose mechanical properties and compliance can be finely tuned. Moreover, with respect to the other 3DP technology, prints are relatively fast, with the printable volume slightly above average (255x252x200 mm). The employed software for preparing the job was GrabCAD Print (Stratasys, USA, version 1.49). The position and orientation of the model on the printing platform were properly set, to minimize the printing time, followed by material assignment.

The *VeroWhitePlus* and *Agilus30Clear* are the materials selected for this work. The former is a rigid and opaque resin, with good mechanical properties, while the latter is rubber-like, characterized by elongation at break equal to 220-240%, as declared by the producer. As support, SUP706 has been selected: it dissolves in a solution of caustic soda and sodium metasilicate. Preliminary tests were conducted exploiting a single material: a blend between *VeroWhite* and *Agilus* was adopted, varying the relative proportion and so the global shore hardness: in this way, however, we had always to keep a balance between the resulting properties of thin atrial wall and the thicker ventricle one.

The results was not satisfactory, hence the model has been divided into two parts: to guarantee the possibility of a differentiated material assignment, the STL model has been split into two parts, exploiting Boolean functions applied to the model. The process was always performed in 3-matic environment and the result is shown in Fig. 2. In this way, the materials assignment was easier: for the thin atria walls a blend of 60% VeroWhite and 40% Agilus, as suggested by Stratasys to guarantee a final material shore hardness equal to 50 A, was chosen, while for the thicker ventricles pure Agilus, whose shore A hardness is 30, was thought to be the best choice. The mechanical properties of the two materials, as declared by the supplier, are summarized in the Tab. 1.

The overall printing time was about 14 hours. At the end, the model was soaked in soda solution for about

	Agilus30	***Blend***
Tensile strength [MPa]	2.4 - 3.1	0.5 - 1-5
Elongation at Break [%]	220 - 240	130 - 150
Shore (Scale A)	30 - 35	36 - 50

Table 1: Mechanical properties of Agilus and the blend of Agilus and VeroWhite.

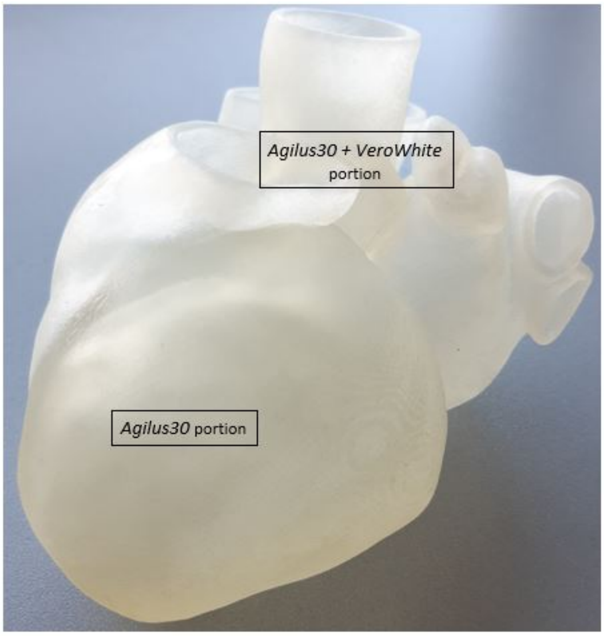


Figure 3: Print result, after supports removal and washing.

2 hours, then support structure was manually removed, with the help of water jet. After the final washing, the model was ready, as can be seen in Fig. 3.

4 RESULTS

Before evaluating the dimensional accuracy, some preliminary qualitative visual and haptic evaluations can be done. The print is characterised by a very smooth surface, with an excellent finishing and reproduction of anatomical details. Moreover, it is translucent, allowing the operator to see through the structures and observe instruments when inserted. As regards the distensibility, the behavior of the portion made with the blended material seems to be very good, characterized by high deformability and elastic recovery. On the contrary, the portion made of pure Agilus has a different behavior: in correspondence with the right ventricle, where the wall is not so thick, properties are satisfactory, while at the left ventricle location, where the wall is much thicker, the model results to be a bit too stiff and not adequately deformable. Moreover, at the transition from a material to the other, no clear distinction is perceptible both from a visual and a tactile point, suggesting excellent integration between parts.

To evaluate the dimensional accuracy of our printed heart model, with respect to the virtual one, a

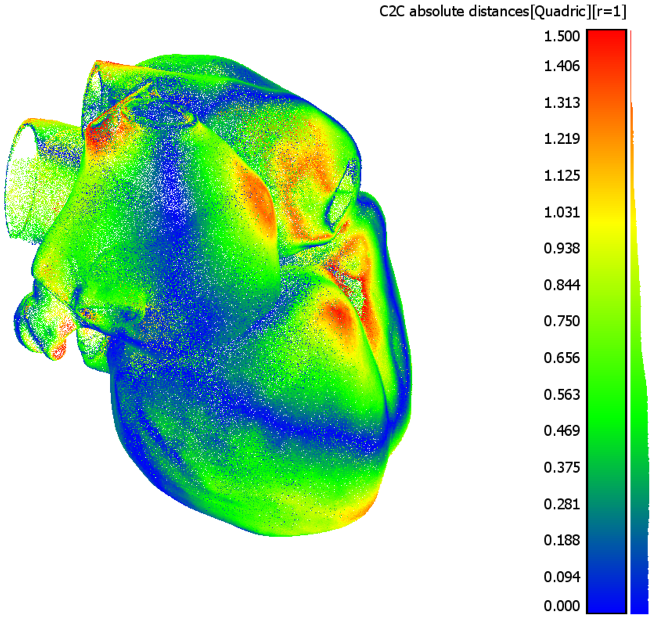


Figure 4: Cloud-to-Cloud distance estimation between the CAD and the printed model. The colors represent the unsigned distance between the two point cloud and the values are in mm. The mean distance and the standard deviation are 0.49 mm and 0.33 mm, respectively.

NextEngine Ultra HD laser scanner has been used. This is a triangulation-based scanner, endowed with a rotating table. Because of the translucency of the model, it was uniformly sprayed with a matting agent. The following acquisition setup has been adopted: the scanning distance was set at 250 mm, with a capture density at 440 $points/mm^2$. In this way, the scanner's accuracy is of 100 μm. The model has been surveyed with 12 scans, around 360 degrees. After a first-round, a second one was performed, modifying the orientation of the print on the rotating table, to allow the scanner to see the whole object. This procedure was repeated three times, to reach statistical relevance. The raw data of each acquisition were registered and then post-processed in ScanStudio 2.0, the proprietary software of the scanner. Results were then exported as point clouds. The comparison between this point clouds and the original geometry was performed in CloudCompare (version 2.11, GPL software, http://www.cloudcompare.org/). The CAD model was imported and sampled with 1 million points, in order to get more robust results while comparing with the acquired point cloud. Then, the two point clouds - i.e. the one from the original model and the one acquired - has been registered taking advantage of the ICP algorithm exposed by CloudCompare. At this point, the distance between the two point clouds has been computed. The original point cloud has been taken as reference and the Cloud-to-Cloud distance tool by CloudCompare has been used: the algorithm computes the nearest neighbour distance globally, that is then refined using a least-square best-fitting plane with a quadratic formulation. The results of the geometrical comparison are shown in Fig. 4. The contour map indicates the unsigned distances, in millimeters, between the two point clouds. The mean and the standard deviation are 0.49 mm and 0.33 mm, respectively. Considering we are dealing with flexible materials, these results can be considered fully satisfactory, especially if compared with those ones obtained from rigid material models (see e.g. [14]), whose values are in the same order of magnitude. These data can be taken as a preliminary confirmation about the validity of the proposed workflow to produce compliant multi-material heart models for education or surgical training purposes.

5 DISCUSSION AND CONCLUSION

With this first phase of the work we were able to produce a realistic compliant patient-specific heart model. However, to make it effectively useful in clinical practice, some improvements are needed. First of all, the mechanical behavior of the print should be finely tuned, in order to reach a better agreement between the model and its biological counterpart. We understood how, if we want a fully compliant model printed with this kind of technology, the wall thickness must be not to exceed a certain threshold: indeed, even if the softest available material was chosen, walls with a thickness close to 10 mm end up to be too rigid eventually.

A solution could be artificially tuning the wall thickness and the material blend, in a way that the final behavior of the printed object is consistent with surgeon's expectations. In other words, an optimization procedure whose objective would be to confer at the printed heart stiffness very close to the real one could be desirable. To this end, a deeper understanding of the biomechanics of the heart, very complex because of its non-linear and anisotropic properties, and the capability of predicting the behavior of the printed materials are needed. To the best of authors' knowledge, such knowledge is not totally available in the literature. So, comprehensive experimental tests on available materials will constitute a key step: tensile and cycling loading, as well as dynamic mechanical analysis (DMA) and stress relaxation tests should be conducted [1], then comparing obtained data with those ones that can be found in the literature. Rising in complexity, a methodology for mapping and rescaling the mechanical properties of the heart on the set of materials available for the printer could be implemented. This could be performed both computationally and experimentally. In the former case, a constant pressure could be assigned in correspondence with the region occupied by the blood pool; in this way, once assigned the materials properties, the deformations field could be extracted and evaluated. In the latter case, a proper test bench could be conceived, exploiting for example pressurized liquid to be injected in the flexible print [3]. By means of adequately located strain gauges, obtained deformation values could be compared with those ones from *in silico* simulations.

As regards the accuracy evaluation performed, it was said how obtained results can be considered satisfactory. Just very limited portions of the printed model exceed 1 mm of difference with respect to the original STL. Anyway, we have to consider we are dealing with a flexible model: even if its stiffness is enough to guarantee the print not be macroscopically affected by deformations due to gravity, the employed gripping system could have slightly altered the final shape.

Another key point is related to introduction of heart valves in the printed model. From the available CT images, these structures cannot be effectively isolated: for this reason, they are not present in our model. The combination of the CT images with ad-hoc imaging techniques, such as 3D echocardiography, would be able to capture valves leaflets, allowing us to enrich the model in sense. Even the introduction of an accurate replica of *chordae tendineae*, cords of connective tissue that connect the papillary muscles to atrioventricular valves, would be a very demanding challenge, to our knowledge not yet tackled in the literature.

A further goal is related to the possibility of speeding up and automating the segmentation procedure, in order to make this technology more suitable for operative environments. Segmentation is a very time-consuming and tedious activity, subject to intra- and inter-observer variability and requires dedicated expert operators [12], especially in case of very complex anatomies. For example, the process to obtain our hollow heart model, from the raw images to the final smoothed model, without considering the printing phase, took the operator a few tens of hours. In order to implement an effective technology transfer to an operating environment, one week-person to provide clinicians with a patient's model could be in many cases excessive. Novel techniques that allow automatic segmentation of complex cardiac and extracardiac structures (e.g. atlas-based segmentation) are gradually taking place, making 3DP easier to be effectively operated in clinical contexts [11], but we are still far away from accessible and robust tools.

To conclude, a summary cost analysis can be done. In general, 3DP technology has now reached a cost that is not prohibitive [8]. Indeed, prices of 3DP printers for this kind of application have noticeably decreased in recent years, giving the possibility to multiple subjects to access them. This is especially true for Fused

Deposition Modelling (FDM) or Stereolithography (SLA) printers, while for printers as employed in this work costs are still a bit higher. Excluding the cost of the machine itself, the price for the raw material is affordable (excluding the support, the weight of our model is about 500g, with a cost per kg of resins of about 300€). We have also to consider that in many cases just specific portions of the heart can be printed, according to the particular needs: in this way, costs further reduce. The other potential cost item is given by segmentation software license: even if nowadays valid open-source segmentation solutions are available, they are oftentimes limited to research activities and not mature for a real operative environment (e.g. due to lack of approvals from regulatory bodies). Even if with some limitations, open-source solutions anyway can guarantee accurate segmentation results in most of the cases (see e.g. [2]). The need for well-trained personnel, with competencies ranging from Medicine to Engineering, is however a must to get successful results.

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