

Design, fabrication, and computational biomechanical evaluation of a multi-material synthetic proximal femur surrogate using PolyJet technology

L Mendoza^{1,2}, M S Malloni^{2,3}, F K Ciliberti², P Gargiulo^{2,4} and Y A Tesfahunegn⁵

¹International University of Catalonia, Barcelona, Spain

²Institute of Biomedical and Neural Engineering, Reykjavik University, 101 Reykjavik, Iceland

³University of Padua, Padua, Italy

⁴Department of Science, Landspítali, University Hospital of Iceland, 101 Reykjavik, Iceland

⁵Department of Engineering, Reykjavik University, 101 Reykjavik, Iceland

E-mail: mariastella.malloni@studenti.unipd.it, laia.mg722@gmail.com, federicac@ru.is, yonatan@ru.is, paolo@ru.is

Abstract. Synthetic bone surrogates offer a promising alternative to cadaveric specimens for orthopaedic biomechanics research. This study presents an integrated experimental–computational framework for designing and evaluating multi-material PolyJet-printed surrogates of the proximal femur. Patient-derived CT data were used to reconstruct healthy and osteoporotic geometries, while printable materials and trabecular lattice porosities (69.44–93.65%) were screened through displacement-controlled tensile tests on dog-bone specimens. Finite element models of the specimens reproduced the experimental stresses within 10% for the selected materials. The characterised properties were then assigned to proximal femur models under compressive loading and compared with a literature-based bone reference. The synthetic femora reproduced the von Mises stress distribution of native bone (maximum stress differences below 0.6%), but the cortical stiffness gap ($E \approx 3.3$ vs. 17.5 GPa) produced approximately fourfold greater deformation, and the osteoporotic synthetic model fell below yield (FoS = 0.96) at 800 N. No printed femur was mechanically tested; femoral behaviour was assessed computationally using experimentally calibrated properties. These findings support PolyJet printing as a route to reproducible, patient-specific femoral surrogates, pending physical validation.

1 Introduction

Additive manufacturing (AM) is increasingly used in healthcare to produce customizable, patient-specific anatomical models for surgical planning, device testing, and biomechanical research [1, 2]. In particular, multi-material PolyJet printing combines materials with different mechanical properties in a single print, allowing the heterogeneous behaviour of biological tissues to be reproduced [1]. Printed models are a reproducible alternative to cadaveric specimens, whose use is limited by availability and large inter-specimen variability [3].

The proximal femur is a particularly relevant target. Hip fractures are a major consequence of osteoporosis, with high mortality and healthcare costs in the elderly [4, 5]. Cortical bone (approximately 17,500 MPa) and trabecular bone (approximately 400 MPa) differ substantially in their contribution to load transfer [6, 7, 8], and osteoporosis degrades both compartments [9, 10], so healthy and osteoporotic conditions must be represented. However, bone behaviour depends not only on the base material but also on architecture and porosity [11, 12]. In a printed surrogate, trabecular stiffness can be tuned through lattice parameters, although PolyJet is limited by its minimum feature size and by support removal from fine cavities [13, 14]. Experimental characterisation and numerical validation, such as FEA of subject-specific femora [15, 16], are therefore needed.

Previous 3D-printed bone models have used FDM, SLA, or PolyJet, mostly for single-material trabecular structures or scaffolds [17, 18, 19, 20], while commercial composite femurs are not patient-specific and osteoporotic variants are obtained mainly by changing the foam density [21]. To our knowledge, no study has combined experimental material characterisation, numerical verification of the material model, and patient-specific healthy and osteoporotic proximal femora within a single workflow.

In this study, an experimental-computational workflow was developed to evaluate PolyJet-printed materials as synthetic bone surrogates for the proximal femur. Candidate materials were characterised on tensile specimens, and the resulting properties were used to verify finite element models and then assigned to patient-specific healthy and osteoporotic femoral geometries under compressive loading. The main contributions are: (i) independent tuning of the photopolymer formulation and lattice architecture to represent healthy and osteoporotic trabecular bone; (ii) an integrated characterisation-FEA patient-specific workflow; and (iii) a quantification of current PolyJet limitations (cortical stiffness gap and lattice resolution limit) and their effect on femoral compliance. The anatomical models were evaluated exclusively through FEA; no physical femur was fabricated or tested.

2 Materials and methods

2.1 Anatomical design

Patient-specific CT images of the proximal femur of two female patients aged 60–65 years were segmented using Materialise Mimics [22]. One dataset corresponded to a healthy patient and the other to a patient with osteoporosis, resulting in two independent patient-specific proximal femoral models. CT images were imported in DICOM format and segmented based on intensity thresholds expressed in Hounsfield Units (HU). The HU range of the acquired datasets was from -1024 to 1851 HU. The femoral neck was isolated from the surrounding anatomical structures using threshold-based segmentation and region-growing operations, and the cortical compartment was segmented by intensity thresholding. Because the CT resolution does not resolve the individual trabeculae, the trabecular compartment was not segmented as bone tissue but defined as a continuous solid volume filling the femoral interior (the region enclosed by the cortical shell), using a wide HU range (-400 to 1617 HU) so that the whole medullary cavity was included. CT intensities were used only to define the geometry; no HU–density–modulus relationship was

applied. Manual slice-by-slice corrections were performed using the available mask-editing tools to remove segmentation artefacts and ensure anatomical continuity. For the osteoporotic model, additional manual refinement was required in regions with reduced bone density and lower contrast between adjacent tissues. The cortical and trabecular masks were then converted into three-dimensional models, followed by Wrap and Smooth operations to improve surface quality and remove minor geometric irregularities. Finally, the resulting geometries were exported as watertight STL files for subsequent finite element analysis. The main segmentation workflow is illustrated in Figure 1.

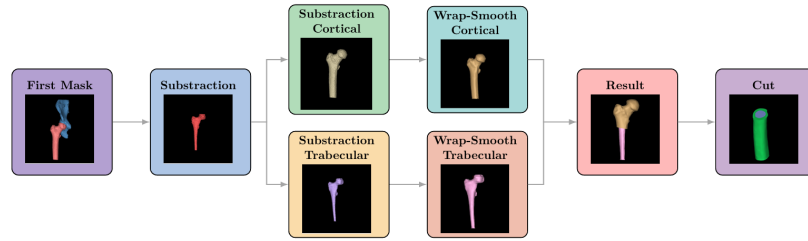


Figure 1: Workflow used for the segmentation and reconstruction of the cortical and trabecular compartments of the patient-specific proximal femur models.

2.2 Multi-material design and fabrication

Three ASTM D638 Type IV dog-bone specimens[23] were manufactured for each combination using a Stratasys J850TM Digital AnatomyTM printer based on PolyJet technology [24]. Following fabrication, the soluble support material (SUP706) was removed according to the manufacturer’s recommendations, and all specimens were visually inspected before testing. The geometry and dimensions of the ASTM D638 Type IV specimens are shown in Figure 2.

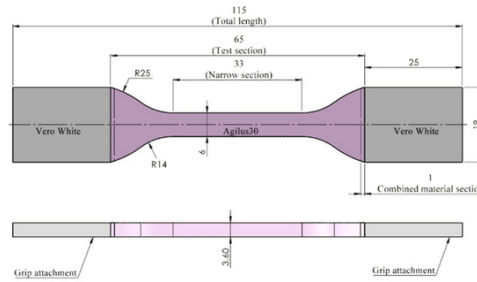


Figure 2: Geometry and dimensions of the ASTM D638 Type IV dog-bone specimens used for tensile testing, based on [23]. All dimensions are in mm.

Candidate materials included Flexible Digital Materials (FLXA series), Rigid Digital Materials (RGDA series), and VeroClear, selected to cover a broad range of mechanical responses relevant to cortical and trabecular bone. Material selection was based on experimental mechanical characterisation rather than solely on manufacturer specifications.

As shown in Table 1, for the healthy trabecular configuration, specimens T1–T6 were fabricated using FLXA formulations with different nominal Shore hardness values while maintaining the same geometric lattice. The theoretical volumetric fractions of Agilus30 Clear and VeroClear were estimated from the nominal hardness of each formulation using a linear rule-of-mixtures approach:

$$H_c = H_{\text{Agilus}}V_{\text{Agilus}} + H_{\text{Vero}}V_{\text{Vero}}, \quad (1)$$

where H_c is the nominal Shore hardness of the formulation, and V_{Agilus} and V_{Vero} are the corresponding volumetric fractions of Agilus30 Clear and VeroClear, respectively, such that

$$V_{\text{Agilus}} + V_{\text{Vero}} = 1. \quad (2)$$

Nominal Shore hardness values of 30 for Agilus30 Clear and 100 for VeroClear were used as the constituent material values. The resulting theoretical fractions for each formulation are reported in Table 1.

The same trabecular lattice architecture was applied to specimens T1–T8, with a lattice spacing of 1.0 mm and a filament thickness of 0.2 mm, corresponding to an analytically determined macrostructural porosity of 69.44%. T1–T6 used FLXA formulations, whereas T7 and T8 used RGDA 8625 and RGDA 8630, respectively. The corresponding material and architectural parameters are summarised in Table 1. The trabecular architecture was generated using the lattice parametrisation available in the Stratasys printing software (spacing and filament thickness), rather than a custom-designed topology. Parameters were chosen to span the achievable range of void fraction within the printer’s resolution limits.

Table 1: Healthy trabecular material formulations.

Sample	Material	Agilus30 (%)	VeroClear (%)	Spacing (mm)	Thickness (mm)	Porosity (%)
T1	FLXA 9940	85.7	14.3	1.0	0.2	69.44
T2	FLXA 9950	71.4	28.6	1.0	0.2	69.44
T3	FLXA 9960	57.1	42.9	1.0	0.2	69.44
T4	FLXA 9970	42.9	57.1	1.0	0.2	69.44
T5	FLXA 9985	21.4	78.6	1.0	0.2	69.44
T6	FLXA 9995	7.1	92.9	1.0	0.2	69.44
T7	RGDA 8625	6.0	94.0	1.0	0.2	69.44
T8	RGDA 8630	10.0	90.0	1.0	0.2	69.44

Agilus30 and VeroClear percentages represent theoretical volumetric fractions estimated from the nominal Shore hardness values using a linear rule-of-mixtures approach. The exact compositions of the commercial digital materials are proprietary.

To investigate the feasibility of reproducing osteoporotic trabecular bone, additional specimens were manufactured using the FLXA 9985 and FLXA 9995 formulations, which were identified as the mechanically strongest candidates for the trabecular compartment. Osteoporotic-like trabecular degradation was simulated by modifying the internal lattice architecture to reproduce the increased structural rarefaction and void space associated with osteoporosis [25, 26]: compared with the healthy configuration (69.44% porosity), the lattice spacing was increased from 1.0 mm to 2.0 or 3.0 mm and the filament thickness was reduced to 0.1 mm [24], giving macrostructural porosities of 90.70% and 93.65%, respectively. These values were determined by the achievable combination of lattice spacing and filament thickness rather than selected from a predefined target. The 90.70% configuration substantially increased the void fraction while maintaining sufficient structural integrity for reliable PolyJet fabrication, whereas the 93.65% configuration, intended to represent more severe rarefaction, approached the geometric resolution limits of the process and resulted in lower structural stability. Both configurations were

therefore constrained by the minimum printable feature size and the need to maintain structural continuity during post-processing (Table 2).

Table 2: Osteoporotic trabecular configurations.

Config.	Materials	Spacing (mm)	Thickness (mm)	Porosity (%)
OP-A	FLXA 9985, FLXA 9995	3.0	0.1	93.65
OP-B	FLXA 9985, FLXA 9995	2.0	0.1	90.70

For the cortical compartment, solid RGDA and VeroClear formulations were evaluated without an internal lattice. VeroClear was selected as the reference material for healthy cortical bone based on its experimentally measured mechanical response, while RGDA formulations with reduced mechanical properties were considered as candidates for representing degraded cortical tissue. The final material assignments were based on the experimentally measured mechanical properties of the printed specimens rather than solely on nominal manufacturer specifications. The material formulations and corresponding configurations are summarised in Table 3.

Table 3: Material formulations evaluated for the cortical compartment.

Sample	Material	Architecture	Application
C1	VeroClear	Solid	Healthy cortical
C2	RGDA 8630	Solid	Cortical candidate
C3	RGDA 8425	Solid	Cortical candidate
C4	RGDA 8430	Solid	Osteoporotic cortical
C5	RGDA 8625	Solid	Cortical candidate

2.3 Mechanical characterization

Uniaxial tensile tests were performed on ASTM D638 Type IV dog-bone specimens using an Instron Static 34TM-50 universal testing machine equipped with a 50 kN load cell [27]. Tests were conducted at room temperature under displacement-controlled loading at a constant crosshead speed of 50 mm/min. Force and displacement were continuously recorded until specimen failure.

Prior to testing, the width and thickness of each specimen were measured individually using a digital caliper to determine the actual initial cross-sectional area. The measured dimensions were used for the calculation of engineering stress, accounting for geometric variability associated with the 3D-printing and post-processing procedures.

The experimental force–displacement data were processed using a custom MATLAB R2023b script to obtain engineering stress–strain curves [28]. The resulting stress–strain curves are presented in Figure 4. The initial seating of the specimen in the grips produced a non-zero initial load, which was significant for the highly porous osteoporotic lattices. For material characterisation, this initial offset was removed by subtracting the initial stress value from the whole stress curve, and the strain was referenced to zero at the start of loading. The corrected curves were then used to determine Young’s modulus, the yield stress (0.2% offset method) and the ultimate tensile strength. The resulting mechanical properties are reported in Tables 5, 6 and 7. For the cortical and healthy trabecular specimens, this offset is negligible and the corrected stresses coincide with the values used in the numerical–experimental comparison (Table 8). In contrast, for the osteoporotic lattices (porosity above 90%), which are very soft and whose cavities may retain residual support material, the initial offset is significant. The numerical simulations of these specimens (Section 2.4) were run with the total force recorded by the machine, including this initial load, so the σ_{exp} values in Table 8 are higher than the corrected σ_{max} in Table 6.

2.4 Finite element analysis

FEA was performed using ANSYS Mechanical [29]. in two consecutive phases: (1) validation of the numerical model using the experimental tensile specimens and (2) biomechanical evaluation of the proximal femur under compressive loading. An overview of the two-phase FEA workflow is presented in Figure 3.

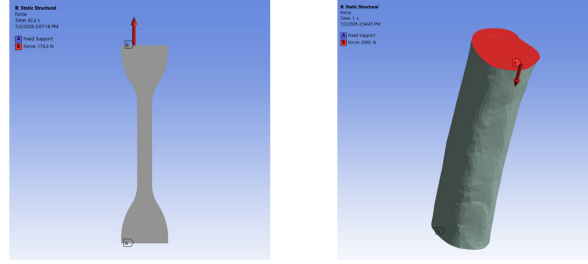


Figure 3: Boundary conditions applied in the finite element models for (a) the experimental tensile specimen and (b) the proximal femur.

Phase 1: Validation of the numerical model using standard test specimens.

The first phase was performed to assess the ability of the numerical model to reproduce the mechanical response experimentally measured for the selected 3D-printed material formulations. Numerical models reproducing the geometry and loading conditions of the ASTM D638 Type IV tensile specimens were developed in ANSYS. For each formulation, the experimentally determined Young's modulus, together with the corresponding density and Poisson's ratio, were assigned to the numerical model.

The equivalent von Mises stress obtained from the numerical simulations (σ_{ANSYS}) was compared with the experimental tensile tests (σ_{exp}). The percentage deviation between both values was calculated to assess the predictive accuracy of the numerical model. It should be noted that, for the cortical and healthy trabecular specimens, the initial load offset described in Section 2.3 is negligible, so the σ_{exp} values coincide with the corrected values reported in Section 3.1. For the osteoporotic trabecular specimens, in contrast, σ_{exp} was obtained from the total force recorded by the testing machine, including the initial load due to specimen seating, and the simulations were run with this same force. Consequently, these σ_{exp} values are higher than the corrected σ_{max} reported in Section 3.1. The total force was used so that the numerical results could be compared directly with the experimentally measured response under the same nominal loading conditions. A deviation below 10% was considered an acceptable level of agreement for engineering purposes. The validation procedure was applied to the cortical, healthy trabecular, and osteoporotic trabecular configurations.

Phase 2: Anatomical femur simulation.

Following the experimental characterisation and numerical validation, two patient-specific proximal femoral models, one healthy and one osteoporotic, were generated from the corresponding anatomical segmentations. Only the proximal femoral segment was considered; each model was sectioned at the subtrochanteric level to provide a planar distal surface for the application of the boundary condition. The cortical shell and trabecular compartment were independently reconstructed and maintained as separate anatomical regions.

The two anatomical models were discretized using an unstructured mesh of tetrahedral elements. A mesh convergence analysis was performed with element sizes of 3.0, 2.0 and 1.4 mm, using the maximum total deformation as the convergence criterion. The 3.0 and 1.4 mm meshes

resulted in variations of -0.06% and $+0.03\%$, respectively, relative to the 2.0 mm mesh, so a 2.0 mm element size was selected as a converged solution with an appropriate balance between numerical accuracy and computational cost.

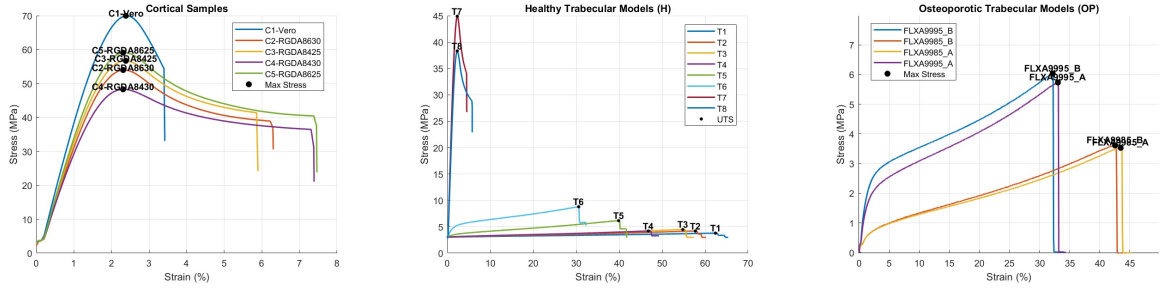
Two material configurations were evaluated for each anatomical model: a literature-based configuration, using the properties reported for healthy or osteoporotic human bone in Table 4 [6, 7, 30], and a synthetic-material configuration, using the experimentally characterised properties of the selected 3D-printed materials. For the healthy model, VeroClear was assigned to the cortical compartment and FLXA 9995 to the trabecular compartment; for the osteoporotic model, RGDA 8430 and FLXA 9995-B were assigned, respectively. These formulations were selected for their experimentally characterised mechanical properties and their correspondence with the mechanical behaviour of the respective bone tissues. The FEA thus comprised four anatomical simulations (healthy and osteoporotic femur, each with literature-based and synthetic-material properties), run on the same segmented geometry with identical boundary and loading conditions to enable direct comparison of their structural responses.

The biomechanical behaviour under compressive loading was reproduced by applying the following boundary conditions (Figure 3b). The distal planar surface, obtained from the subtrochanteric cut, was defined as a fixed support, with displacements constrained in all three spatial directions ($u_x = u_y = u_z = 0$). The opposite flat surface was defined as the compression side, on which each compressive force was applied axially, along the long axis of the model and normal to the loaded surface. The cortical and trabecular regions were obtained from complementary segmentations, so that they share a common interface when assembled, and were connected through bonded contact, assuming perfect adhesion between the two compartments.

Compressive loads ranging from static posture to severe overloads (800, 1150, 1510, 2000 and 4000 N) were simulated. Structural integrity was evaluated through the Minimum Factor of Safety (FoS_{min}), calculated locally in ANSYS as the ratio of the material yield strength (σ_{yield}) to the equivalent von Mises stress (σ_{ANSYS}). For the literature-based biological models, the onset of yielding was additionally contextualized using Bayraktar’s biological limit-strain criterion ($\epsilon_y = 0.8\%$) [31]. For the synthetic models, yielding was evaluated directly against the experimental yield strength (σ_{yield}) determined from the uniaxial tensile characterization of the respective photopolymer formulations (VeroClear, RGDA and FLXA). Consequently, 1510 N was selected for the healthy model and 800 N for the osteoporotic model as representative loading conditions for comparison, under static body weight ($1 \times BW$) and load-to-failure conditions, respectively.

Table 4: Mechanical properties assigned to the literature-based femoral models. The elastic modulus values were obtained from Bittner-Frank et al. [7].

Condition	Compartment	E [MPa]	ν	ρ [g/cm ³]	ϵ_y [%]	σ_y [MPa]
Healthy	Cortical	17,500	0.30	1.9	0.7–1.0	100–130
Healthy	Trabecular	400	0.30	1.4	0.8–1.0	5
Osteoporotic	Cortical	10,900	0.30	1.7	0.7–0.9	70–90
Osteoporotic	Trabecular	100	0.30	1.2	0.8–1.0	1.5



(a) Tensile stress-strain curves of cortical bone samples (C1–C5). (b) Tensile stress-strain curves of healthy trabecular models (T1–T8). (c) Tensile stress-strain curves of osteoporotic trabecular models (OP).

Figure 4: Tensile stress-strain curves of cortical (a), healthy trabecular (b), and osteoporotic trabecular (c) models. Black dots indicate peak stress points.

3 Results and discussion

3.1 Experimental Mechanical Characterisation of Materials

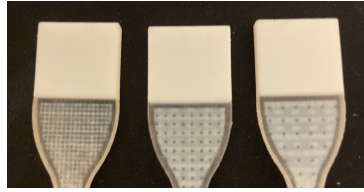


Figure 5: Porosity comparison between the healthy and osteoporotic conditions: healthy bone (left), osteoporotic condition A (middle), and osteoporotic condition B (right).

Uniaxial tensile testing was performed to determine the experimental elastic modulus (E), maximum stress ($\sigma_{\max}$), and yield stress (σ_{yield}) of the photopolymer formulations across solid and porous configurations.

Healthy trabecular models (H Configuration)

The porosity distribution for configuration H is shown in Figure 5, with the corresponding MATLAB-processed tensile stress-strain curves reported in Figure 4b. The derived mechanical parameters are summarized in Table 5.

While the fully dense FLXA 9995 formulation exhibits an elastic modulus compatible with healthy trabecular bone, its bulk tensile strength ($\sigma_{\max} = 12.85$ MPa) overestimates the biological benchmark. Introducing a controlled geometric porosity of 69.44% (specimen T6-FLXA 9995) resulted in a maximum tensile stress of 8.88 ± 0.30 MPa and a yield stress of 5.63 MPa, both within the reported ranges for healthy trabecular bone. However, its Young's modulus (150.37 MPa) was substantially lower than the literature reference (400 MPa). This difference may be partly attributed to the polymeric nature of the material, whose molecular structure and chain mobility can contribute to lower stiffness. Therefore, T6 reproduces the strength of healthy trabecular bone within the selected reference ranges, but does not fully reproduce its elastic stiffness.

The yield stress of specimen T4-FLXA 9970 was not reported because the mechanical test was unsuccessful, making the resulting data unreliable. Therefore, no reliable yield stress value could be determined for this specimen.

Osteoporotic trabecular models (OP Configuration)

To replicate pathological bone tissue, mechanical characterisation focused on the FLXA 9985 and FLXA 9995 formulations evaluated across two porosity levels: Configuration A (nominal porosity 93.65%) and Configuration B (nominal porosity 90.70%), as shown in Figure 5. The resulting MATLAB-processed tensile stress-strain curves are presented in Figure 4c, while the extracted mechanical parameters are summarized in Table 6.

Although FLXA 9985 formulations exhibited elastic moduli closer to clinical targets, Configuration A pushed PolyJet technology to its geometric resolution limits, leading to structural instability and higher experimental deviations. Consequently, **FLXA 9995-B** was selected as the optimal osteoporotic surrogate; it provided adequate printing stability and a yield strength of $\sigma_{\text{yield}} = 1.96$ MPa, consistent with the pathological target range for osteoporosis (1–3 MPa). The slightly elevated elastic modulus observed in these high-porosity lattices is attributed to a minor clogging effect caused by residual support gel trapped within the micro-cavities.

Table 5: Mechanical properties of healthy porous trabecular samples (Configuration H). *Values for natural healthy trabecular bone in the last row are reported from literature [7].

Sample	σ_{max} (MPa)	E (MPa)	σ_{yield} (MPa)	Density (g/cm ³)	σ_f (MPa)
T1-FLXA 9940	3.79	3.58	3.69	0.352	2.98
T2-FLXA 9950	4.19	7.10	4.09	0.354	2.99
T3-FLXA 9960	4.48	10.14	4.02	0.355	3.00
T4-FLXA 9970	4.31	10.27	–	0.356	3.00
T5-FLXA 9985	6.21	42.59	4.16	0.358	3.00
T6-FLXA 9995	8.88	150.37	5.63	0.360	3.01
T7-RGDA 8625	44.90	2129.40	44.87	0.360	19.57
T8-RGDA 8630	38.32	1768.70	38.22	0.360	16.54
Trabecular Bone(H)*	5–10	400	3–6	1.400	–

Table 6: Mechanical properties of osteoporotic (OP) trabecular bone specimens evaluated through tensile testing. *Values for natural osteoporotic trabecular bone in the last row are reported from literature [32].

Sample	σ_{max} (MPa)	σ_{yield} (MPa)	E (MPa)	Porosity [%]
FLXA 9995-B	6.03	1.96	176.70	90.70
FLXA 9985-B	3.61	0.62	43.85	90.70
FLXA 9985-A	3.53	0.59	46.31	93.65
FLXA 9995-A	5.73	1.58	139.57	93.65
Trabecular OP*	2–5	1–3	100.00	–

Cortical models

The specimens for the cortical compartment were produced in fully dense, homogeneous configurations without induced porosity. Their MATLAB-processed tensile stress-strain curves are presented in Figure 4a, while the extracted mechanical properties are summarized in Table 7 alongside reference values for human cortical bone.

Photopolymer resins are inherently limited in matching the extreme stiffness of the mineralized matrix of natural cortical bone ($E \approx 17.5$ GPa). Consequently, **C1–VeroClear** was

selected for the healthy cortical compartment ($\sigma_{\max} = 69.98$ MPa, $E = 3311.80$ MPa) as it provides the highest modulus available within the PolyJet system. For the osteoporotic cortical compartment, **C4–RGDA 8430** was chosen, exhibiting a tensile strength of $\sigma_{\max} = 48.30$ MPa and a yield stress of $\sigma_{\text{yield}} = 47.75$ MPa, closely matching the target stress reduction for pathological cortical tissue.

Table 7: Mechanical properties of fully dense cortical samples.

Sample	$\sigma_{\max}$ (MPa)	σ_{yield} (MPa)	E (MPa)	UTS (MPa)
C1–VeroClear	69.98	69.33	3311.80	33.04
C2–RGDA 8630	53.99	53.40	2675.10	30.57
C3–RGDA 8425	56.71	56.48	2755.50	24.21
C4–RGDA 8430	48.30	47.75	2487.30	21.07
C5–RGDA 8625	59.09	58.94	2867.40	23.81
Cort. Healthy*	130.00	80.00	17500.00	–
Cort. Osteoporotic*	90.00	60.00	10900.00	–

*The last two rows report reference literature values for healthy and osteoporotic human cortical bone [6].

3.2 Results of numerical simulations (FEA)

Phase 1: Validation of the numerical model on standard test specimens (“Dog-Bone”) The initial computational phase assessed the predictive accuracy of the ANSYS model by comparing the experimental tensile stresses (σ_{exp}) with the equivalent von Mises stresses from FEA (σ_{ANSYS}), assigning the experimental Young’s modulus to each formulation and literature values of density and Poisson’s ratio (ν). The comparison is summarized in Table 8. Relative errors remained below 10% for the cortical, healthy trabecular, and 90.70% porosity osteoporotic configurations, whereas larger deviations were observed for the 93.65% porosity configurations. For the osteoporotic specimens, σ_{exp} corresponds to the total force recorded by the testing machine, including the initial seating load (Sections 2.3 and 2.4), and is therefore higher than the corrected $\sigma_{\max}$ in Table 6.

Table 8: Validation of the FEA model: Comparison between experimental stress (σ_{exp}) and numerical von Mises stress (σ_{ANSYS}) for standard test specimens (“dog-bone”).

Category	Specimen / Mixture	Force [N]	σ_{exp} [MPa]	σ_{ANSYS} [MPa]	Rel. Error [%]
Cortical	C1 – Vero Clear	1350.0	69.98	74.12	5.90
	C4 – RGDA8430	950.3	48.30	52.16	7.90
Trabecular (H)	T6 – FLXA 9995	179.3	8.88	9.75	9.80
	T7 – RGDA 8625	851.4	44.90	46.75	4.11
Trabecular (OP)	FLXA-9985-B	142.6	7.32	7.83	6.90
	FLXA-9995-B	191.3	9.70	10.50	8.20
	FLXA-9995-A	144.4	9.42	7.93	15.80
	FLXA-9985-A	185.8	7.23	10.19	40.00

Phase 2: Compression Simulation on the Anatomical Femoral Model Based on experimental characterisation, the selected optimal polymer surrogates (C1 for healthy cortical, C4 for osteoporotic cortical, T6 for healthy trabecular, and FLXA 9995-B for osteoporotic

trabecular) were implemented into the macro-anatomical finite element model of the proximal femur. Table 9 explicitly compares their properties against physiological targets.

Table 9: Explicit comparison between experimental mechanical properties of selected bone surrogates and human bone literature benchmarks.

Bone Condition	Selected Sample	Young’s Modulus E [MPa]		$\sigma_{\max}$ [MPa]		σ_{yield} [MPa]	
		Exp.	Lit.	Exp.	Lit.	Exp.	Lit.
Healthy Cortical	C1 – VeroClear	3311.80	17500.00 [6]	69.98	130.00 [6]	69.33	80.00 [6]
Osteoporotic Cortical	C4 – RGDA 8430	2487.30	10900.00 [6]	48.30	90.00 [6]	47.75	60.00 [6]
Healthy Trabecular	T6 – FLXA 9995	150.37	400.00 [7]	8.88	5.00–10.00 [7]	5.63	3.00–6.00 [7]
Osteoporotic Trabecular	FLXA 9995-B	176.70	100.00 [32]	6.03	2.00–5.00 [32]	1.96	1.00–3.00 [32]

The biomechanical performance of the proximal femur under compressive loading is summarized in Table 10. A detailed analysis reveals the following key insights:

- **Stress state distribution and ‘shell’ effect:** The maximum von Mises stress showed numerical agreement between biological and synthetic models, with deviations of only 0.45% in the healthy scenario (−1510 N) and 0.51% in the osteoporotic scenario (−800 N). Graphically (Figure 6a,b and Figure 7a,b), the synthetic twin reproduces the functional “shell effect” of native bone: high stress concentrations are borne along the outer cortical wall, effectively shielding the inner trabecular core.
- **Total Deformations (TD) and energy analysis:** Conversely, kinematic displacement analysis demonstrated a macro-level discrepancy exceeding 300%, primarily associated with the stiffness mismatch between the synthetic and biological models. The lower elastic modulus of the photopolymer resins contributes to the greater deformability of the synthetic structure. Under an osteoporotic compressive load of −800 N, the synthetic femur displaced 0.784 mm versus 0.192 mm in the natural benchmark (Table 10).

The accumulated elastic strain energy U is defined as:

$$U = \frac{1}{2} F \Delta x \quad (3)$$

Applying Equation 3 to the osteoporotic configuration yields $U_{\text{Synth}} = \frac{1}{2} \cdot 800 \text{ N} \cdot 0.784 \text{ mm} = 0.314 \text{ J}$ for the synthetic model, compared to $U_{\text{Lit}} = \frac{1}{2} \cdot 800 \text{ N} \cdot 0.192 \text{ mm} = 0.077 \text{ J}$ for the biological benchmark. As quantified by these energy calculations, U_{Synth} is more than four times greater than U_{Lit} , reflecting the greater compliance and elastic energy storage of the synthetic photopolymer structure. Since the present FEA model assumes linear-elastic, quasi-static behaviour, this strain energy represents elastic energy storage rather than energy dissipation.

- **Minimum Factor of Safety ($\text{FoS}_{\min}$):** Evaluated using material-specific yield strength (σ_{yield}) for synthetic models and cross-referenced with Bayraktar’s biological strain threshold ($\epsilon_y = 0.8\%$) for the biological reference [31], $\text{FoS}_{\min}$ approached unity at critical compressive loads, correctly predicting yield initiation. This criterion was not considered representative of the yield or failure strain of the synthetic photopolymers, whose mechanical behaviour was experimentally characterised. In the healthy model, the synthetic femur

yielded a slightly higher and more uniform FoS (1.27 vs. 1.01), owing to structural material isotropy. In the osteoporotic model, localized strain accumulation within the flexible cortical shell led to a FoS below unity (0.956 at -800 N).

Table 10: Summary of FEA biomechanical results for healthy and osteoporotic proximal femur models under compressive loading.

Scenario	Material	Load [N]	$\sigma_{\max}$ [MPa]	Total Def. [mm]	Main Eq. Strain	FoS min
Healthy	Literature	1510	17.74	0.076	0.0123	1.01
	Synthetic	1510	17.66	0.338	0.0296	1.27
Osteoporotic	Literature	800	21.37	0.192	0.0291	1.03
	Synthetic	800	21.27	0.784	0.0116	0.96

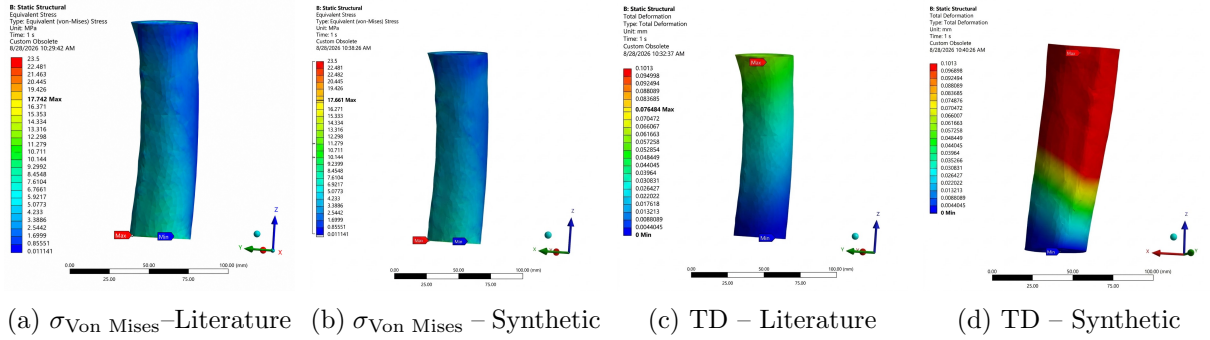


Figure 6: FEM results for the healthy scenario under an axial compressive load of 1510 N: cut section of the proximal femur obtained from CT scan image segmentation.

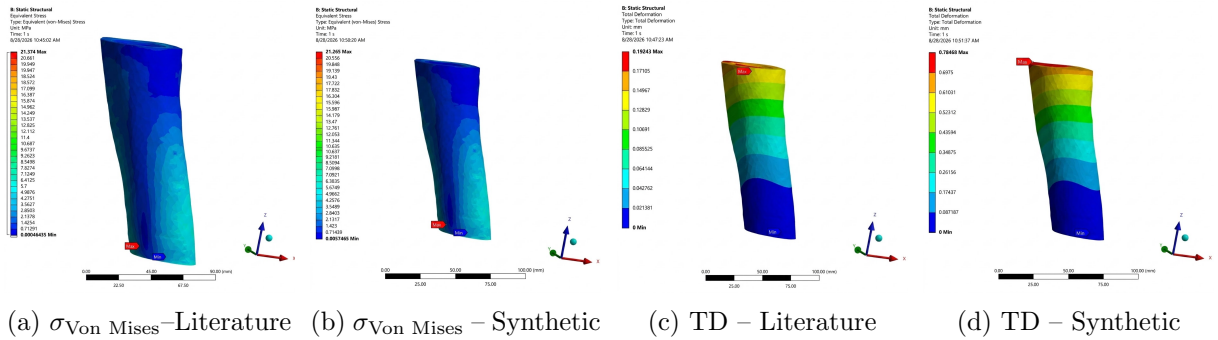


Figure 7: FEM results for the osteoporotic scenario under an axial compressive load of 800 N: cut section of the proximal femur obtained from CT scan image segmentation.

3.3 Limitations and future work

Despite the high reproducibility and predictive capability of the proposed workflow, several limitations should be noted. The primary constraint is the cortical stiffness mismatch: rigid PolyJet photopolymers ($E \approx 2-3$ GPa) cannot match mineralized human cortical bone (11–17.5 GPa),

increasing overall compliance and deformation under load. In the most porous lattices (93.65%, Configuration A), incomplete removal of the water-soluble support gel caused local clogging, artificially altering the lattice stiffness. Moreover, the osteoporotic material properties were obtained from curves corrected for the initial seating load, whereas Table 8 used the total recorded force (Section 2.3); a single data-processing protocol should be adopted in future work.

From a modelling perspective, the isotropic, linear-elastic, quasi-static assumption simplifies the anisotropic and viscoelastic response of real tissue. Material characterisation was performed only under uniaxial tension, whereas the femur is loaded mainly in compression, where low-density lattices may exhibit micro-strut buckling, cell-wall contact and localised collapse. Further limitations are the small sample size, the lack of CT-based spatially varying properties, and the simplified trabecular architecture. Most importantly, the femoral surrogate was evaluated only computationally, so experimental validation with complete printed femora is required before its use in orthopaedic device testing or other biomechanical applications. Additionally, only two patient-specific geometries were analysed, and the loading was simplified to a single axial compression with a fixed distal section and perfectly bonded cortical–trabecular contact. The FEA was validated only on uniaxial tensile coupons, and the 93.65% porosity specimens exceeded the 10% error threshold.

Future work will therefore include compressive coupon testing of all formulations, photopolymer composites reinforced with ceramic micro/nanoparticles (e.g., hydroxyapatite or bioactive glasses) to raise the cortical modulus without compromising printability, improved post-processing (e.g., enhanced ultrasonic cleaning) to ensure open lattice porosity, and dynamic, cyclic loading with in vitro fatigue testing under physiological gait conditions.

3.4 Conclusions

This study demonstrated an integrated experimental–computational workflow for designing patient-specific PolyJet surrogates of the proximal femur under healthy and osteoporotic conditions. Tensile characterisation and finite element verification on standard specimens (errors below 10% for the selected materials) provided the material inputs for the anatomical models. In the femoral simulations, the synthetic models reproduced the von Mises stress distribution and the cortical “shell effect” of native bone, with maximum stress differences below 0.6%. However, the cortical stiffness gap of current PolyJet materials ($E \approx 3.3$ GPa vs. 17.5 GPa) led to approximately fourfold greater deformation and elastic strain energy, and the osteoporotic synthetic model fell below yield at 800 N (FoS = 0.96). The 93.65% porosity lattice approached the printer’s resolution limit, so 90.70% was the practical limit for osteoporotic trabecular bone. Because the femoral models were assessed only computationally and under linear-elastic assumptions, physical compression testing of printed femora is required before these surrogates can be used for implant testing or surgical planning. Overall, the workflow is a promising basis for reproducible, patient-specific femoral surrogates, and stiffer cortical formulations and compressive material characterisation are the main priorities for future work.

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