Characterization and Design Optimization of Periodic Cellular Solids for Optimal Bone Ingrowth in Osseointegration Implants

By

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Abstract

Post-operative complications of spinal lumbar fusion include cage subsidence and non-union. The lack or delayed osseointegration of the fusion cage may be one of the problems resulting in poor outcomes. The use of micro-architected truss-like lattice materials to improve implant osseointegration is getting more attention beyond spinal fusion. Therefore, the main goal of this research is to define the optimal lattice microscopic parameters to maximize bone ingrowth, for fusion spinal cages. A tool based on previous mechano-regulatory models is implemented numerically, to determine the fraction of void space within a structure that receives optimal stimulation for bone ingrowth. The model was used to replicate previous *in silico* and *in vivo* results and showed very good agreement when comparing the different lattice topologies, unit cell orientation, microscopic strut cross-section shape as well as strut size. Characterization of the performance of lattice materials, relevant for osseointegration, is conducted for fourteen unit cell topologies, with ten relative densities, under four axial loadings. Results suggested that Diamond, X-Shape, Tesseract and Rhombic Dodecahedron with horizontal orientation provide the best biomechanical environment for cells to grow and form bone tissue.

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Mom, merci de m'avoir permis de performer à mon plein potentiel.

*"To develop a complete mind: Study the science of art; Study the art of science.
Learn how to see. Realize that everything connects to everything else."*

- Leonardo Da Vinci

Preface

This research project is an original work developed by Ariane Parisien, under the co-supervision of Professor Mostafa El Sayed and Professor Hanspeter Frei. The thesis is conforming to Carleton University mandatory requirements for the Master of Applied Science in Biomedical Engineering. During the research project, a literature review investigating subsidence occurrence following five lumbar interbody fusion surgery procedures, was done. Chapter 2.3.2 is an excerpt of this review paper entitled "Subsidence of spinal fusion cages: a systematic review" which will be submitted for publication [1].

Table of Contents

Abstract	i
Acknowledgements.....	ii
Preface	iv
Table of Contents	v
List of Tables.....	viii
List of Figures	x
Nomenclature.....	1
List of Abbreviations.....	2
CHAPTER 1 - INTRODUCTION	4
1.1 - Motivation.....	4
1.2 - Research Objectives	5
1.3 - Organization of the Thesis.....	6
CHAPTER 2 - LITERATURE REVIEW	7
2.1 - Bone Tissue	7
2.1.1 Bone structure.....	7
2.1.2 Bone Remodeling and Regeneration	8
2.2 - Mechanobiological Model (FEA)	10
2.2.1 Biophysical Stimulus	10
2.2.2 Diffusion and Random walk model for cell migration	12
2.2.3 Angiogenesis	14
2.2.4 Limitations	15
2.2.5 History of the mechanobiological models.....	15
2.3 - Spine.....	17
2.3.1 Spine Anatomy	17
2.3.2 Intervertebral Disc Fusion Surgery.....	17
2.3.3 Intervertebral Disc Fusion Surgery Complications	20
2.2.4 Spinal Cages and mechano-regulatory model.....	22
2.4 - Lattice Materials	23
2.4.1 Mechanics of Lattice Materials	23
2.4.2 Tissue growth and Lattice Materials.....	27
CHAPTER 3 - MODEL DEVELOPMENT	32
3.1 - Introduction.....	32
3.1 - Boccaccio et al. Simplified Method	32
3.2 - Bone Growth Estimation Tool Development.....	34
3.2.1 Parametrized lattices	35
3.2.2 Poroelastic Materials.....	39

3.2.3 Mesh Preparation	40
3.2.4 Material Properties.....	43
3.2.5 Boundary Conditions	44
3.2.6 Data Collection	47
3.3 - Convergence Analysis	48
CHAPTER 4 - VALIDATION	52
4.1 - Boccaccio and Rodriguez's Trend Validation.....	52
4.1.1 Introduction	52
4.1.2 Method	52
4.1.3 Results	54
4.1.4 Discussion	56
4.2 - Canines In vivo Comparison (Octet versus Tetrahedron)	59
4.2.1 Introduction	59
4.2.2 Method	59
4.2.3 Results	61
4.2.4 Discussion	62
4.3 - Rabbits In Vivo Validation (Diamond, Cubic, Tetrahedron)	63
4.3.1 Introduction	63
4.3.2 Method	64
4.3.3 Results	65
4.3.4 Discussion	68
CHAPTER 5 - Bone Ingrowth in Ti6Al4V Lattice Topology	69
5.1 - Introduction.....	69
5.2 - Method	70
5.2.1 Pressure	71
5.2.2 Unit Cell Dimensions, Relative Densities, Topology and Element number.....	72
5.3 - Results	74
5.3.1 Bone Percentage Comparison Between Topologies	74
5.3.2 Fluid Flow Velocity and Octahedral Shear Strain Distribution	78
5.3.3 Distribution of Elements Optimal for Bone Ingrowth	78
5.4 - Discussion	82
5.4.1 Stiffness and Bone ingrowth	82
5.4.2 Fluid flow velocity.....	84
5.4.3 Lattice Response as Bone Grows	85
5.4.4 Comparison with previous study	85
5.5 - Limitation	86
CHAPTER 6 - CONCLUSION	88
6.1 - Summary.....	88
6.2 - Contribution	88
6.3 - Future Work.....	89
References	91
Appendix A - Structure Characteristic	100

Appendix B - Convergence Analysis.....104

Appendix C - In Silico Paper Replication105

Appendix D - In Vivo Paper Replication.....109

Appendix E - Titanium Lattice Characterization Data118

Appendix F - Result Plot of All Topology Evaluated126

Appendix G - Python Script for SpaceClaim Lattice CAD Creation.....155

Appendix H - ACT-Python Script for Material Assignment169

Appendix I - APDL Script for Boundary Conditions171

Appendix J - APDL Script for Post-Processing Solutions172

Appendix K - MATLAB Script for Biophysical Stimulus Calculation173

Appendix L - ACT-Python Script for Element Optimal for Bone179

List of Tables

Table 1. Tissue phenotype associated with each Biophysical Stimulus inequalities	13
Table 2. Surgical procedure's description of the LIF methods	19
Table 3. Lumbar spinal cages samples.....	21
Table 4 Comparison of unit cell topology mechanical properties	26
Table 5. Unit cell and lattice structure CAD prepared at a 15% relative density	38
Table 6. Material properties.....	43
Table 7. Implant lattice description	59
Table 8. Percentage of void space optimal for bone ingrowth with pressures of 0.5 MPa, 0.625 MPa, 0.75 MPa, 0.875 MPa and 1 MPa compared with the percentage of bone ingrowth found in structures with the same lattice parameter implanted in canine (<i>in vivo</i> data published by Arabnejad et al. 2016 [17]).....	62
Table 9. Implant lattice description	64
Table 10. Percentage of void space optimal for bone ingrowth with pressures of 0.4 MPa, 0.5 MPa, 0.6 MPa and 0.7 MPa, compared with the percentage of bone ingrowth found in structures with the same lattice parameter implanted in rabbits (<i>in vivo</i> data published by Deng et al. 2021 [16])	66
Table 11. Element with optimal environment for bone growth, under 0.5 MPa of pressure	67
Table 12. Lattice parameters evaluated for the characterization	70
Table 13. Lattice characteristics at the lowest and highest relative density investigated (5% and 50%) corresponding to pore sizes between 130 and 1016 micron.....	73
Table A1. Pore size estimation for 1200 microns unit cells, with 5 and 50% relative density	100
Table C1. Comparison of the strut diameter maximizing the percentage of the total volume that can stimulate bone based on the current simulation with data published by Rodriguez et al. 2018 [14] for the Diamond structure.....	105
Table C2. Comparison of the strut diameter maximizing the percentage of the total volume that can stimulate bone based on the current simulation with data published by Rodriguez et al. 2018 [14], for the Truncated Cube structure	105
Table C3. Comparison of the strut diameter maximizing the percentage of the total volume that can stimulate bone based on the current simulation with data published by Rodriguez et al. 2018 [14] for the Rhombic Dodecahedron with horizontal orientation structure	106
Table C4. Comparison of the strut diameter maximizing the percentage of the total volume that can stimulate bone based on the current simulation with data published by Rodriguez et al. 2018 [14], for the Rhombic Dodecahedron with vertical orientation structure.....	106

Table C5. Comparison of the strut diameter maximizing the percentage of the total volume that can stimulate bone based on the current simulation with data published by Boccaccio et al. 2018 [15], Rhombicuboctahedron structure made of polymer with different material young's modulus	108
Table D1. Element with optimal environment for bone growth, under 0.4 MPa of pressure	109
Table D2. Octahedral Shear Strain/0.0375, representing the strain part of the biophysical stimulus, under 0.4 MPa of pressure.....	110
Table D3. Octahedral Shear Strain/0.0375, representing the strain part of the biophysical stimulus, under 0.5 MPa of pressure.....	111
Table D4. Element with optimal environment for bone growth, under 0.6 MPa of pressure	112
Table D5. Octahedral Shear Strain/0.0375, representing the strain part of the biophysical stimulus, under 0.6 MPa of pressure.....	113
Table D6. Element with optimal environment for bone growth, under 0.7 MPa of pressure	114
Table D7. Octahedral Shear Strain/0.0375, representing the strain part of the biophysical stimulus, under 0.7 MPa of pressure.....	115
Table D8. Octahedral Shear Strain/0.0375 representing the strain part of the biophysical stimulus, and element with optimal environment for bone growth, under 0,5 MPa, 0.625 MPa, 0.75 MPa, 0.875 MPa and 1 MPa of pressure, for the Octet and Tetrahedron structures.	116
Table E1. Percentage of pores with optimal environment to grow bone and structure characteristics under pressures of 0.5 MPa, 1 MPa, 1.5 MPa and 2 MPa	121
Table E2. Biophysical stimulus average under pressures of 0.5 MPa, 1 MPa, 1.5 MPa and 2 MPa	124

List of Figures

Figure 1. Degeneration of IVD leading to compression of the nerve roots [3]	4
Figure 2. Bone structure diagram [18].....	8
Figure 3. The mesogenic differentiation process forming mature tissues, by Bruder, Fink, & Caplan, 1994 [24].....	9
Figure 4. The tissue differentiation model proposed by Prendergast et al. 1997 [4] based on distortional strain and interstitial fluid flow, illustration adapted by Liu & Niebur, 2008 [26].....	10
Figure 5. Functional spinal unit description and axial compression distribution between vertebral and facet joint, in the sagittal plane, by Kushchayev et al., 2018 [2]	17
Figure 6. Surgery pathway to insert a spinal cage implant during anterior lumbar interbody fusion (ALIF), lateral lumbar interbody fusion(LLIF), oblique lumbar interbody fusion(OLIF), posterior lumbar interbody fusion (PLIF) and transverse lumbar interbody fusion (TLIF) [70]	18
Figure 7. a) Sagittal CT scan showing non-union of a PEEK cage following LIF surgery, by Behrbalk et al 2013 [85], b) subsidence of an ALIF titanium cage in the anterior part of the endplate and c) healthy ALIF fusion following the insertion of a titanium cage, by Cabraja et al 2012 [86].....	20
Figure 8. Eleven polyhedron cells: (1) Square pyramid; (2) Triangular prism; (3) Cubic; (4) Hexagonal prism; (5) Octagonal prism; (6) Rhombi Dodecahedron; (7) Cuboctahedron; (8) Rhombicuboctahedron; (9) Truncated Cube; (10); Truncated Octahedron; (11) Truncated Cuboctahedron [103].....	24
Figure 9. Pores of Truncated Cuboctahedron unit cell, with a) 5% and b) 50% relative density	29
Figure 10. Octet unit cell with parameters definition.....	35
Figure 11. Strut cross section with adjustable dimensions, made from the	36
Figure 12. Two-phase model representing poroelastic material [145]	39
Figure 13. APDL script for coupled pore pressure, element activation.	41
Figure 14. Finite element mesh of an octet lattice, granulation tissue and plate, with element size of 92 microns, resulting in a total of 367,195 elements.	42
Figure 15. a) Octet unit cell b) 4X4X4 lattice structure made of octet unit cells with circle cross section at 15% relative density b) model setup, boundary conditions and constraints	45
Figure 16. Average octahedral shear strain and velocity within a 2x4x4 Octet lattice, under a pressure of 0.5 MPa, in function of the loading time	46
Figure 17. Percentage of void space, optimal for bone ingrowth within a 2x4x4 Octet, under a pressure of 0.5 MPa, in function of the loading time.	46

Figure 18. Element stimulus average within a 2x4x4 Diamond lattice with unit cells of 637 microns and strut diameter of 85 microns, made of polymer strut. Convergence analysis made in function of the number of elements used.	49
Figure 19. Percentage of void space, optimal for bone ingrowth within a 2x4x4 Diamond lattice with unit cells of 637 microns and strut diameter of 85 microns, made of polymer strut. Convergence analysis made in function of the number of elements used.....	49
Figure 20. Element stimulus average within a 2x4x4 Octet lattice, made of titanium strut. Convergence analysis made in function of the number of elements used. Strain/0.0375 and velocity/3 are the scaling factors added together to get the BS	50
Figure 21. Percentage of void space, optimal for bone ingrowth within a 2x4x4 Octet lattice, made of titanium strut. Convergence analysis made in function of the number of elements used	50
Figure 22. Comparison of the optimized strut diameter maximizing the percentage of the total volume that can stimulate bone based on the current simulation with data published by Rodriguez et al. 2018 [10], Bone ingrowth and diameter under pressures of 0.05 MPa, 0.1 MPa, 0.15 MPa, 0.5 MPa, 1 MPa and 1.5 MPa, inside a) Diamond and b) Truncated Cube structures made of polymer with circle and square cross section	54
Figure 23. Comparison of the optimized strut diameter maximizing the percentage of the total volume that can stimulate bone based on the current simulation with data published by Rodriguez et al. 2018 [10], Bone ingrowth and diameter under pressures of 0.05 MPa, 0.1 MPa, 0.15 MPa, 0.5 MPa, 1 MPa and 1.5 MPa, inside Rhombic Dodecahedron structure made of polymer with vertical (Rhombic (V)) and horizontal (Rhombic (H)) cell orientation	55
Figure 24. Comparison of the optimized strut diameter maximizing the percentage of the total volume that can stimulate bone based on the current simulation with data published.....	55
Figure 25. Percentage of the volume that can stimulate bone growth in function of the total and the void space volume inside a polymer structure made of 637 microns diamond unit cells with different strut diameter.....	57
Figure 26. Element with optimal environment for bone growth (brown elements) inside lattice structures made of Diamond unit cell with a strut diameter of a) 85 microns (biologically irrelevant solution) and b) 290 microns (biologically relevant solution), loaded with 0.5 MPa of axial compression.....	58
Figure 27. Implants inserted perpendicular to the lateral femoral cortex of the canines from Arabnejad et al., 2016 [17]	59
Figure 28. Transverse electron micrograph showing bone ingrowth after A) 4 weeks and B) 8 weeks, in 1) Octet truss and 2) Tetrahedron from Arabnejad et al., 2016 [17]	61

Figure 29. Percentage of void space, optimal for bone ingrowth with pressures of 0.5 MPa, 0.625 MPa, 0.75 MPa, 0.875 MPa and 1 MPa compared with the percentage of bone ingrowth found in structures with the same lattice parameter implanted in canine after 4 and 8 weeks (<i>in vivo</i> data published by Arabnejad et al. 2016 [17]), in red octet and blue tetrahedron structures.....	62
Figure 30. Close up view of the elements having an optimal environment to grow bone (brown elements) inside an octet lattice	63
Figure 31. Titanium lattice implanted in the lateral femoral condyle of a rabbit at low speed, from Deng et al., 2021 [16]	64
Figure 32. Percentage of void space optimal for bone ingrowth with pressures of 0.4 MPa, 0.5 MPa and 0.6 MPa, compared with the percentage of bone ingrowth found in structures with the same lattice parameter implanted in rabbits 6 and 12 weeks after surgery (<i>in vivo</i> data published by Deng et al. 2021 [16]) in red Diamond, in blue Tetrahedron and green Cube	66
Figure 33. Bone ingrowth forming a circular shape when growing in square pores a) and b) Optimal area for bone to growth (brown elements) within a cubic tessellated structure c) tissue formed in cell culture inside a square HA plates, after 30 days Rumpler et al., 2008 [137]	67
Figure 34. Unit cells	69
Figure 35. Preparation of the octet topology with 15% relative density a) SpaceClaim CAD model b) Load and boundary conditions c) Mesh generated for the model	70
Figure 36. Percentage of the void space having optimal stimulation for bone growth in function of the relative density of the 14 topologies, under a pressure of 0.5 MPa.....	74
Figure 37. Percentage of the void space having optimal stimulation for bone growth in function of the relative density of the 14 topologies, under a pressure of 1 MPa.....	75
Figure 38. Percentage of the void space having optimal stimulation for bone growth in function of the relative density of the 14 topologies, under a pressure of 1.5 MPa.....	75
Figure 39. Percentage of the void space having optimal stimulation for bone growth in function of the relative density of the 14 topologies, under a pressure of 2 MPa. The data point	76
Figure 40. Percentage of the void space having optimal stimulation for bone growth in function of the relative density of the 14 topologies, under pressures of 0.5 MPa, 1 MPa, 1.5 MPa and 2 MPa, for low (10%), medium (30%) and high relative density (50%)	77
Figure 41. Distribution of the velocity and the strain part of the biophysical stimulus, the Diamond sample at 40% relative density, under a pressure of 2 MPa	79
Figure 42. Distribution of the velocity and the strain part of the biophysical stimulus, the Cubic sample at 40% relative density, under a pressure of 2 MPa.....	80

Figure 43. Elements with optimal stimulus for bone growth, within a structure made of Cubic topology, with 10% and 40% relative density, under a pressure of 0.5 MPa and 2 MPa.....	81
Figure 44. Elements with optimal stimulus for bone growth, within a structure made of Rhombic Dodecahedron (Horizontal) topology, with 10% and 40% relative density, under a pressure of 0.5 MPa and 2 MPa	81
Figure 45. Elements with optimal stimulus for bone growth, within a structure made of X-Shape topology, with 10% and 40% relative density, under a pressure of 0.5 MPa and 2 MPa.....	82
Figure 46. Percentage of the void space having optimal stimulation for bone growth in function of the relative density for Rhombic Dodecahedron (V) in light green and Rhombic Dodecahedron (H) in dark green under pressures of 0.5 MPa, 1 MPa, 1.5 MPa and 2 MPa.....	84
Figure 47. Percentage of the void space having optimal stimulation for bone growth as a function of the relative density for Octet in light blue and Tetrahedron in pink under pressures of 0.5 MPa, 1 MPa, 1.5 MPa and 2 MPa.....	86
Figure B1. Element stimulus average within a 2x4x4 Diamond lattice with unit cells of 637 microns and strut diameter of 100 microns, made of pyroclastic polymer strut. Convergence analysis made in function of the number of elements used. Strain/0.0375 and velocity/3 are the scaling factors added together to get the BS.....	104
Figure B2. Bone ingrowth within a 2x4x4 Diamond lattice with unit cells of 637 microns and strut diameter of 100 microns, made of pyroclastic polymer strut. Convergence analysis made in function of the number of elements used	104
Figure C1. Comparison of the optimized strut diameter maximizing the percentage of the total volume that can stimulate bone based on the current simulation with data published by Rodriguez et al. 2018 [10], the structure made of polymer is tessellated with Diamond, Truncated Cube, Rhombic Dodecahedron (H) and Rhombic Dodecahedron (V) unit cells, with circle cross section, under pressures of 0.05 MPa, 0.1 MPa, 0.15 MPa, 0.5 MPa, 1 MPa and 1.5 MPa	107
Figure E1. Pore size of the topologies evaluated, with 5% and 50% relative density	118
Figure E2. Strut diameter of the topologies evaluated, with 5% and 50% relative density	118
Figure E3. Surface Area for a 2x2x4 lattice structure made of 1200 microns unit cells, with 5% and 50% relative density	119
Figure E4. Young's modulus (GPa) in the z direction with 5% relative density for different topologies	120
Figure E5. Young's modulus (GPa) in the z direction with 50% relative density for different topologies	120

Figure F1. Percentage of the void space that procures an optimal environment for bone growth (%) in function of the relative density, under a pressure of 0.5 MPa, 1 MPa, 1.5 MPa and 2 MPa, for the Octet topology	126
Figure F2. Percentage of the void space that procures an optimal environment for bone growth (%) in function of the relative density, under a pressure of 0.5 MPa, 1 MPa, 1.5 MPa and 2 MPa, for the Cubic Center topology	126
Figure F3. Percentage of the void space that procures an optimal environment for bone growth (%) in function of the relative density, under a pressure of 0.5 MPa, 1 MPa, 1.5 MPa and 2 MPa, for the Diamond topology	127
Figure F4. Percentage of the void space that procures an optimal environment for bone growth (%) in function of the relative density, under a pressure of 0.5 MPa, 1 MPa, 1.5 MPa and 2 MPa, for the Vintile topology	127
Figure F5. Percentage of the void space that procures an optimal environment for bone growth (%) in function of the relative density, under a pressure of 0.5 MPa, 1 MPa, 1.5 MPa and 2 MPa, for the Cubic topology	127
Figure F6. Percentage of the void space that procures an optimal environment for bone growth (%) in function of the relative density, under a pressure of 0.5 MPa, 1 MPa, 1.5 MPa and 2 MPa, for the Body Centered (X-Shape) topology	128
Figure F7. Percentage of the void space that procures an optimal environment for bone growth (%) in function of the relative density, under a pressure of 0.5 MPa, 1 MPa, 1.5 MPa and 2 MPa, for the Rhombic Dodecahedron (vertical) topology	128
Figure F8. Percentage of the void space that procures an optimal environment for bone growth (%) in function of the relative density, under a pressure of 0.5 MPa, 1 MPa, 1.5 MPa and 2 MPa, for the Rhombic Dodecahedron (Horizontal) topology	128
Figure F9. Percentage of the void space that procures an optimal environment for bone growth (%) in function of the relative density, under a pressure of 0.5 MPa, 1 MPa, 1.5 MPa and 2 MPa, for the Rhombicuboctahedron topology	129
Figure F10. Percentage of the void space that procures an optimal environment for bone growth (%) in function of the relative density, under a pressure of 0.5 MPa, 1 MPa, 1.5 MPa and 2 MPa, for the Tetrahedron topology	129
Figure F11. Percentage of the void space that procures an optimal environment for bone growth (%) in function of the relative density, under a pressure of 0.5 MPa, 1 MPa, 1.5 MPa and 2 MPa, for the Truncatedcuboctahedron topology	129

Figure F12. Percentage of the void space that procures an optimal environment for bone growth (%) in function of the relative density, under a pressure of 0.5 MPa, 1 MPa, 1.5 MPa and 2 MPa, for the Truncated Cube topology	130
Figure F13. Percentage of the void space that procures an optimal environment for bone growth (%) in function of the relative density, under a pressure of 0.5 MPa, 1 MPa, 1.5 MPa and 2 MPa, for the Tesseract topology	130
Figure F14. Percentage of the void space that procures an optimal environment for bone growth (%) in function of the relative density, under a pressure of 0.5 MPa, 1 MPa, 1.5 MPa and 2 MPa, for the X Cross Grid topology	130
Figure F15. Average biophysical stimulus for 10% relative density structures, under a pressure of 2 MPa	131
Figure F16. Average biophysical stimulus for 30% relative density structures, under a pressure of 2 MPa	131
Figure F17. Average biophysical stimulus for 50% relative density structures, under a pressure of 2 MPa	132
Figure F18. Elements with optimal stimulus for bone growth, within a structure made of Diamond topology, with 10% and 40% relative density, under a pressure of 0.5 MPa and 2 MPa	132
Figure F19. Distribution of the velocity and the strain part of the biophysical stimulus, the Octet sample at 40% relative density, under a pressure of 2 MPa	133
Figure F20. Elements with optimal stimulus for bone growth, within a structure made of Octet topology, with 10% and 40% relative density, under a pressure of 0.5 MPa and 2 MPa	134
Figure F21. Distribution of the velocity and the strain part of the biophysical stimulus, the Cubic Center (Star) sample at 40% relative density, under a pressure of 2 MPa	135
Figure F22. Elements with optimal stimulus for bone growth, within a structure made of Cubic Center (Star) topology, with 10% and 40% relative density, under a pressure of 0.5 MPa and 2 MPa	136
Figure F23. Distribution of the velocity and the strain part of the biophysical stimulus, the Vintile sample at 40% relative density, under a pressure of 2 MPa	137
Figure F24. Elements with optimal stimulus for bone growth, within a structure made of Vintile topology, with 10% and 40% relative density, under a pressure of 0.5 MPa and 2 MPa	138
Figure F25. Distribution of the velocity and the strain part of the biophysical stimulus, the X-Shape sample at 40% relative density, under a pressure of 2 MPa	139
Figure F26. Distribution of the velocity and the strain part of the biophysical stimulus, the Rhombic Dodecahedron (Horizontal) sample at 40% relative density, under a pressure of 2 MPa	140

Figure F27. Distribution of the velocity and the strain part of the biophysical stimulus, the Rhombic Dodecahedron (vertical) sample at 40% relative density, under a pressure of 2 MPa	141
Figure F28. Elements with optimal stimulus for bone growth, within a structure made of Rhombic Dodecahedron (vertical) topology, with 10% and 40% relative density, under a pressure of 0.5 MPa and 2 MPa.....	142
Figure F29. Distribution of the velocity and the strain part of the biophysical stimulus, the Rhombicuboctahedron sample at 40% relative density, under a pressure of 2 MPa	143
Figure F30. Distribution of the velocity and the strain part of the biophysical stimulus, the Tetrahedron sample at 40% relative density, under a pressure of 2 MPa.....	145
Figure F31. Elements with optimal stimulus for bone growth, within a structure made of Tetrahedron topology, with 10% and 40% relative density, under a pressure of 0.5 MPa and 2 MPa	146
Figure F32. Distribution of the velocity and the strain part of the biophysical stimulus, the Truncatedcuboctahedron sample at 40% relative density, under a pressure of 2 MPa.....	147
Figure F33. Elements with optimal stimulus for bone growth, within a structure made of Truncatedcuboctahedron topology, with 10% and 35% relative density, under a pressure of 0.5 MPa and 2 MPa.....	148
Figure F34. Distribution of the velocity and the strain part of the biophysical stimulus, the Truncated Cube sample at 35% relative density, under a pressure of 2 MPa	149
Figure F35. Elements with optimal stimulus for bone growth, within a structure made of Truncated Cube topology, with 10% and 40% relative density, under a pressure of 0.5 MPa and 2 MPa	150
Figure F36. Distribution of the velocity and the strain part of the biophysical stimulus, the X-Cross-Grid sample at 40% relative density, under a pressure of 2 MPa.....	151
Figure F37. Elements with optimal stimulus for bone growth, within a structure made of X-Cross-Grid topology, with 10% and 40% relative density, under a pressure of 0.5 MPa and 2 MPa	152
Figure F38. Distribution of the velocity and the strain part of the biophysical stimulus, the Tesseract sample at 40% relative density, under a pressure of 2 MPa.....	153
Figure F39. Elements with optimal stimulus for bone growth, within a structure made of Tesseract topology, with 10% and 40% relative density, under a pressure of 0.5 MPa and 2 MPa.....	154

Nomenclature

Symbol	Description
Φ	Porosity
$\varepsilon_1, \varepsilon_2, \varepsilon_3$	Principal strains
ρ	Lattice structure volume
ρ_o	Total volume of the lattice material
$\bar{\rho}$	Relative density
α	Radius of specimen
c	Concentration of MSC
δ_{ij}	Second order identity tensor
D	Diffusion coefficient
ε_v^e	Trace of infinitesimal strain tensor of solid matrix
F_i	Body force per unit volume
k	Cell death removal rate
k_m	Biot modulus
q_i	Rate of volumetric flow
T_{ij}	Biot stress tensor
v	Fluid velocity
y	Octahedral shear strain

List of Abbreviations

Abbreviation	Description
3D	Three Dimensional
ACT	ANSYS Customization Toolkit
ALIF	Anterior Lumbar Intervertebral Fusion
APDL	ANSYS Parametric Design Language
ATP	Adenosine Triphosphate
BDL	Bending Dominated Lattice
BMPs	Bone Morpho-genetic Proteins
BS	Biophysical Stimulus
CT	Computed Tomography
CPT	Coupled Pore-Pressure-Thermal
DNA	Deoxyribonucleic Acid
FEA	Finite Element Analysis
IFGS	Insulin-like Growth Factors
IVD	Intervertebral Disc
LIF	Lumbar Interbody Fusion
LLIF	Lateral Lumbar Intervertebral Fusion
MSC	Mesenchymal Stem Cells
OLIF	Oblique Lumbar Intervertebral Fusion
PEEK	Polyether Ether Ketone
PLIF	Posterior Lumbar Intervertebral Fusion
RAM	Random-Access Memory

RD	Relative Density
RVE	Representative Volume Elements
SDL	Stretch Dominated Lattice
SEM	Scanning Electron Microscopy
TGFs	Transforming Growth Factors
TLIF	Transverse Lumbar Intervertebral Fusion
TPMS	Triply Periodic Minimal Surface
USDFLD	User-Defined Field Variable
VEGF	Vascular Endothelial Growth Factors

CHAPTER 1 - INTRODUCTION

1.1 - Motivation

One of the leading causes for low back pain is the compression of nerve roots and spinal cord, due to degeneration of intervertebral discs (IVD) [2] (Figure 1). Degeneration of IVD mostly happens in the lumbar region of the spine. Lumbar intervertebral disc fusion (LIF) is a surgical treatment to relieve pain. It consists of replacing the IVD with a spinal fusion cage, which encourages osseointegration and finally complete fusion of the adjacent vertebrae. Some patients still experience poor post-surgery outcome, either from cage subsiding into the endplate or due to a lack of bone growth around the spinal cage. Besides biochemical stimulation of bone growth, fusion cages should be designed to optimize the biomechanical environment to avoid subsidence and maximize osseointegration.

Figure 1. Degeneration of IVD leading to compression of the nerve roots [3]

The advancement of 3D printing has allowed the development of lattice-based implants. This emerging technology provides the opportunity to design micro architectures that can maximize both the mechanical and the biological performance of orthopaedic devices. This leads to the main idea of creating a lattice-based spinal cage that stimulates bone ingrowth. To

determine the best lattice structure for the implant, the field of mechano-regulatory modelling was explored in this thesis. Numerous mechano-regulatory studies using FEA have been previously developed and supported, to simulate bone growth based on the strain and shear flow velocity osteoblasts and osteocyte would encounter in the growth space [4]–[8]. Due to excessive computational cost, most research apply those models to study one specific geometry or implant. Until now, only one research team has simplified the mechano-regulatory algorithm to allow the evaluation of numerous lattice structures and optimized certain of their parameters. [9] However, they have focused on investigating polymer lattices and modeled only a few geometries. [9]–[15].

The goal of this thesis was to apply Boccaccio et al. [9], simplified method to compare the performance, relevant for osseointegration, of various lattice structure made of Ti6Al4V, subjected to compression loads observed on spinal cages. Characterization was done to determine the relative densities (RD) maximizing the amount of pore volume offering an optimal biophysical stimulus to sustain bone growth. The characterization was performed for fourteen different unit cell topologies. The results obtained from this research are not only valuable for the development of lumbar intervertebral implant but can also be beneficial for the improvement of numerous other Ti6Al4V prosthesis devices implanted in other regions of our body.

1.2 - Research Objectives

The general objective of this thesis is to define optimal lattice parameters to maximize bone ingrowth stimulation for spinal fusion cages, using a simplified mechano-regulatory models. Three main research objectives had to be reached individually to accomplish the main goal.

1. Develop an automated tool, based on Boccaccio et al. [9] mechano-regulatory simplified model, that can evaluate the capacity of numerous structures to stimulate bone formation, without having the burden of major computation time.

2. Evaluate the bone estimation capacity of the proposed tool; firstly, by comparing the resulting trends to data from Rodriguez et al. 2018 [10] and Boccaccio et al. 2018 [13] secondly, by comparing the trends obtain to *in vivo* studies that implanted different 3D printed lattices made of Ti6Al4V in rabbits [16] and canines [17].
3. Investigate the bone ingrowth stimulation potential of fourteen different topologies, under loads replicating the spinal pressure that would be applied on a spinal implant (0.5 MPa, 1 MPa, 1.5 MPa and 2 MPa). Creating a database for each topology including the percentage of pores optimal to grow bone within a tessellated structure of 5%, 10%, 15%, 20%, 25%, 30%, 35%, 40%, 45%, 50% relative density.

1.3 - Organization of the Thesis

This thesis is organized as follows: Chapter 2 is a summary of literature review covering bone structure, mechano-regulatory theory, spinal implants, and an introduction to lattice structures. Chapter 3 explains the development of the model created to determine which element procures the right environment for bone to be stimulated. In Chapter 4, two *in silico* studies are replicated to validate the model. Furthermore, two published *in vivo* studies were also simulated with the model to gage the reliability of the trend found. Eventually, in Chapter 5 a database is created to describe the performance of different Ti6Al4V lattice structures in function of their relative density. Finally, Chapter 6 concludes with a short overview of potential future work.

CHAPTER 2 - LITERATURE REVIEW

2.1 - Bone Tissue

2.1.1 Bone structure

Bone is a connective tissue composed of 20% water, 35% organic molecules including collagen, morphogenetic protein and fibrillin as well as 45% inorganic mineral salts, including calcium phosphate, calcium carbonate, sodium, magnesium and fluoride [18]. Bone can have two different macrostructures (Figure 2); a compact arrangement also referred as cortical bone and a trabecular arrangement also referred as spongy bone. The compact bone is a dense structure made of osteons all aligned in the same direction. It has a porosity of 3 to 5%, a compressive strength between 100 and 230 MPa and a Young's modulus between 3 and 30 GPa. Inside the osteons there are small blood vessels separated by 200 to 300 microns that provide nutrients and oxygen. As for the trabecular bone, it is structured by load optimized lattice structure. It has a porosity varying from 50% to 90%, a compressive strength between 2 and 12 MPa and a Young's modulus between 0.02 to 0.2 GPa. [19], [20] The porosity typically increases as we get older. Trabecular bone is also filled with bone marrow containing mesenchymal stem cells (MSC). Those cells are nonhematopoietic progenitor cells, meaning they can differentiate into specialized cells that are not implicated in the production of blood. They are usually in a quiescent phase (inactive), until they receive external chemical and physical stimulation to activate them for repair purposes.

It is important to acknowledge that the human body is a fluidic environment. The extracellular fluids consist of the interstitial fluid and blood plasma where the cells, the extracellular matrix and the blood vessels bathe. The bone consists of both a solid and fluid phase. In cortical bone, when there is a load, the fluid flows within the canaliculi (bone canals)

which simulates osteocytes [21]. However, there are also osteoblasts at the outside of the trabeculae which likely get stimulated by the pressure/ flow in the bone marrow or granular tissue during osseointegration. The mechanical stimulation will induce fluid shear, simulating gene expression which ultimately result in mineralisation. The flow is also important for the cell distribution [22]

Figure 2. Bone structure diagram [18]

2.1.2 Bone Remodeling and Regeneration

Bone is constantly exposed to a cycle called bone remodeling. In this process, cells called osteoclasts resorb mineralized bone and cells called osteoblast fill the new cavities with new bone. In the lacunae (bone cavity), osteocytes are present (Figure 2. Microstructure). Osteocytes are mechano-sensors, that sense the mechanical stimulus and transfers instructions to osteoblasts and other osteocytes. When osteoclasts absorb bone, it reduces the strength and stiffness of the bone structure, in return there is an increase of stimulation sensed by the osteocyte. Consequently, they transduce this signal to activate bone deposition. This bone remodeling cycle continuously takes place to repair the tissue and to adjust the bone structure in function of the

body environmental changes. This bone adaptation equilibrium to the mechanical stresses is also known as Wolff's law [23].

Figure 3. The mesogenic differentiation process forming mature tissues, by Bruder, Fink, & Caplan, 1994 [24].

Counter to bone remodeling, bone regeneration implicates the proliferation, differentiation, and migration of MSC, for the creation of new tissues (Figure 3). Migration first occurs when the MSC are activated and start colonizing a new area. Then, they undergo proliferation which is the cell division augmenting the MSC colony. Finally, depending on the mechanical stimulation cells receive, they change their characteristic and become specialized cells forming mature tissues. The measurement of the mechanical stimulus they sense is referred as the biophysical stimulus (BS). For bone tissue to grow the stimulation must be low but above a certain threshold. When the BS is high, it favours fibroblast differentiation, when the stimulus is intermediate it favours chondrocyte differentiation and low stimulus favour osteoblast differentiation (Figure 3). In regions that experience too high stimulus there is cell apoptosis (cell

death); those regions usually have lower cell concentration [6]. Regardless of the stimulation, if there is low concentration of oxygen, the cell differentiates into cartilage. Cartilage is an avascular low-density tissue, composed mostly of water and collagen. Later, if the environmental condition changes, they can take the endochondral ossification pathway to become bone. Fibrous tissue is made of a network of collagen fibre with fibroblast cells, this dense tissue can also change into bone through the desmoid ossification process [25]. The bone regeneration concept explained above, can also be applied to bone growth around or within an implanted structure. Tissue differentiation is significantly dependent on the load applied to the structure.

2.2 - Mechanobiological Model (FEA)

2.2.1 Biophysical Stimulus

A mechano-regulatory model was developed in 1997 by Prendergast et al., [4] (Figure 4). Since then, it has been used extensively.

Figure 4. The tissue differentiation model proposed by Prendergast et al. 1997 [4] based on distortional strain and interstitial fluid flow, illustration adapted by Liu & Niebur, 2008 [26]

This research team found that the biophysical stimulus that determines the fate of the cells can be estimated with Equation (1) based on fluid flow velocity and octahedral shear strain, where y is octahedral shear strain and v is fluid velocity, a and b are constants. The constant a is

0.0375 and b is $3 \mu\text{m/s}$ according to Prendergast et al [4], [6], [15], [26]–[37]. The octahedral shear strain is calculated using the three principal strains ε_1 , ε_2 , ε_3 , in Equation (2) [14], [38]–[42]. To develop the Equation (1), a micromotion device was inserted in the condyles of dogs to analyze the formation of tissue on its surface. A biphasic finite element model of the implant was created to further investigate the tissue differentiation results. After compiling the experimental and FEA data, they could describe the mechano-regulatory pathway with a plot based on the interaction between stimuli, shear strain and velocity. From this analysis they derived the constants a and b [4], [34].

$$BS = \frac{y}{a} + \frac{v}{b} \quad (1)$$

$$y_{\text{oct}} = \frac{2}{3} \sqrt{(\varepsilon_1 - \varepsilon_2)^2 + (\varepsilon_2 - \varepsilon_3)^2 + (\varepsilon_3 - \varepsilon_1)^2} \quad (2)$$

The octahedral shear strain is used since the tissue development is related to changes in cells shape and local extracellular matrix deformation [21]. Osteocyte senses mechanical strains, then the mechano-transducer of the cells will convert this signal to biological output. When osteoblasts and osteocytes sense mechanical strain and fluid flow shear stress, they release nitric oxide, which inhibits bone absorption and stimulates bone deposition [43]. In cortical bone itself, canalicular fluid flow is caused by mechanical loading of the bone. Fluid flow also protects the osteocytes against apoptosis, activates gene transcription and translation and is implicated in the dendrite elongation. Other important molecules for bone health maintenance are released when the cell sense fluid flow following the mechanical loading [43]. If the strain is too high there is a decrease in the osteoblast formation and if the fluid flow is too high, it will lead to bone resorption which can potentially cause failure of an implant [8].

To apply those concepts with FEA, a component is modelled to represent the space where bone is expected to grow, most often it is the void space around the structure of interest.

However, typically granulation tissue invades this space quickly long before bone is deposited.

Granular tissue, a vascularised tissue, can be modelled as a porous media, a solid matrix bathing in saline water [44]–[46].

2.2.2 Diffusion and Random walk model for cell migration

Diffusion or random walk models are used to simulate cell migration in models based on Pendergast et al., 1997 [4]. In the diffusion method, at the beginning of the FEA simulation, MSC cell concentration assumed inside the structure, where bone is expected to grow is set to 0 and the concentration outside is set to 1. Linear diffusion equations are applied to simulate the diffusion of the cells inside the structure. The method uses the diffusion Equation (3) with c the concentration of MSC, and D the diffusion coefficient [47]. Some studies add a term to represent the cell mitosis and a term for the death of cells, k being the death removal rate constant (Equation 4) [5], [7].

$$\frac{\partial c}{\partial t} = D \nabla^2 c \quad (3)$$

$$\frac{\partial n}{\partial t} = D \nabla^2 n + ns(c) - kn \quad (4)$$

The cell diffusion coefficient used is chosen with the assumption that after 4 weeks there is uniform MSC cell density in the entire structure. The biophysical stimulus is calculated for each element containing MSC to then determine the tissue phenotype between fibrous tissue, bone and cartilage according to the predefined ranges (Table 1) [4], [10], [11], [26], [29], [31], [36], [37], [47], [48]. The strains and velocity values from Figure 3, found experimentally by Prendergast et al [4], were used in the BS equation to define those tissue phenotype ranges (Table

1). The ranges are based on dog bone response and do not consider intraspecies and interspecies genetic variability. [29], [31], [49]–[51] For every loading cycle, once the MSC cell concentration is found, they update the material properties of each element according to the tissue type corresponding to the stimulus [6], [27], [36], [52], [53]. The tissue material properties associated with each element is averaged on the ten previous iterations, to simulate a gradual cell differentiation with time [6], [53]. Most paper that included the cell diffusion concept would use ABAQUS with the subroutine USDFLD to have a solution-dependent field variable at element integration, this program facilitates solution dependent material properties. [33]. Iteration keeps going for a defined period or once the granulation tissue is fully converted to specific tissue phenotype.

Table 1. Tissue phenotype associated with each Biophysical Stimulus inequalities

Biophysical Stimulus (BS) ranges	Tissue phenotype
$BS > 3$	Fibroblast (Fibrous tissue)
$1 < BS < 3$	Chondrocytes (Cartilage tissue)
$0.53 < BS < 1$	Osteoblasts (Immature bone tissue)
$0.01 < BS < 0.53$	Osteoblasts (Mature bone tissue)
$0 < BS < 0.01$	Bone resorption (Bone resorption)

The random walk model is an alternative to the diffusion models, which is another way to estimate the proliferation and migration of the cells inside a structure. There is no clear evidence if the diffusion or random walk model provide better estimate, both showed comparable results. [5], [6]. The random walk model consists of the assumption that each cell is surrounded by six possible positions that a daughter cell can occupy. A random position is selected for the daughter cell to lodge, out of the 6 possible sites. If one of the 6 positions is already occupied, another random site is chosen. This proliferation process continues until the entire lattice is filled with cells. The fibroblasts are known to be more motile than the other cell types, thus a migration

model is also applied to those cells. This migration process represents the displacement of the cell. It works the same way as the proliferation but instead of creating a daughter cell, it displaces the cell randomly between the 6 surrounding sites and its current position. [6], [28] A study comparing both models showed that the random-walk model gives similar results compared with the diffusion method. However, the tissue type created was varying, depending on the method used. The random walk is not cell phenotype specific, since it assumes that the 3 types of cells proliferate at the same rate, leading to an heterogenous tissue distribution. [28]

The diffusion and the random walk model are important for the distribution of cells in the lattice. A study implanted porous structures injected with an autologous serum containing MSC, in sheep. The coverage of the serum was shown to significantly affect the bone formation results. According to Bensaïd et al., 2005 [54], lattice material with layers of MSC cells surrounding the lattice forms significantly more bone compared to lattice with MSC cells homogeneously distributed inside the lattice at the starting point. This study used the same number of cells in both *in vitro* tests. The result of that study was explained by the fact that the cell at the periphery is more in contact with the healthy bone and has more access to vascularization.

2.2.3 Angiogenesis

Vascularization also plays a role in the cell's differentiation. For instance, in regions of poor vascularization, the cells tend to become chondrocyte rather than osteoblast. There are ways to take the vascular endothelial growth factor (VEGF) concentration in consideration during bone growth simulation [55]. In those models, the cells differentiate depending on both the biophysical stimulus and the oxygen availability. High biophysical stimulus can also prevent or slow down capillary vessels development [32]. In simulations, most researchers relate the blood vessel branching to the vessel length. In the human body, the vessel branching is mostly regulated by

the presence of the growth factors. While the inclusion of angiogenesis in the simulation brings the result closer to the human response, it adds complexity and computation time to the modelling.

2.2.4 Limitations

The mechano-regulatory models rely purely on mechanical stimulation to simulate bone growth, which is a tremendous simplification of very complex biophysical and biochemical processes. For example, growth factors as well as the accessibility of nutrients and oxygen play a massive role in bone growth and currently, they are not possible to consider in *in silico* studies. It was also shown, *in vivo* that the cells response to mechanical stimulus can have slight variations between subjects [48], [56], [57]. Those variations are found in the tissue phenotype inequalities that are predefined in Table 1. Furthermore, the path cells take during migration is estimated with the same diffusion equation, whereas each cell type migrates and proliferate differently. Additionally, depending on the location of the implantation, the concentration of MSCs in the surrounding tissues can vary. For porous structures, the quantity of blood that infiltrates the granulation space can potentially create small blood clots in the pores, which can alter the cell diffusion.

2.2.5 History of the mechanobiological models

(1941) - The first mechano-regulation theory was issued based on observation in fracture callus, by Pauwels in 1941 and 1960 [7], [28], [58]. He stated that the first and second invariant of the strain tensor could describe the volume and shape change in tissues, due to hydrostatic stress and shear stress, respectively. He then suggested that the tissue phenotype could be determined by the combination of this hydrostatic stress and shear stress.

(1980) - Perren and Cordey 1980, hypothesized that the interfragmentary strain ratio in a fracture gap is related to the tissue differentiation. A strain under 2% would lead to bone growth, between 2 and 10% lead to cartilage, between 10 and 100% lead to fibrous tissue and a strain more than 100% prevent tissue growth [59], [60]. However, this theory is an oversimplification of the mechanical cellular response.

(1988 – 1998) - Carter et al., 1988 [61] also suggested a mechano-regulation model based on the hydrostatic stress and octahedral shear stress in the tissues. In 1998, Carter et al [62] replaced the octahedral shear stress by the octahedral strain considering the maximum principal. This modification was made following their study results, showing that cell elongation and cell shape change influence the differentiation. They also found that higher distortion of the chondrocyte and fibroblast increases the synthesis of fibrous matrix, mitosis, and proliferation.

(1997) - In 1997 Prendergast et al., developed the model described and studied in this thesis, based on octahedral shear strain and fluid velocity [4]. This model developed an equation to quantify the mechano-regulation stimulation based on the strain, velocity, and stimuli. Till this day, this method remains the one giving the closest result to experiments [63], [64]. In 2009, Heyward & Morgan [65] corroborated experimentally Prendergast et al [4], model and confirmed that strain and fluid flow are the key tissue differentiation regulators.

(1998) - Cleas & Heigele [66], established a mechano-regulatory model that relies on the hydrostatic stress and the minimum principal strain applied on the tissues. When strain is less than 5% and hydrostatic pressure is less than 0.15 MPa, bone will form. If strain is less than 15% and hydrostatic pressure is more than 0.15 MPa, cartilage will form. When the strain or the hydrostatic pressure is higher than 15% and 0.15 MPa, fibrous tissue is formed. This model leads to comparable result to the Prendergast et al [67] model.

2.3 - Spine

2.3.1 Spine Anatomy

The spine is composed of 24 vertebrae including 7 cervical, 12 thoracic and 5 lumbar. Between each vertebra, there is an intervertebral disc (IVD) (Figure 5) [2]. The IVD centre is called the nucleus pulposus and is composed of water with some collagen and proteoglycans. Healthy nucleus pulposus are well hydrated and generate intradiscal pressure that resist compression. On the periphery there is the annulus fibrosis, made of proteoglycans, fibroblasts, collagen, and elastic fibres, structured in the form of 15 to 25 concentric lamellae. Both structures easily absorb water since proteoglycan is a charged molecule that binds well with water, a polar molecule. The spine is the intermediary between the upper and lower body, it acts as a shock absorber. The combination of the 3 main curvatures of the spine and the IVDs helps absorbing and transferring energy from axial loads.

Figure 5. Functional spinal unit description and axial compression distribution between vertebral and facet joint, in the sagittal plane, by Kushchayev et al., 2018 [2]

2.3.2 Intervertebral Disc Fusion Surgery

The following section is an excerpt from a review paper entitled "Subsidence of spinal fusion cages: a systematic review" which will be submitted for publication [1].

Different pathologies such as sciatica, disc prolapse, nucleus pulposus herniation, spinal stenosis, spondylolisthesis, scoliosis amongst others can require the removal of the IVD disc followed by the fusion of the adjacent vertebrae. Most often these pathologies are the result of IVD degeneration. The degeneration is caused by changes in permeability and water content of the annulus and nucleus pulposus, causing a decrease in disc space [68].

When the degeneration occurs between the lumbar vertebrae, surgery is performed to remove the IVD and replace it with a spinal cage. Eventually bone grows around the implant and a full fusion is created between the inferior and superior endplates. The surgeons may choose between five different approaches to perform the vertebral fusion, namely, Anterior Lumbar Fusion (ALIF), Lateral Lumbar Fusion (LLIF), Posterior Lumbar Fusion (PLIF), Oblique Lumbar Fusion (OLIF) and Transverse Lumbar Fusion (TLIF) (Figure 6). In addition to the implantation of a cage, surgeons may decide to add pedicle screws, to give additional stability. Those pedicle screws are often used when the cage is smaller. Their presence can reduce the pressure the implant sustains. [69]

Figure 6. Surgery pathway to insert a spinal cage implant during anterior lumbar interbody fusion (ALIF), lateral lumbar interbody fusion (LLIF), oblique lumbar interbody fusion (OLIF), posterior lumbar interbody fusion (PLIF) and transverse lumbar interbody fusion (TLIF) [70]

Table 2. Surgical procedure's description of the LIF methods

Surgical procedure	Description
ALIF	The ALIF technique consists of an anterior retroperitoneal approach that provides access to the disc, with the patient in supine position. It involves a three- to five-inch midline and paramedian incision, the opening of the longitudinal ligament and an important vascular mobilization, leaving the paraspinal muscles intact [70]–[72]. This approach allows the insertion of a cage that covers the entire end plate surface, including the apophyseal ring which stabilizes the motion segments which may not require adjunct pedicle screws.
LLIF	LLIF is performed by a lateral retroperitoneal incision on a laterally positioned patient. This creates a transpsoas corridor to access the disc space and insert the implant [70]. Like the cages used in ALIF, the LLIF implant is placed medially and has a large footprint covering parts of the apophyseal ring.
OLIF	The OLIF surgery requires patients to be positioned on their side. It involves a lateral and paramedian incision between the peritoneum and the psoas muscle to access the disc space.[70] A smaller implant than for ALIF and LLIF is inserted, covering the interior one-third portion of the endplate. It will rarely cover the apophyseal ring and mostly be used in conjunction with posterior fixations.
PLIF	PLIF, one of the first procedures used for IVD fusion surgery, accesses the IVD space from a posterior direction, with the patient in a prone position. A midline incision dissecting bilateral muscle strip or splitting paramedian muscle is performed. Before inserting the cage, a laminectomy and a partial facetectomy are performed to navigate around nerve roots [73], [74]. Depending on the cage design, one or two cages is inserted within the apophyseal ring [70], [72]–[74].
TLIF	The transforaminal TLIF provides access to the intervertebral space directly through a small unilateral incision on one side of the neural foramen while the patient is in prone position, minimizing nerve manipulation [70]. This may involve extensive muscle retraction and dissection with the removal of the facet joint in order to place a straight or curved cage. TLIF cages have significantly smaller footprint coverage than ALIF and LLIF cages [72], [75], [76]. Depending on the cage used and the surgeon's approach, the TLIF cage can either be placed on the interior or medial part of the end plate. TLIF does not provide enough segmental stability without the use of posterior fixation.

2.3.3 Intervertebral Disc Fusion Surgery Complications

Subsidence

While these surgical approaches are effective treatment options for IVD disease, they each have their benefits and potential risks. All can lead to cage subsidence: a significant loss of disc space occurring when the implant migrates into the vertebral bodies (Figure 7b). To this day, there is still subsidence reported in patient receiving LIF surgery, regardless of the surgery method. Complications resulting from cage subsidence can vary from loss of disc height and lumbar lordosis to the narrowing of the intervertebral foramen and foraminal stenosis. A change in the lordosis will hinder the sagittal balance and kinematics of the spine and may lead to back pain [77]–[79]. If the lordosis angle and height are not properly corrected, spinal nerve decompression and the strength of the fusion are compromised [80]. Subsidence can also jeopardize the alignment of the spine during fusion and lead to cyst formation [78], [81], [82]. All those complications can potentially result in non-union, adjacent level degeneration, and/or cause recurrent symptoms requiring revision surgery [83], [84].

Figure 7. a) Sagittal CT scan showing non-union of a PEEK cage following LIF surgery, by Behrbalk et al 2013 [85], b) subsidence of an ALIF titanium cage in the anterior part of the endplate and c) healthy ALIF fusion following the insertion of a titanium cage, by Cabraja et al 2012 [86]

The main key factor affecting cage subsidence is the cage size. Bigger cage surface area in contact with the endplate resulted in less subsidence since it reduces endplate stress [87]–[90]. For instance, both ALIF without pedicle screw and LLIF with pedicle screw are showing less

subsidence compared with methods using smaller cages such as TLIF and PLIF. See Table 3, for cages example with dimensions, for each surgery procedures. Another crucial aspect is the placement of the implant, since the endplate is weaker in the centre [87]. Studies evaluating different implants, on cadaveric spinal units, during cyclic compressive loads, suggested that regardless of the cage type, placing the implant on the peripheral subchondral bone is one of the most effective methods to prevent the cage from penetrating the vertebral body interphase [91], [92].

Non-union

Following the LIF surgery, non-union consisting of an incomplete fusion between the two adjacent vertebrae may occur following all surgery methods (Figure 7a) [85]. When the bone has trouble growing around the spinal cage, the two vertebrae cannot merge and fuse completely, leading to failure and requires revision surgery [70], [85], [89]. Clinically unstable fixations allow more mobility, including higher shear displacement which may disturb the healing process and can lead to this non-union [93]. There is the possibility of filling the implants holes with a sponge soaked with recombinant human bone morpho-genetic protein-2 (BMP-2) to enhance the fusion. Additionally, the design of an implant that creates an optimal biomechanical environment for cells to differentiate in mature bone may accelerate the healing.

Table 3. Lumbar spinal cages samples

Spinal cage	Surgery Procedure	Cage size	
CAPSTONE@ PEEK Spinal System -Medtronic [94]	LLIF, PLIF and TLIF	Smallest: 22mm X 10mm Largest: 36mm X 10mm
BIO AVS UniLIF - PEEK – Stryker [95]	LLIF, PLIF and TLIF	Smallest: 25mm X 9mm Largest: 30mm X 11mm

Aleutian Anterior-Lumbar Interbody System -PEEK - Stryker [96]	ALIF	Smallest: 30mm X 24mm Largest: 36mm X 28mm
TITAN ENDOSKELETON TA – Titanium – Medtronic [97]	ALIF and OLIF	Standard: 32mm x21mm Large: 36mm x 24mm X-Large: 40mm x 27 mm
Aleutian Lateral Interbody System – PEEK -Stryke [98]	LLIF	Smallest: 45mm X 18mm Largest: 60mm X 22mm

2.2.4 Spinal Cages and mechano-regulatory model

There are few studies that have investigated spinal cage designs based on mechano-regulatory theories. Postigo et al 2014 [67] compared the fusion performance of a LIF solid spinal cage to a cage with an extrusion in the middle, under 500 N, 1000 N and 1500 N. They also compared a lower Young's modulus of 1 GPa to a higher Young's modulus of 100 GPa. They prepared two different FEA models to collect their data. The first relying on Prendergast et al., (1997)[4] models, consisting of calculating the BS value with the octahedral shear strain and fluid-flow velocity. The second one was inspired by Claes & Heigele's fracture healing studies from 1998 [66], where they calculated the BS value from the minimum principal strain and the hydrostatic stress. Both methods gave similar results. They were able to show that there was a better fusion around materials with lower stiffness implant. Whether a study is based on the fluid velocity model or the fluid shear stress model, the BS should not vary too much, if the respective constant α is 3 $\mu\text{m/s}$ for Prendergast's model and 10 $\text{mPa}\cdot\text{s}$ for Claes' model. Both measurements will lead to the same trends since the fluid shear stress is linearly proportional to the velocity gradient, in Newtonian fluids. In those mechano-regulation studies the fluid is always modelled as Newtonian fluids, respecting Newton's law of viscosity, where the viscosity is not depending

on shear rate [99]. As for the strain measurement, octahedral shear strain, it is calculated based on the principal strains ϵ_1 , ϵ_2 , ϵ_3 , while Cleas' model uses the minimum principal strain.

Bashkuev et al., 2015 simulated bone ingrowth around different solid spinal fusion cage geometry such as hollow, small, large, conic, T-shape and ring cages with different materials [40]. They also used the minimum principal strain and the hydrostatic stress to calculate the BS. After loading their implant under compressive loads of 250 N, 500 N and 1000 N, they found that complex cage geometry such as a cone and a T-shape implant were producing more bone, than a simple cubic implant.

2.4 - Lattice Materials

2.4.1 Mechanics of Lattice Materials

Micro-architected periodic cellular solids, also known as lattice materials, can be multiscale designed to tailor its stiffness, strength, density, and permeability, among others. These micro structures are able to withstand higher strains making them better at absorbing kinetic energy [100]. In mechanical design, research often focus on developing lightweight structures with the optimal specific stiffness and strength properties. On the other hand, lattice structures developed for medical applications are tailored for more compliant performance. When the implants are made of titanium, lattice structures can help to reduce the stiffness while still providing the strength necessary to resist loading [19].

Lattice Description

Strut based or truss-like lattice micro-structures are a non-stochastic form of cellular solids. Here, a unit cell, the Representative Volume Element (RVE), is tessellated periodically to create the lattice material. Different topologies of lattice materials are available in the literature

spanning 2D planar and 3D spatial topologies [101]. Some spatial lattices are tessellated with only one fundamental polyhedron (Figure 8) and others are made by combining different polyhedron. For example, an octet unit cell is made of an octahedron with eight tetrahedron connected on its faces. Different arrangement of struts can also modify the polyhedron to create unit cells with different material properties (Table 4). other types of lattice materials include lattices with closed unit cell such as the triply periodic minimal surface (TPMS). They are made of connected curved sheets without nodes. Those TPMS cellular material have almost no curvature discontinuity, which reduces the stress concentration, consequently increasing the fatigue strength. [102]

Mechanical performance of lattice materials is evaluated at multiscale. Homogenized material properties are considered as the mechanical response of the lattice at the macroscopic scale, while lattice micro-structural characteristics are evaluated at the microscopic scale. Homogenized mechanical properties of lattice materials vary with the cell topology, the parent material properties, and the relative density.

Figure 8. Eleven polyhedron cells: (1) Square pyramid; (2) Triangular prism; (3) Cubic; (4) Hexagonal prism; (5) Octagonal prism; (6) Rhombi Dodecahedron; (7) Cuboctahedron; (8) Rhombicuboctahedron; (9) Truncated Cube; (10); Truncated Octahedron; (11) Truncated Cuboctahedron [103].

The relative density $\bar{\rho}$ of the lattice is one of the most important parameters. It is also expressed in terms of porosity Φ , in most biological studies. Relative density is defined by the lattice structure volume ρ divided by the total volume of the lattice material ρ_0 . While the apparent porosity Φ is one minus the relative density. The lower the relative density the more the architecture impacts the mechanical properties. [104] Comparisons between unit cell topologies are more precise when they are done at the same relative density.

The lattice can have either a bending or a stretching dominated behaviour. The struts of bending dominated lattice (BDL) are mostly deformed by deflection, they collapse from local bending. BDL are very compliant. In stretching dominated lattice structure (SDL) some struts withstand tensile forces and most struts sustain compression from axial loads [105]. SDL has stiffness and strength significantly higher than BDL and is linearly scaled with the relative density [104], [106]. However, they do not dissipate the energy well, causing sudden failure. See Table 4 for the specific structural properties of each topology studied in this thesis.

Some topologies have a high anisotropy such as Tesseract, X-Shape, Cubic or Rhombic Dodecahedron, while others have more isotropic behaviour such as Cubic Center, Octet and Vintile [107]–[109]. Also, for some topologies like Tesseract, as the relative density decrease they become more anisotropic [109]. The anisotropy affects the stiffness response of the lattice material depending on the direction of the applied load [110]. By rotating the unit cell by 45% the stiffness can be evaluated along the axial direction. X-Shape, Cubic Center and Octet are examples of unit cells that have higher stiffness when they are tessellated in their diagonal directions. On the opposite, Cubic, Tesseract and Vintile have a higher stiffness along the axial direction. [109]

Table 4 Comparison of unit cell topology mechanical properties

Unit cell	Mechanical Properties	Unit cell	Mechanical Properties
<p>Body Center Cubic (X-Shape)</p>	X-Shape is a BDL, that has no strut parallel or perpendicular to the axial loading. The inclined struts are the only one bearing loads. Like Diamond its struts bend more than the struts in most other topologies. Stress concentrations are found at the joint [108], [109], [111]	<p>Tesseract</p>	Tesseract is an SDL topology that contains struts parallel and perpendicular to the axial loading direction. Compared to the other structure its stiffness is average [109]
<p>Cubic</p>	Cubic is an SDL structure; its vertical struts are the only bearing the loads creating stress concentration at the 8 junction points. It has one of the highest stiffness [107], [109], [112]	<p>Tetrahedron</p>	Tetrahedron with the Sommerville arrangement #3 is an SDL that contains strut parallel and perpendicular to the axial loading direction [113]
<p>Cubic Center (Star)</p>	Star is an SDL. Under uniaxial load, the star lattice was shown to have high stiffness and strength. They contain struts parallel and perpendicular to the axial loading direction [114]	<p>Truncated Cube</p>	Truncated Cube is one of the stiffest lattice structures. It is a very similar structure to Cubic, but the strut arrangement at the corners allows to transmit the forces preventing stress concentration at the junctions. [107]
<p>Diamond</p>	Diamond is a BDL structure that behaves like the X-Shape. Their struts are bending more easily than the other structures. It doesn't have strut parallel or perpendicular to the axial loading and it creates the most compliant structure. [107], [111], [115]	<p>Truncated Cuboctahedron</p>	Truncated Cuboctahedron is a relatively stiff topology with a very high fatigue strength [101], [116]
<p>Octet</p>	Octet strut is often used in the mechanical field for its high strength to weight ratio, it has a high stiffness and is considered a SLD topology. [113], [114]	<p>X-Cross Grid (FBCCXYZ)</p>	They contain struts parallel and perpendicular to the axial loading direction.
<p>Rhombic Dodecahedron</p>	Rhombic Dodecahedron has properties that vary significantly depending on its anisotropy. Under a horizontal orientation as seen in the image, it is very compliant. When it as vertical orientation, it becomes one of the stiffest unit cells. No vertical or horizontal struts are present to bear load thus stress concentrations happen at the intersection joints. [107], [108]	<p>Vintile</p>	Vintile does not have struts parallel to the axial loading, the angle struts bear the load, they are the ones deforming the most. These struts distribute the stress uniformly between their adjacent inclined and vertical struts, reducing stress concentration at the joints. [109]
<p>Rhombic Cuboctahedron</p>	Rhombicuboctahedron is a relatively stiff topology [107]		

2.4.2 Tissue growth and Lattice Materials

The application of lattice structures can benefit cell attachment by providing a larger surface area. Also, lattice structures increase the accessibility of bone morpho-genetic proteins (BMPs), insulin-like growth factors (IFGs) and transforming growth factors (TGFs) needed for bone growth. To date, most objective for lattice optimization are based on strain energy, stiffness, permeability and stress distribution [69], [117]. Before discussing the relationship between bone ingrowth and implant structures, it is important to understand the methods used to analyze the biological environment. Typically, *in vitro*, studies used scanning electron microscopy (SEM) to evaluate cell attachment within a lattice, to quantify the proliferation. Also, DNA synthesis and ATP concentration were measured. Alkaline phosphatase was used to identify osteoblasts, whereas, alizarin red-S was used to identify the mineralization [113]. Additionally, it should be noted that few studies evaluating bone and material structure relationship were done on 3D printed porous structures built with arbitrary pore arrangement like foam [118]–[123] and numerous studies were done on non-stochastic cellular materials [18], [19], [44], [113], [124]–[134].

Relative Density and Porosity

It has been shown that higher porosity contributes to better bone ingrowth. However, too high porosity will reduce the mechanical strength of the structure and may result in failure of the implant [19]. The increase of porosity increases the accessibility of nutrients and oxygen and facilitate the removal of cell debris due to a higher fluid flow [18], [129], [130]. The lower the porosity, the slower the cells can invade the lattice structure [131]. If the porosity is less than 40%, the migration and the ability of cells to invade the structure is prohibited. [113]

Permeability

The permeability of the structure is characterized by the gradient of pressure pushing the fluid through the pores. The permeability is directly related to the porosity of the structure. An increase in the porosity will subsequently increase the permeability due to less structure obscuring fluid flow [69]. Permeability allows to quantify the capacity of MSC's to penetrate a structure. When it is too high, the increase of the fluid flow velocity provides less time for the osteoblast to attach to the surface, this is also known as cell washout [133], [134]. However, higher fluid velocity allows the nutrients and oxygen transport deeper in the structure [118], [124], [127], [134]. A unit cell with higher complexity can decelerate the fluid flow and cause higher surface friction. Those two events both favour the osteoblast attachment to the lattice or porous structure. [134]

Pore Size

Pore size has previously been investigated by many *in vitro* and *in vivo* studies. The pore itself is the opening between the strut, that allows MSCs, blood vessels and nutrient to enter the lattice structure (Figure 9). It was shown that when the pore size is bigger there is a shift toward osteoblast maturation, instead of proliferation [119]. The osteoblast also spreads only on the struts surface [133]. As for small pores, there is more calcium deposition compared to large pores and bone can bridge from one surface or one strut to another, this is called pore bridging. This shows that small pores increase the osteogenic activity and lead to more calcium-containing mineralized matrix. [119] However, if the pores are smaller than 100 microns the oxygen and nutriment transport to the centre of the structure is obstructed. Pore bridging also can alter the entrance of blood vessels furthermore [133]. The lack of oxygen and nutriment consequently inhibits both the cell proliferation and maturation [19], [119], [120], [125], [130], [133].

Figure 9. Pores of Truncated Cuboctahedron unit cell, with a) 5% and b) 50% relative density

The result of a study investigating pore sizes between 450 and 1200 micron, suggested that the optimal range is between 300 and 600 microns. Lower cell activity was found in pores between 900 and 1200 microns. [126] Furthermore, an *in vivo* study comparing a Diamond lattice with pore size of 300 microns, 600 microns and 900 microns, found that Diamond 600 microns had the highest bone growth [44]. An *in vitro* experiment made with rats' osteoblasts showed that smaller pore size (40 microns) has more cells, alkaline phosphatase activity and osteocalcin. While bigger pores (100 microns) allow faster cell migration due to better oxygen and nutrients transport [121]. The ideal pore size would be bigger than the one explored in this study, since it was done with rat cells and the vascularization cannot be included *in vitro*. In fact, two *in vivo* studies found higher levels of alkaline phosphatase activity, osteocalcin content and new bone formation in structures with 300 to 400 micron pore sizes [122] As for the blood vessels, pores between 470 microns and 590 microns are the ones leading to the highest amount of blood vessels. [125] Finally, according to a review paper, pore size between 300 and 400 microns are optimal to improve tissue ingrowth [128].

Surface Area, Roughness and Shape

Firstly, lattice with the highest surface area provides more surfaces for cells to attach and are shown to increase bone regeneration [127], [133]. Secondly, the surface type has an impact on the regulation of the bone cell during healing. In Ti6Al4V material, sandblasting and acid etching with hydrogen chloride and sodium hydroxide, provides a rough surface which enhances the bone regulation [123]. The migration of cells inside a lattice starts with focal adhesion, which is when the lamellipodia and the filopodia protrude at the front of the cell and adhere to the surface of the biomaterial. Those extensions of the cell membranes are like little arms that pull the cell forward. The front lamellipodia of the cell adhere to the surface and the back ones release their contraction like if the cell was walking on the surface [135]. This concept is important to understand since, the ability to have a strong grip on the surface can determine their cell's response and gene expression, consequently affecting the proliferation. Thirdly, the growth rate is affected by the pore shape [136]. The bone always starts growing in corners. No matter what shape the pores have, as the bone grows it forms smaller circular pores. The more corners there are in a structure, the more there is occlusion of bone [133], [136], [137]. Furthermore, surfaces with concave shape show an increase in osteocyte attachment, faster cell migration and proliferation as well as better tissue differentiation compared to flat and convex shapes. [9], [128]

Unit Cell Parameters

The unit cell length can also play a major role in orthopedic implants. The larger the unit cells are, the fewer number of unit cells will fill the implant. Moreover, some implants such as hip implants, necessitate a structure that can sustain high stress in all direction. The use of isotropic topologies in the design of those implants can prevent improper deformation and implant failure by having the same strength in all direction.

While there isn't enough information in the literature to discuss the details of bone response toward the different lattice topologies, a special attention should be given to Diamond unit cells. This unit cell is composed of 4 main nodes connected by 4 struts, all oriented at an angle of 109.57 degrees, making its structure and mechanical behaviour the closest to trabecular bone [138]. It is the topology the most utilized in biomedical research and implant development. It has been used in previous research for biomedical applications such as for the femoral stem of hip implants [139]–[141]. Those studies all choose the Diamond topology due to its high compliance and its isotropic mechanical properties. In the study made by Jetté et al., 2018 [141] the usage of Diamond lattice was able to reduce the stem stiffness by 31% while providing enough strength.

CHAPTER 3 - MODEL DEVELOPMENT

3.1 - Introduction

The described mechano-regulatory algorithms are quite complex and are computationally expensive. In 2016, Antonio Boccaccio, Antonio Emmanuele Uva and Michele Fiorentino developed a slightly simplified method of evaluating bone growth through various porous structures. In 2016, they published the first research paper briefly describing their concept [15]. This research team is the only one, to this day, that has adapted the mechano-regulation theories of Prendergast al., 1997 [4] for the optimization of lattices. The method suggested allows the estimation of the capacity of a structure to offer a proper environment for bone growth within a reasonable computational time. Overall, the BS equation presented in Chapter 2 is the same one used for the simplified method. However, the bone growth potential is assessed at one specific instant in time and not over the total healing period. Despite the simplification, when comparing similar structure, they showed that bone was estimated in the same regions in a structure as the bone ingrowth found *in vivo* [13].

This chapter will first describe the path taken to reproduce Boccaccio's simplified model. Secondly, steps taken to develop a new way to automate the evaluation of any strut-based topology, with adjustable parameters will be explained. Thirdly, convergence analysis was performed to determine the minimum number of elements required.

3.1 - Boccaccio et al. Simplified Method

Like previous studies, Boccaccio prepared his FEA model, by creating two components. The first being the solid lattice structure and the second representing the space where bone is expected to grow (the void space within the lattice). As explained earlier, the space within the

lattice is assumed to be filled with granulation tissue. This granulation tissue is modelled as a poroelastic material, and its material properties are the same as those of previous models. They were determined experimentally by Claes et al., (1998) [142] and Goodship and Kenwright (1985)[143]. Boccaccio et. al. modeled lattice as an unspecified biodegradable polymer with similar mechanical properties to bone. It was also compared with polycaprolactone but with a lower Young's modulus [6]. Their polymer lattice is also modelled as a poroelastic material.

The main idea behind Boccaccio's method is to apply pressure on top of a lattice structure to induce structure deformation, leading to fluid movement in the lattice. The pressure they apply on the structure is ramped over a period of 1 second, which can be considered as the time an implant can be loaded *in vivo* condition [144]. After one loading cycle, the BS acting on hypothetical cells in the granulation phases is calculated for each element. This calculation was made using Equation (1) and (2) described in Chapter 2 based on octahedral shear strain and fluid velocity. As previously mentioned, in Prendergast's model the biophysical stimulus would change in time, as the granulation tissue is replaced by, fibrous tissue, cartilage tissue, immature bone, or mature bone. Additionally, the diffusion theory would be applied to simulate the gradual entrance of the cell, inside the structure. The inclusion of this diffusion process and the tissue change as a function of time, increases the computational cost to carry out analysis by at least two orders of magnitude. [11] The strategy used to prevent this challenge is to remove the iteration process and neglect the transient phase of cell diffusion by assuming that MSC are already homogeneously distributed throughout the granulation tissue [9]–[13], [15]. The second major difference is the phenotype of the cells considered. It is still possible to estimate the different cell phenotypes and their corresponding tissue, for each element. However, in Boccaccio's model the focus was the promotion of bone tissue. In that case, only elements having BS in the range of 0.01 and 0.53 were considered to procure an optimal environment for bone to grow (Table 1).

Boccaccio's model was used for 7 different analyses. In the first one they investigated structures with pores distributed in an aligned periodic fashion, where a comparison between, elliptic, circular, rectangular, square and staggered elliptic pores of different sizes were made [9]. In their second study, they optimized a functionally graded lattice with circular pores [15]. Then they investigated their first strut-based topology; Rhombicuboctahedron. They evaluated the lattice with both square and circle cross-sections, made of different strut diameters and with polymers of 3 different Young's moduli [13]. Another study following the same goal was done to investigate four additional lattice topologies: Truncated Cube, Truncated Cuboctahedron, Rhombic Dodecahedron (with vertical and horizontal orientation), and Diamond [10]. In their last three studies, they investigated irregular load adapted scaffold optimization [10], they compared structures with spherical pores to elliptic and rectangular extrusions [11], and they tried to optimize the distance between the strands in cylindrical scaffolds[12].

3.2 - Bone Growth Estimation Tool Development

In this section, the development of an automated tool to determine which elements procure the right environment for bone to grow will be explained; starting with the computer-aided design (CAD) modelling, followed by the FEA preparation and data collection. The model is prepared to collect data for both lattices made of solid Ti6Al4V material and for lattice made of poroelastic material representing the polymer previously studied.

3.2.1 Parametrized lattices

Ansys SpaceClaim

The first step undertaken was to write a PYTHON script in Ansys SpaceClaim (Appendix G) to automatically generate lattice topologies with specific parameters; unit cell length, strut diameter, pore size, cross-section and topology (Figure 10). The unit cell length, strut diameter and pore size can be any dimension inputted in the script. For the topology, the location of the unit cell nodes as well as the node connection forming the struts were determined. They were added as a 3D coordinate system x, y, z, in the script. Afterward circle, square, rectangle, triangle, diamond, and I-beam cross section shape were also scripted with modifiable dimensions (Figure 11). Sketch planes perpendicular to each strut's direction were created. Then the location of the cross-section's geometry nodes was calculated based on the circle's radius, diamond's height, triangle's width, the square and rectangle's height and width as well as the height, width, web, and flange of the I-beam inputted as parameters. Those nodes were eventually sketched on the plane with lines connecting them. Once the topology and the cross-section were sketched, the script uses the extrusion tool to extrude the cross-section following the strut trajectory.

Figure 10. Octet unit cell with parameters definition.

Figure 11. Strut cross section with adjustable dimensions, made from the PYTHON script built to generate the CAD of the structures, a) Circle, b) Rectangle, c) Triangle, d) Square, e) Diamond, f) I-Beam, in the X-Shape unit cell

Spheres were created at each node with the same radius as the cross-section's radius for the circle, with a radius of half the hypotenuse of the width and base of the rectangle, I-beam, and square cross-sections, with a radius the scale of half the largest width for the diamond and with a radius the scale of the distance between the centre and the vertex of the triangle cross-sections. Those spheres prevent having any protrusions from the strut intersections, that could cause unintended stress concentrations and FEA convergence issues. Once the unit cell was created the Boolean function was used to get the granulation tissue domain, covering all void space in the unit cell.

The lattice explored in this thesis is made from the tessellation of 4-by-4-by4 unit cell. It is the biggest lattice that is computationally reasonable to process on the FEA software and it is also the lattice size explored by previous studies [9]–[13], [15]. To facilitate the computation, process, a 2-by-2-by-4 lattice had to be generated with the application of symmetry condition on two of its sides to simulate that 4-by-4-by-4 lattice. The creation of this 2-by-2-by-4 lattice CAD,

was done by scripting the usage of the mirroring tool to tessellate the cell. Finally, the script also incorporated a plate on top of the lattice structure to apply uniform pressure. The plate was created with the material properties of steel, preventing deformation from the load. Its thickness was $1/3$ of the unit cell length in loading direction. $1/3$ of the length was sufficient to prevent any observational deformation of the plate. Each individual unit cell had a dimension of the length x , y and z inputted as parameters before running the script. See Table 5. for examples of the 14 lattices structures, automatically generated by the script based on pre-defined parameters.

At the end of the script, few coding lines were added to select the lattice surface, to output the solid phase surface area, then output the lattice volume and the total design space volume (excluding the plate), to calculate the relative density. The solid phase surface area and the relative density were then stored in a text file. Prior to the characterization in Chapter 5, this part of the script was used to determine the diameter that needed to be inputted to get 5%, 10%, 15%, 20%, 25%, 30%, 35%, 40%, 45% and 50% relative density. Furthermore, pore sizes were estimated manually using the *Three-Point Circle* and the *Measuring tool* in ANSYS SpaceClaim, by selecting the center point on the surface of 3 opposite struts. Those 3 selected points create a circle representing the pore size, while the measuring tool measures the diameter of that circle. See Table A1 for illustration of the pores. Predefined equations were available in the literature to calculate the exact pore size of the Tetrahedron and Cubic topologies [16].

Table 5. Unit cell and lattice structure CAD prepared at a 15% relative density

Unit cell	2x2x2 lattice	4x4x4 lattice and granulation tissue	Unit cell	2x2x2 lattice	4x4x4 lattice and granulation tissue
Body Center Cubic (X-Shape)			Rhombicuboctahedron
Cubic			Tesseract
Cubic Center (Star)			Tetrahedron
Diamond			Truncated Cube
Octet			Truncated Cuboctahedron
Rhombic Dodecahedron (H)			X-Cross Grid (FBCCXYZ)
Rhombic Dodecahedron (V)			Vintile

3.2.2 Poroelastic Materials

To allow coupling between deformation of the granulation tissue and fluid flow poroelastic elements were used. Treating the granulation tissue as a poroelastic structure containing a solid matrix saturated with a fluid is the most direct method to evaluate the fluid displacement induced by mechanical loading. The application of poroelastic materials for the granulation tissue can be pictured as the cells (Soil grains) bathing in the extracellular matrix (pore fluid), filling the void space of a lattice (Figure 12).

Figure 12. Two-phase model representing poroelastic material [145]

The fluid is considered inviscid and incompressible while the solid phase is considered linearly elastic and incompressible. These two phases are modelled by the combination of two governing equations. Equation (5) relates the mechanical equilibrium to the fluid pressure, where; T_{ij} is the Biot stress tensor, α is a ratio representing the radius of the specimen, δ_{ij} is the second order identity tensor and F_i is the body force per unit volume [146]. Equation (6) is a diffusion equation of the fluid pressure coupled with the time of change of the volumetric strain where; ϵ_V^e is the trace of the infinitesimal strain tensor of the solid matrix, K_m is the Biot modulus, p is the interstitial fluid pressure and q_i is the rate of volumetric flow crossing a unit area. Both equations are based on Biot's consolidation theory describing fluid-solid couplings in poroelastic materials

[146]. These problems can be entirely solved by ANSYS, FEA Software. When Bulk modulus of the grain and Bulk modulus of the fluid are specified, ANSYS can calculate a Biot's coefficient relating compressibility of grain to the bulk material.

$$\nabla \cdot (T_{ij} - \alpha \rho \delta_{ij}) + F_i = 0 \quad (5)$$

$$\alpha \frac{\partial \varepsilon_V^e}{\partial t} + \frac{1}{K_m} \frac{\partial \rho}{\partial t} + \nabla \cdot q_i = s \quad (6)$$

Some studies such as those published by Konstantinos et al. 2019 [147], Jun et al. 2020 [148], and Chou & Müftü. 2013 [149], applied 2D coupled pore pressure elements (Ansys CPT213 and Ansys CPT212) with the purpose of finding a biophysical stimulus for bone growth simulation based on Prendergast et al 1997 mechano-regulatory model. Those three studies also included the diffusion process of the MSC cells. Konstantinos et al and Chou & Müftü simulated fracture healing whereas Jun et al simulated bone growth around titanium dental implants. Likewise, these Ansys CPT elements are often used to study the cartilage behaviour [146].

Newtonian fluid models can also be used in simulations based on mechano-regulatory theories. While they are computationally less expensive than the poroelastic model, an inlet velocity must be assigned instead of relying on the deformation of the structure to create fluid motion. For instance, Sandino & Lacroix., 2011 [42] assigned an inlet of $10 \mu m/s$ based on trial and error aiming to have enough velocity for cell activity. If the Newtonian fluid model is coupled with the structural deformation, then the computation gets higher compared to the poroelastic model. Overall, both method led to comparable results [42].

3.2.3 Mesh Preparation

To mesh both the lattice structure and the granulation tissue, tetrahedral elements with 10 nodes, 4 degrees of freedom were selected (Figure 14). They allow a quadratic displacement

behaviour, ideal for modelling irregular geometries. To specify the usage of tetrahedrons as the meshing method of both components, A *Patch Conforming Method* was set up in Transient structural, assigning the “Method” as *Tetrahedrons*, the “Algorithm” as *Patch Conforming* and the “Element Order” as *Use Global Setting*. The global setting selection makes the model use quadratic element order, having mid-side nodes. For the granulation tissue, coupled pore pressure element CPT217 were used. Those elements can make direct coupling to simultaneously satisfy both mechanical equilibrium equations and the fluid continuity equation altogether. For the Ti6Al4V, Ansys SOLID187 elements were used, they have the capacity to compute mechanical response following large deflection and large strain. To apply coupled pore element such as the CPT217, an APDL macro script was written and inputted as a command directly in Transient Structural Module (Figure 13). In the script the TB and PM command representing Permeability and Biot coefficient were activated to allow coupled pore fluid diffusion. Their material properties such as Young’s modulus, Poisson’s ratio, permeability, porosity, fluid weight and the bulk modulus of the solid and fluid also needed to be added to the APDL script.

```

/UNITS,uMKS
E=0.2      !Young's modulus x (MPa)
NU=0.167   !Poisson's ratio
KX=1E-4    !Permeability X (μm/s)
KY=1E-4    !Permeability Y (μm/s)
KZ=1E-4    !Permeability Z (μm/s)
WEIGHT=1E-8 !Weight (μN/μm^3)
BMS=2300   !Bulk Modulus of solid (MPa)
BMF=2300   !Bulk Modulus of fluid (MPa)
PR=0.8     !Porosity
/PREP7

ET,1,CPT217 !Choose the element type (Coupled-pore-pressure, 10 nodes, tetrahedral elements, poroelastic,fully direct coupling)
KEYOPT,1,12,1 !element type number 1, then knum number 12 Structural-pore-fluid-diffusion
MP,EX,1,E     !Assing the Young's modulus
MP,EY,1,E
MP,EZ,1,E
MP,NUXY,1,NU  !Assing the Poisson's ratio
MP,NUYZ,1,NU
MP,NUXZ,1,NU
TB,PM,1,,,PERM !Assing permeability
TBDATA,1,KX,KY,KZ
TB,PM,1,,,SP   !Assing Bulk modulus solid
TBDATA,1,BMS,WEIGHT
TB,PM,1,,,FP   !Assing Bulk modulus and porosity for Fluid
TBDATA,1,BMF,WEIGHT,PR

SHPP,ON
VMESH,ALL

```

Figure 13. APDL script for coupled pore pressure, element activation.

As for the Steel plate, SOLID186, 3D 20 nodes quadratic element were use, while SURF154 are automatically generated, to cover the top surface of the plate where pressure is applied. The 3D 8-node contact elements (CONTA174) were also automatically generated by the software between the lattice structure surface and the granulation tissue, the plate and the granulation tissue as well as between the plate and the lattice. They are recommended for coupled field contact analysis. While those elements represent the contact and sliding between 3D surfaces, the contacts between the surface of all three components were defined as *Bonded Contact*. Bonded contact also known as linear contact, ties the surfaces together in all direction and allows direct transfer of forces. This also implies that small sliding option is enabled, meaning that the contact interface won't be subjected to sliding motion more than 20% of the contact length. Since the lattice and the granulation tissue component were created using the SpaceClaim Boolean/cutting function, no gap or overlap can be found in the model. This avoids issues in the surface bonded contact, preventing unstable solutions. The element size is manually defined for the granulation tissue and the scaffold, beforehand. More details regarding mesh refinement will give in section 3.3.

Figure 14. Finite element mesh of an octet lattice, granulation tissue and plate, with element size of 92 microns, resulting in a total of 367,195 elements.

3.2.4 Material Properties

To compare the results obtained, to data from Rodriguez et al. 2018[10] and Boccaccio et al. 2018[13], both the lattice structure and the granular tissue were modelled with poroelastic elements. The material properties assigned for that part was the same as used by Rodriguez et al. 2018[10], the exact name of the polymer used was not given in their study (Table 6). In contrast, for the characterization of the lattice structure in Chapter 5, Ti6Al4V material properties were applied to the lattice (Table 6). Only linear elastic and physical properties were assigned to the materials. During automatization, the APDL macros are removed when a new geometry gets created. To automatically reassign the material's macros, an ACT extension was written and was linked to the main workbench script, which control the entire characterization (Appendix H).

Table 6. Material properties

Material properties	Rodriguez et al, Polymer Lattice [10]	Granulation tissue	Ti6Al4V Lattice [150]–[152]
Permeability ($\mu\text{m}/\text{s}$)	1×10^{-4}	1×10^{-4}	-
Porosity	0.5	0.8	-
Solid Bulk Modulus (MPa)	13 920	2300	-
Fluid Bulk Modulus (MPa)	2300	2300	-
Weight ($\mu\text{N}/\mu\text{m}^3$)	N/A	1×10^{-8}	-
Density ($\text{kg}/\mu\text{m}^3$)	1.14×10^{-15}	1×10^{-15}	4.43×10^{-15}
Young's modulus (MPa)	1000	0.2	113 800
Poisson's ratio	0.3	0.167	0.342
Compressive Strength (MPa)	-	-	970

3.2.5 Boundary Conditions

Figure 15 b) illustrates the boundary conditions applied to the model. A steel plate was created on top of the lattice to have uniform pressure applied on the surface of both the solid and the fluid phase nodes. This is also representative of the way the vertebral endplate would apply pressure on a lattice cage. Since the top plate was impermeable, the fluid flow across the top plate is automatically set to zero. Additionally, all degrees of freedom were constrained by a fixed support at the bottom of the structure. Then symmetry conditions were applied on two perpendicular surfaces, which simulates a 4-by-4-by-4 lattice structure. The DSYM command was activated to assure degree-of-freedom symmetry constraint on all the nodes of the surfaces of two perpendicular sides of the lattice, including the plate and granulation tissue. The orientation of the symmetry is normal to the surface orientation. The two other surfaces were representing the periphery of the structure. On periphery surfaces a fluid pressure of zero was assigned meaning that the fluid was able to flow across. To set those boundary conditions another APDL pre-processor (/PREP7) macro was written and directly inputted as a command in the Transient Structural Setup branch (Appendix I).

Figure 15. a) Octet unit cell b) 4X4X4 lattice structure made of octet unit cells with circle cross section at 15% relative density b) model setup, boundary conditions and constraints

As in the previous studies, the pressure applied to the top plate was ramped over one second period [144]. In fact, the average human physiologic walking speed creates a dynamic sinusoidal displacement at a frequency between 0.1 and 1 Hz [153]. To evaluate the significance of the time of pressure applications, a sensitivity analysis was done with pressure ramped between 0.5 and 4 seconds. This analysis was executed on one of the lattice structures evaluated in the characterization part. It is an Octet truss lattice structure, made of 1200 microns unit cell length, with a relative density of 15%, subjected to a 0.5 MPa pressure. The element size used was 100 microns, generating a mesh containing a total of 316,253 elements. The average

octahedral shear strain and velocity as well as the estimated bone percentage of pores having optimal stimulation for bone growth were collected for the granulation tissue elements, under pressure ramped of 0.5; 1; 1.5; 2; 2.5; 3; 3.5 and 4 second. Three plots were generated to represent the octahedral shear strain, velocity and bone growth estimation against the pressure time.

Figure 16. Average octahedral shear strain and velocity within a 2x4x4 Octet lattice, under a pressure of 0.5 MPa, in function of the loading time

Figure 17. Percentage of void space, optimal for bone ingrowth within a 2x4x4 Octet, under a pressure of 0.5 MPa, in function of the loading time.

The time of pressure applications had a very small impact on the velocity, the strain, and the final percentage of pores optimal for bone (Figures 16 and 17). The velocity was showing a small noticeable decrease. Since the average velocity of the granulation phase is relatively small, changing the ramped time by a few seconds doesn't make a meaningful difference in the bone percentage comparison made between structures. However, it is not known if the implication of the ramped time has more impact on other topologies or under higher loads. Ramping the pressure over a much longer period to simulate a person sleeping or sitting can potentially have a bigger impact on the results. Considering that the patients receiving the implant perform a variety of daily activities, choosing one specific ramping time of 1 second seems reasonable to compare the performance of the lattice structure.

3.2.6 Data Collection

To collect the information related to the porous media, an APDL post-processor (/POST1) macro was written and executed as a command in the Transient Structural Solution branch (Appendix J). The fluid flow and the principal strains ϵ_1 , ϵ_2 , ϵ_3 , were retrieved and used to calculate the BS value in each element, using custom MATLAB code (Appendix K). When the BS value was between 0.01 and 0.53, the element volume was saved. The total volume of elements containing optimal stimulation was calculated and divided by the total pore volume. This ratio was used to quantify the lattice performance in terms of the percentage of the void space that can promote bone ingrowth. Since Boccaccio's method looks at a ratio in terms of the total design space volume (including the lattice), this ratio was also calculated for the validation part. The element number of each element considered good for bone ingrowth were saved in a file. A second ACT script was written to input the bone element numbers in ANSYS and render a visual of their location inside the lattice (Appendix L). Another file, containing the element

numbers associated with the results from the velocity part of the BS equation was created and imported in ANSYS to get a visual of the fluid velocity distribution.

3.3 - Convergence Analysis

The high amount of poroelastic elements necessary to process the model requires a computer with a lot of random-access memory (RAM). For the structure analysis of the validation part, both the lattice and the granulation tissue are modelled as poroelastic elements. Thus, the computation is more demanding and takes longer. Three convergence analyses were made. Their results were plotted for the percentage of pores optimal for bone growth, the biophysical stimulus as well as the velocity and the strain part of the BS. It was necessary to separate velocity and strains in the convergence analysis since they might not require the same number of elements to reach sensitivity. The first and second analyse were done on a diamond lattice made of the polymer material with small strut size of 85 microns (Figures 18 and 19) and bigger strut size of 100 microns (Figure B1 and B2). The third and last analysis was done on a lattice made of Ti6Al4V. This analysis was done to evaluate the lowest number of elements that could be used in the characterization part of this thesis. Octet topology was chosen since pre-test results showed that it was one of the two most computationally demanding topologies. This is most likely due to the higher amount of truss leading to more surface contact and more joints. The maximum number of elements the computer could run was around 600 000 elements for the polymer and 800 000 for the Ti6Al4V. Rodriguez et al. 2018 [10] were able to use 5 million elements under 1.5 hours. This could be explained by the fact that they were using ABAQUS software, which is better suited to process poroelastic materials.

Figure 18. Element stimulus average within a 2x4x4 Diamond lattice with unit cells of 637 microns and strut diameter of 85 microns, made of polymer strut. Convergence analysis made in function of the number of elements used. Strain/0.0375 and velocity/3 are the scaling factors added together to get the BS

Figure 19. Percentage of void space, optimal for bone ingrowth within a 2x4x4 Diamond lattice with unit cells of 637 microns and strut diameter of 85 microns, made of polymer strut. Convergence analysis made in function of the number of elements used

Figure 20. Element stimulus average within a 2x4x4 Octet lattice, made of titanium strut. Convergence analysis made in function of the number of elements used. Strain/0.0375 and velocity/3 are the scaling factors added together to get the BS

Figure 21. Percentage of void space, optimal for bone ingrowth within a 2x4x4 Octet lattice, made of titanium strut. Convergence analysis made in function of the number of elements used

Following all convergence analysis, the fluid velocity was always overestimated, and the strain was always slightly underestimated, when less elements were used (Figure 18, Figure 20 and Figure B1). For Ti6Al4V the usage of at least 300 000 elements, (average element size = 0.11 mm for 1.44 mm unit cell length) the change in number of elements resulting in bone is $\pm 2.5\%$ (Figure 21). Moreover, smaller struts and lower relative density required more elements to reach the same deviation (Figure 18). In fact, when struts and relative density were smaller, there was a higher proportion of the elements forming the granulation tissue.

In a study evaluating a few structures, using a very fine mesh is reasonable since the computation time only goes up to 11 hours. It can reduce the risk of errors and result in smoother visuals of the elements with the right stimulation for bone growth. For this project, the computational time mattered since the goal was to carry out a total of 532 analysis, for the characterization. The choice was made based on the fewest element number necessary to get enough sensitivity. Following Figure 21 the usage of more than 250 000 elements was high enough to determine the percentage of void space, optimal for bone growth. As for the biophysical stimulus components independently, more than 300 000 elements would be more appropriate for the velocity (Figure 20). The target to get good results was to use between 300 000 and 400 000 elements for all the models with a Ti6Al4V lattice. It was determined that a mesh refinement of 90 microns was sufficient to assure every model of the characterization part would have more than 300 000 elements, regardless of their relative density and their topology. Both the lattice and the granulation tissue were assigned the same element size.

CHAPTER 4 - VALIDATION

4.1 - Boccaccio and Rodriguez's Trend Validation

4.1.1 Introduction

This first part of the validation chapter discusses the replication of two studies that are using Boccaccio's method. The goal is to confirm that the results collected with the model developed in this thesis are correct. First, the paper by Rodriguez et al., 2018 [10] was replicated to evaluate if changing the topology of the lattice, the cross-section and the orientation of an anisotropic unit cell outputs the same trends. Then, the paper by Boccaccio et al., 2018 [13] was also replicated to see if changing the material properties affects the lattice potential for bone.

4.1.2 Method

Rodriguez et al., 2018 [13] were optimizing Diamond, Truncated Cube, Truncated Cuboctahedron, Rhombic Dodecahedron with horizontal and vertical orientation, under 7 loads (0.05 MPa; 0.1 MPa; 0.15 MPa; 0.5 MPa; 1 MPa and 1.5MPa). Their objective function was to maximize the bone ingrowth for each load, by changing the struts diameters while keeping the unit cell length as 637 microns. By changing the diameter, it changes the relative density of their lattice structure which affects the stiffness and consequently the bone stimulation. Bone ingrowth percentage was based on the ratio of the total element volume having optimal BS to the total design space volume. It is important to understand that this paper was optimizing with the usage of MATLAB *fmincon* function. They would need up to 30 different analysis per pressures for each topology. They also used 5 million elements to get the most accurate results, which took a total of 3150 hours of computation, for their entire study.

While the goal of this part was to see if the set up created in Chapter 2 was able to generate the same percentage trends with the same optimal strut diameter, it was not reasonable

to reproduce a complete optimization. Instead, a csv file containing different discrete strut diameters, with maximum 20 microns of variance were entered. Then those data point were used to estimate the optimal strut diameter for bone stimulation and the bone ingrowth percentage associated with that optimal strut diameter. With this manual method the peak bone ingrowth percentage could be estimated within a strut diameter variation of ± 20 microns. This means that the optimal bone ingrowth found could potentially increase slightly if the strut diameter was 19 microns more or less than the optimal diameter found. However, the difference is not big enough to change the trends. As for the modelling, the FEA method describes in Chapter 2 was applied with the usage of the polymer lattice material properties (Table 6), and unit cell length of 637 microns.

Furthermore, at the time of the data collection for the validation, the computer used wasn't powerful enough and 120 000 elements (average element size = 0.09 mm for 0.637 mm unit cell length) was the maximum it could process. When the number of elements is that low, BS tends to be overestimated (Figure 18 and Figure B1). When the average BS is overestimated (with smaller strut), a low element count leads to lower bone growth, since some elements who should have been optimal for bone will have passed the upper threshold for bone. The opposite happens when the BS is in the lower range (with bigger strut), some elements are considered bone because they stay within the mature bone optimal range with low element counts. This results in higher bone percentage than it should with smaller struts and lower bone percentage for bigger struts.

For the comparison of the topology trends and the anisotropy, the Diamond, the Truncated Cube, and the Rhombic Dodecahedron with both the horizontal (Rhombic H) and vertical (Rhombic V) anisotropy were prepared with the circle cross section, using the method described

in Chapter 3. For the comparison of the cross section, Diamond and Truncated Cube were also prepared with square cross sections. As for Boccaccio et al., 2016 [13], Rhombicuboctahedron was prepared with a Young's modulus of 500 MPa, 1000 MPa and 1500 MPa. Only five pressures were tested in this paper (0.05 MPa; 0.1 MPa; 0.5 MPa; 1 MPa and 1.5 MPa). The Unit cell length parameter was set to 637 microns in all models, just like the length used in those comparative studies.

4.1.3 Results

Figure 22. Comparison of the optimized strut diameter maximizing the percentage of the total volume that can stimulate bone based on the current simulation with data published by Rodriguez et al. 2018 [10], Bone ingrowth and diameter under pressures of 0.05 MPa, 0.1 MPa, 0.15 MPa, 0.5 MPa, 1 MPa and 1.5 MPa, inside a) Diamond and b) Truncated Cube structures made of polymer with circle and square cross section

Figure 23. Comparison of the optimized strut diameter maximizing the percentage of the total volume that can stimulate bone based on the current simulation with data published by Rodriguez et al. 2018 [10], Bone ingrowth and diameter under pressures of 0.05 MPa, 0.1 MPa, 0.15 MPa, 0.5 MPa, 1 MPa and 1.5 MPa, inside Rhombic Dodecahedron structure made of polymer with vertical (Rhombic (V)) and horizontal (Rhombic (H)) cell orientation

Figure 24. Comparison of the optimized strut diameter maximizing the percentage of the total volume that can stimulate bone based on the current simulation with data published by Boccaccio et al. 2018 [13], Structure is tessellated with Rhombicuboctahedron unit cells made of polymer with young's modulus of 500 MPa, 1500 MPa and 1000 MPa, under pressures of 0.05 MPa, 0.1 MPa, 0.5 MPa, 1 MPa and 1.5 MPa

When comparing the square and circle cross sections, the same trends could be found with both the Diamond and the Truncated cube topology. In general, circle cross section led to higher bone ingrowth (Figure 22). Interestingly, for the Diamond, the circle cross section was less accurate than the square cross section. Additionally, for Truncated cube, the optimal strut diameter was overestimated for circle cross sections and underestimated for square cross sections in the replication. As for the anisotropy, Rhombic (H) showed lower bone potential compared to the vertical anisotropy in both the original paper and the replication. The Rhombic (H) was also slightly more accurate compared to Rhombic (V) in terms of optimal strut diameter (Figure 23). In Figure C1, the combination of all topology results in one graph, showed the same trends for this paper and the replication in terms of percentage of total space with bone. However, all the bone percentages were 10 to 20 % lower in the replication while all the optimal strut diameters were slightly higher (Table C1; C2; C3; C4).

In both the paper and the replication results, higher Young's moduli led to a higher percentage of bone, and smaller strut sizes were optimal (Figure 24). Additionally, with all three Young's moduli, both data collections showed higher bone potential under lower pressures (Figure 24 and Table C5).

4.1.4 Discussion

Overall, the low number of elements can explain most of the differences between the original papers and the replication results. Additionally, the only material property information that was provided is presented in Table 6. The exact name of the material and information such as the density of the material was not given. Thus, polycaprolactone density was used. From Figure 22, it was noticeable that the Diamond results with square cross-section were more accurate,

compared to the results of circle cross-section. As previously mentioned, the inaccuracy due to the lack of elements, leads to an overestimation of the velocity. Diamond is a highly permeable structure, and the circle cross section is more hydrodynamic than the square cross-section, causing less drag in the flow. This could explain why the velocity inaccuracy could have potentially played a bigger role in the biophysical stimulus of the Diamond with circle cross section. Nevertheless, all the trends compared were having the same behaviour using the model built in this thesis, which supports the usage of the new tool.

Figure 25. Percentage of the volume that can stimulate bone growth in function of the total and the void space volume inside a polymer structure made of 637 microns diamond unit cells with different strut diameter

Both papers are analyzing bone ingrowth compared to the entire volume of the structure instead of considering only the void space. If you increase the strut diameter, then you reduce the void space, in return it reduces the amount of space for bone to grow. Hence, even if different strut diameters can offer a 100% optimal environment for bone in the void space, bigger diameters will always be considered less effective (Figure 25).

Another finding during the validation was that the BS is higher on the side of the struts. A visual example can be examined in Figure 26, where two Diamond structures with different strut size showed similar bone percentage. The first one made with 85 microns diameter struts, resulting in 79% of the void space optimal for bone and the other one made of 290 microns had 83% of the void space optimal for bone. As discussed in Chapter 2, MSC's requires an attachment to the lattice to form bone. However, the simulation can predict bone growth in the granulation tissue without lattice attachment as shown in Figure 26a, which can be considered a biologically irrelevant solution.

Figure 26. Element with optimal environment for bone growth (brown elements) inside lattice structures made of Diamond unit cell with a strut diameter of a) 85 microns (biologically irrelevant solution) and b) 290 microns (biologically relevant solution), loaded with 0.5 MPa of axial compression

4.2 - Canines *In vivo* Comparison (Octet versus Tetrahedron)

4.2.1 Introduction

Only a few studies have evaluated different lattice topologies, *in vivo*, to test their bone ingrowth capacity. Arabnejad et al., 2016 [17], has compared two SDL, the Tetrahedron with Sommerville #3 arrangement, to Octet lattice. Their lattice implants were 3D printed with Ti6Al4V and implanted in canines' lateral femoral cortex (Figure 27). The same topology comparison as described above was made in this section to see if the method applied in this thesis results in similar trends.

Figure 27. Implants inserted perpendicular to the lateral femoral cortex of the canines from Arabnejad et al., 2016 [17]

4.2.2 Method

The study's implants were cylindrical and contained unit cells with the average dimension defined (Table 7). Four- and eight-weeks post-surgery, they used transverse electron micrographs to quantify the bone as a percent of the void volume.

Table 7. Implant lattice description

Unit Cell Characteristics	Tetrahedron	Octet
Unit length (mm)	1.44	1.44
Diameter (mm)	0.39	0.25
Porosity (%)	56	70
Pore Size (micron)	438	772

To generate the CAD model, the same unit length and strut diameter as the paper was used in the PYTHON script. The Ti6Al4V material properties were assigned to the lattice. Once the lattice was created, it was confirmed that it had the corresponding porosity and pore size as the one defined in their research. During the meshing, a minimum of 350,000 elements were used. As for the pressure, it was not possible to know the exact load that was applied to the implant. Therefore, 5 different pressures (0.5 MPa, 0.625 MPa, 0.75 MPa, 0.875 MPa and 1 MPa) were applied to see if the trend of bone growth obtained corroborates the paper's results. The pressure chosen was based on the given weight of the canine (30 – 35 kg) [17][16] and the premise that if their body weight is spread between 4 legs, each leg would support a load between 73.5 N and 85.8 N. Image J was used on Figure 23, to estimate the surface area of the cortical bone that would support this load. While no scale is given with the figure, the implant width of 5 mm is visible on the micrograph, allowing the estimation of the image scale. The surface area was found to be approximately 120 mm². It would be reasonable to assume that a pressure around 0.61 MPa and 0.72 MPa is applied to the implants. This does not consider the other forces applied internally by the muscle and externally during movements. According to a study done on sheep during its walking gait, the axial contact force applied to the tibio-femoral joint is on average: 2.1 times the body weight [154]. While this study looks at the force in the joint and not the femoral cortex, it is essential to keep in mind that the pressure on the implant could still be much higher; up to 2.5 MPa and 5.7 MPa

Figure 28. Transverse electron micrograph showing bone ingrowth after A) 4 weeks and B) 8 weeks, in 1) Octet truss and 2) Tetrahedron from Arabnejad et al., 2016 [17]

4.2.3 Results

The average bone ingrowth found in canine after 4 weeks was 35.5% for Octet and 28.6% for Tetrahedron. While the bone ingrowth found after 8 weeks, it was 56.9% for Octet and 41.3% for Tetrahedron (Table 8). The pressures providing the closest percentage for comparison were 0.875 MPa and 1 MPa (Figure 29). The results obtained from the simulation also revealed that under the unit cell chosen characteristics, Octet had more potential to grow bone compared with the Tetrahedron topology. Table D8 offers a visual of the strain distribution and the location of elements optimal for bone, for all pressures examined. A close-up view, in Figure 25, shows that the elements having an optimal environment were found in the creases and on the surface of the strut. This is the exact location, where bone is typically found *in vivo*.

Table 8. Percentage of void space optimal for bone ingrowth with pressures of 0.5 MPa, 0.625 MPa, 0.75 MPa, 0.875 MPa and 1 MPa compared with the percentage of bone ingrowth found in structures with the same lattice parameter implanted in canine (*in vivo* data published by Arabnejad et al. 2016 [17])

Pressure (MPa)	Octet	Tetrahedron
0.5	9.4	3.6
0.625	19.5	6.2
0.75	30.8	10.0
0.875	42.1	15.1
1	56.2	22.1
Post-surgery time	Octet	Tetrahedron
4 Weeks	35.5	28.6
8 Weeks	56.9	41.3

Figure 29. Percentage of void space, optimal for bone ingrowth with pressures of 0.5 MPa, 0.625 MPa, 0.75 MPa, 0.875 MPa and 1 MPa compared with the percentage of bone ingrowth found in structures with the same lattice parameter implanted in canine after 4 and 8 weeks (*in vivo* data published by Arabnejad et al. 2016 [17]), in red octet and blue tetrahedron structures

4.2.4 Discussion

From these results, the model used in this thesis showed similar trends and was able to show the superior bone ingrowth capacity of the Octet lattice. In terms of limitation, real bone

grows from higher cell concentration and diffuses with time inside a structure, while the simulation assumes that the void space is already filled with cells. For example, looking at figure 23, no bone grew in the implant areas located in the medullary canal. It is not yet possible to consider this phenomenon with the *in silico* model. Thus, the percentage values should not be directly compared between *in silico* and *in vivo* experiment, but rather used as a reference to exclusively compare structures tested with the model. Even if the pressure applied to the implant gets higher, from the graph it is reasonable to expect that the Octet will remain more performant than Tetrahedron. An additional comparison was performed at 2 MPa to confirm that the Octet remained better with 100% of its pores optimal for bone stimulation while tetrahedron had 84% of its pores optimal for bone stimulation.

Figure 30. Close up view of the elements having an optimal environment to grow bone (brown elements) inside an octet lattice

4.3 - Rabbits *In Vivo* Validation (Diamond, Cubic, Tetrahedron)

4.3.1 Introduction

Deng et al., 2021 [16] has also investigated the performance of Ti6Al4V 3D-printed lattices *in vivo*. They compared the bone growth inside Diamond, Cubic, CIR and Tetrahedron lattices, implanted in the lateral femoral condyle of white rabbits. Once again, to test the

reliability of the method used in this thesis, trend comparison was made between the *in silico* result and the *in vivo* bone growth measured in this paper. The advantage of this study is the comparability between their unit cell characteristics. The four different structures explored by Deng et al., were all having the same porosity and similar pore size. This makes the topology itself the only variable that can explain the difference in bone ingrowth.

Figure 31. Titanium lattice implanted in the lateral femoral condyle of a rabbit at low speed, from Deng et al., 2021 [16]

4.3.2 Method

The study's implants in this paper were also cylindrical and contained unit cells with the average dimension defined in Table 9. They quantified the bone as a percent of the void volume, six- and twelve-weeks post-surgery.

Table 9. Implant lattice description

Unit Cell Characteristics	Diamond	Cubic	Tetrahedron
Unit Length (mm)	1.6	1.2	1.84
Diameter (mm)	0.656	0.567	0.43
Porosity (%)	64.5	65	65
Pore Size (mm)	650	660	650

The same procedure as described in section 4.3.2 was applied to generate the CAD and the mesh following the characteristic of Table 8. CIR was not tested since it is not a strut-based

lattice and would have needed to be generated manually. As for the pressure, it is still not possible to know the exact load that is applied to the implant. In that case, 4 different pressures (0.4 MPa, 0.5 MPa, 0.6 MPa and 0.7 MPa) were applied to see if the trend of bone growth found corroborates the paper's results. The chosen pressures were based on the given weight of the rabbits (2.5 kg – 3 kg)[16] and the premise that if their body weight is spread between 2 legs and the load is entirely applied to the circular implant of 5 mm diameter, the implant would support a load of around (0.62 MPa – 0.75 MPa). However, part of the weight can be supported by the remaining femoral condyle surface and by the front legs. This does not consider the other forces applied internally by the muscles and externally during movements. Once again, if the contact force in the joint of rabbit follows the 2.1 weight ratio found by the study performed on sheep [154], the pressure on the implant could go up to 2.6 – 3.1 MPa.

4.3.3 Results

The average bone ingrowth found in rabbits after 6 weeks was 15.2% for Diamond, 13.7% for Tetrahedron and 11.4% for Cubic. While after 12 weeks, it was 23.1% for Diamond, 19.1% for Tetrahedron and 18.3% for Cubic (Table 9). The pressure providing the closest percentage for comparison was 0.5 MPa and 0.6 MPa (Figure 32). The results obtained from both the simulation and the *in vivo* paper revealed that Diamond had most potential to grow bone followed by Tetrahedron. Cubic was having the least amount of bone. In Table 10, the elements optimal for bone are highly located on the periphery for Tetrahedron and Cubic, whereas Diamond's bone elements are more distributed. Table D1 to D7 offers a visual of the strain distribution and the location of elements optimal for bone at all pressures examined. A cross section view in Figure 28, shows that the elements having an optimal environment in structure with square pores, were forming a circular shape just like the tissue formed in cell culture.

Table 10. Percentage of void space optimal for bone ingrowth with pressures of 0.4 MPa, 0.5 MPa, 0.6 MPa and 0.7 MPa, compared with the percentage of bone ingrowth found in structures with the same lattice parameter implanted in rabbits (*in vivo* data published by Deng et al. 2021 [16])

Pressure (MPa)	Diamond	Tetrahedron	Cube
0.4	8.5	5.5	3.8
0.5	18.7	9.7	6.3
0.6	32.2	15.8	9.0
0.7	45.9	23.6	12.2
Post-surgery time			
6 Weeks (Deng et al.)	15.2	13.7	11.4
12 Weeks (Deng et al.)	23.1	19.1	18.3

Figure 32. Percentage of void space optimal for bone ingrowth with pressures of 0.4 MPa, 0.5 MPa and 0.6 MPa, compared with the percentage of bone ingrowth found in structures with the same lattice parameter implanted in rabbits 6 and 12 weeks after surgery (*in vivo* data published by Deng et al. 2021 [16]) in red Diamond, in blue Tetrahedron and green Cube

Table 11. Element with optimal environment for bone growth, under 0.5 MPa of pressure

Cubic – Center of lattice (symmetry side)	Diamond – Center of lattice (symmetry side)	Tetrahedron – Center of lattice (symmetry side)

Cubic – Periphery of lattice	Diamond – Periphery of lattice	Tetrahedron – Periphery of lattice

Figure 33. Bone ingrowth forming a circular shape when growing in square pores a) and b) Optimal area for bone to growth (brown elements) within a cubic tessellated structure c) tissue formed in cell culture inside a square HA plates, after 30 days Rumpler et al., 2008 [137]

4.3.4 Discussion

Once again, the model used in this thesis was able to show if a topology was better than another one. In both comparisons, the difference between the topologies was amplified when compared *in silico* instead of *in vivo*. In addition to the limitations highlighted in section 4.3.4; the 0.01 to 0.53 optimal range for mature bone could vary in different animal populations due to genetic variability. Additionally, the model assumes that there is a single tissue within the lattice (mature bone). It does not consider the potential growth of cartilage and fibrous tissue. Despite those limitations, all validation tests showed that the tool developed is reliable and could be used to improve implants' lattice design. More attention will be assigned to the behaviour of the topology, following the characterization of all fourteen topologies in the next chapter.

CHAPTER 5 - Bone Ingrowth in Ti6Al4V Lattice Topology

5.1 - Introduction

When developing a lattice implant for bone ingrowth, it is important to have appropriate nutrients supply inside the structure, enough space for vascularization, as well as biocompatible material that allows cells to adhere. The structure's mechanical properties must support the load but also lead to the appropriate biophysical stimulus to signal the cells to grow [10], [155]. To this date there is still no study that characterizes titanium lattice structure parameters based on mechano-regulatory theories. In this chapter, characterization, quantifying bone ingrowth for fourteen lattice topologies (Figure 34) at different relative densities under four different pressures is presented.

Figure 34. Unit cells

5.2 - Method

The lattice parameters investigated for the characterization were chosen based on findings from previous *in vivo* and *in vitro* studies (Table 12). The scripts and modelling method developed in Chapter 3 was applied to gather the results for this section, using Ti6Al4V material properties for the lattice (Figure 35).

Table 12. Lattice parameters evaluated for the characterization

Characteristic	Parameters
Pressure	0.5 MPa, 1 MPa, 1.5 MPa and 2 MPa
Unit cell length	1200 micron
Pore size	Between 130 micron and 1016 micron
Relative density	5%, 10%, 15%, 20%, 25%, 30%, 35%, 40%, 45%, 50%
Material	Ti6Al4V for the lattice, Granulation tissue for the fluid phase and Steel for the top plate
Number of elements	Between 300,000 and 400,000

Figure 35. Preparation of the octet topology with 15% relative density a) SpaceClaim CAD model b) Load and boundary conditions c) Mesh generated for the model

5.2.1 Pressure

Since the main objective in defining optimal lattice parameters is to improve LIF cages, the pressure simulated must be representative of the load applied to those implants. A study evaluating the loads on the vertebral body replacement in 5 patients during their post-operative physiotherapeutic exercises showed that the highest force applied on the spine was 600 N and the lowest was 200 N. [156] Another study showed that a patient lifting arms and changing position from lying to sitting can reach a maximum force of 589 N applied on vertebral bodies. [157] Finally, a third study investigated the intradiscal pressure in a 70 kg male for 24 hours and found that the lowest intradiscal pressure was 0.1 MPa while lying prone and the highest pressure was 2.3 MPa while lifting a 20 kg weight, bent over with round back. [158] Assuming no holes in the middle of the implant, the lumbar LIF cages have a footprint varying between 220 mm² (TLIF) and 1320 mm² (ALIF) (Table 3 and 13).

To estimate a possible pressure range that could be applied to the structures, the lowest and highest pressures were calculated as follows. If the patient applies the lowest possible amount of load (200 N) while receiving the biggest spinal cage of 1320 mm², 0.15 MPa of pressure would be applied. If the patient applying the highest amount of load received the smallest possible cage, a pressure of 2.72 MPa would be applied to the cage. Since the cage size varies substantially and surgeons choose the cage in function of the patient size, it is unlikely that the extremity of the range would be reached. Smaller implant cages are often accompanied by posterior fixation, which reduce the load on the implant [69]. For this study, the 4 pressures evaluated were 0.5 MPa, 1 MPa, 1.5 MPa and 2 MPa. It allows identification of which topologies are more suitable for lower pressure versus higher pressure.

5.2.2 Unit Cell Dimensions, Relative Densities, Topology and Element number

Unit cell length of 1200 microns was chosen so that the strut diameter can be a printable size and so that the pore sizes remain within the recommended range for angiogenesis, growth factors, alkaline levels, nutrient accessibility, and the prevention of pore clogging. Including all relative density and topologies, the pore size of the structures evaluated were between 130 microns and 1016 microns, with an average pore size of 500 microns (Figure E1). This respects the limit range of 100 microns minimum and 1000 microns maximum and the optimal pore size of between 400 microns and 600 microns, discussed in Chapter 2.

The proposed ideal porosity for bone growth in the literature varies within each study. However, it is commonly mentioned that porosity should be the closest to trabecular bone. It was shown that under a relative density of more than 60%, the ability of cells to invade the structure is prohibited [113]. For the characterization, relative density of 5%; 10%; 15%; 20%; 25%; 30%; 35%; 40%; 45% and 50% were assessed. As for the topology, the fourteen strut-based lattices described in detail in Chapter 2 were investigated (Figure 34). Based on the convergence analysis describe in Chapter 3, the number of elements for the characterization was set at around 300,000 to 400,000, depending on the strut size and topology.

Table 13. Lattice characteristics at the lowest and highest relative density investigated (5% and 50%) corresponding to pore sizes between 130 and 1016 micron.

Topology	Relative Density (%)	Strut Diameter (micron)	Pore Size (micron)	Surface Area (mm ²)	E in direction of applied pressure (GPa)
Body Center Cubic (X-Shape)	5	122	880	45.0634	0.05
	50	446	510	107.2641	15.21
Cubic	5	184	1016	45.0091	2.24
	50	680	520	107.0561	33.94
Cubic Center (Star)	5	100	530	60.8206	0.85
	50	376	260	138.7648	22.11
Diamond	5	N/A	742	45.67463	N/A
	50	N/A	437	121.6855	N/A
Octet	5	78	390	78.6663	0.86
	50	288	214	105.5916	20.43
Rhombic Dodecahedron (H)	5	106	431	57.8635	0.04
	50	402	130	130.2158	11.98
Rhombic Dodecahedron (V)	5	106	431	57.8635	0.29
	50	402	130	130.2158	27.77
Rhombicuboctahedron	5	82	415	68.8192	0.81
	50	324	173	140.8783	26.30
Tesseract	5	90	526	67.1933	0.79
	50	328	341	157.0839	23.25
Tetrahedron	5	96	607	68.7733	0.89
	50	352	351	164.5839	18.03
Truncated Cube	5	126	1005	53.5681	1.09
	25	326	805	87.4212	N/A
Truncatedcuboctahedron	5	90	667	65.3948	0.32
	40	312	445	126.6924	N/A
X-Cross Grid (FBCCXYZ)	5	74	417	86.8552	0.88
	50	274	220	196.6563	20.18
Vintile	5	110	630	55.2029	0.30
	50	424	326	118.5522	31.38

5.3 - Results

5.3.1 Bone Percentage Comparison Between Topologies

Overall

Regardless of the pressure applied and the topology used, better environment for bone is created when the relative density is between 10% and 30%. (Figures 36-39) Under pressure of 0.5 MPa the choice of topology makes a big difference on its capacity to grow bone (Figure 36).

Under the highest pressures of 2 MPa the choice of the topology matters less and most perform well with a relative density between 10% and 30% (Figure 39).

Figure 36. Percentage of the void space having optimal stimulation for bone growth in function of the relative density of the 14 topologies, under a pressure of 0.5 MPa

Figure 37. Percentage of the void space having optimal stimulation for bone growth in function of the relative density of the 14 topologies, under a pressure of 1 MPa

Figure 38. Percentage of the void space having optimal stimulation for bone growth in function of the relative density of the 14 topologies, under a pressure of 1.5 MPa

Figure 39. Percentage of the void space having optimal stimulation for bone growth in function of the relative density of the 14 topologies, under a pressure of 2 MPa. The data point

Topology Specific

Figure 34 contains 3 plots for each of the 4 pressures applied. The first plots representing a low relative density of 10% clearly shows that every topology apart from Rhombic Dodecahedron (H) was allowing almost 100% of its void space to promote bone growth, regardless of the pressure applied. As for Rhombic Dodecahedron H, at a relative density of 10% its average biophysical stimulation in each element is almost four times higher than the second-highest BS average (Figure F16). The next plot, containing all results obtained with medium relative density (30%) shows that Rhombic Dodecahedron (H), X-Shape and Diamond are displaying the highest percentages. Moreover, at lower pressures Tesseract does stand out from the other topologies. In the third plot, looking at a high relative density (50%), Diamond, X-Shape, Rhombic (V) and Tesseract are still outstanding with the addition of Tetrahedron, showing good results (Figure F19). Under most relative density and pressures, Cubic, Vintile, Rhombic (V) and Rhombicuboctahedron performed the worst.

Figure 40. Percentage of the void space having optimal stimulation for bone growth in function of the relative density of the 14 topologies, under pressures of 0.5 MPa, 1 MPa, 1.5 MPa and 2 MPa, for low (10%), medium (30%) and high relative density (50%)

5.3.2 Fluid Flow Velocity and Octahedral Shear Strain Distribution

Figures showing the strain and velocity part of the BS equation, as well as the elements that had BS within the optimal range, were generated for each topology to compare the distribution of strain and velocity (Appendix from F18 to F39). The images were taken from all data points collected at a relative density of 40% under a load of 2 MPa. Under those parameters, every topology showed different bone growth amount. While all structures had higher stimulation on the periphery and lower stimulation on the inside, the difference seems bigger in some topologies. The topologies showing the best bone results such as Tesseract (Figure F38) and Diamond (Figure 41) have a smaller difference between the velocity and strain found at the periphery and the symmetry plane. Topologies with more bone also showed higher velocity flow at the symmetry plane, such as X-shape (Figure F25) and Tesseract (Figure F38). As for structures with minimal fluid flow velocity in the symmetry plane, such as for Cubic (Figure 42), there is a weak biophysical stimulation in the center of the lattice (Figure 42).

5.3.3 Distribution of Elements Optimal for Bone Ingrowth

A second set of visualization figures were developed for each topology to compare the location of the optimal elements for bone inside the lattice. The images were taken on the symmetry plane with: high relative density plus low pressure, high relative density plus high pressure, low relative density plus low pressure and low relative density and high pressure (Figures 43, 44 and 45). In Figures 44 and 45, both Rhombic Dodecahedron (Horizontal) and X-Shape shows that for a structure with a low relative density, under high pressure, the stimulation gets too high in the corners and around the strut for bone to adhere and grow. This leads to biologically irrelevant solutions when most optimal elements have no point of contact with the strut. The opposite is shown in the structures with high relative density and low pressure, bone is promoted to grow mostly in the crease and around some struts (Figure 44 and 45).

Figure 41. Distribution of the velocity and the strain part of the biophysical stimulus, the Diamond sample at 40% relative density, under a pressure of 2 MPa

Figure 42. Distribution of the velocity and the strain part of the biophysical stimulus, the Cubic sample at 40% relative density, under a pressure of 2 MPa

Figure 43. Elements with optimal stimulus for bone growth, within a structure made of Cubic topology, with 10% and 40% relative density, under a pressure of 0.5 MPa and 2 MPa

Figure 44. Elements with optimal stimulus for bone growth, within a structure made of Rhombic Dodecahedron (Horizontal) topology, with 10% and 40% relative density, under a pressure of 0.5 MPa and 2 MPa

Figure 45. Elements with optimal stimulus for bone growth, within a structure made of X-Shape topology, with 10% and 40% relative density, under a pressure of 0.5 MPa and 2 MPa

5.4 - Discussion

Overall, following the characterization X-shape, Diamond, Tesseract and Rhombic (H) topologies were leading to better environment to grow bone. Cubic, Truncated Cube, Rhombic (V) and Vintile were the topologies the least favourable for bone growth stimulation.

5.4.1 Stiffness and Bone ingrowth

Bone ingrowth performance was found to be highly related to the structures' stiffness in the direction of the applied load. It should be considered one of the most important characteristics. Ti6Al4V itself has a significantly higher stiffness compared to both the bone and the polymer evaluated in Chapter 4. The lattice with higher stiffness reduces the strain and the fluid movement inside the lattice, consequently reducing the BS obtained. Some geometries such as Vintile (Figure F23) and Rhombic(V) (Figure F27), provide a weak environment for bone

growth, even if they have good homogenous stimulus distribution of velocity and strain. This is explained by their structures that are too stiff to have strains and velocities high enough for stimulation. Also, the worst topologies are Cubic (Figure 42), Rhombicuboctahedron (Figure F29) and Truncated cube (Figure F35) since their stiffness is too high and the strain and velocity distribution are heterogenous.

The lower stiffness of Rhombic Dodecahedron (H) compared to all the other topologies made it perform poorly with lower density under high loads. The results from the Rhombic Dodecahedron (H) topology compared with its own topology oriented vertically are completely different (Figure F27 and F26). The main difference is explained by its structural stiffness being 2.3 times higher in the vertical orientation (Table 13). Compared to all the topologies, Rhombic Dodecahedron (H) led to one of the best environments to grow bone, while Rhombic Dodecahedron (V) was one of the worst (Figure 36-39). Rhombic Dodecahedron (H) had the lowest Young's modulus and Rhombic Dodecahedron (V) had the 3 highest moduli of all topologies. Rhombic Dodecahedron (V) could be the proper choice for a more compliant or for lower relative density under higher loads.

This highlights the importance of creating a result database for each specific material to design implants. The difference between the polymer results (Figure 23) and the titanium results (Figure 46) is a good example. The polymer being more compliant than Ti6Al4V, consequently led to relatively high strain and velocity causing higher BS. In that case a Rhombic Dodecahedron (V) was better suited than the Rhombic Dodecahedron (H), at any density under any pressure since it prevented having a too high BS.

Figure 46. Percentage of the void space having optimal stimulation for bone growth in function of the relative density for Rhombic Dodecahedron (V) in light green and Rhombic Dodecahedron (H) in dark green under pressures of 0.5 MPa, 1 MPa, 1.5 MPa and 2 MPa

5.4.2 Fluid flow velocity

Allowing the fluid to pass and exit on the periphery leads to higher velocity on the outside of the structure, which explains the higher BS in elements at the periphery. Since all the topologies have the same constraint, their results are comparable within each other. In the human environment the same phenomenon would be present within the implant. The strut arrangement must provide an efficient flow, else bone will likely grow on the periphery and not inside the lattice. A study evaluated the permeability of cages with constant stiffness and different porosity found that a titanium cage with 35% relative density was matching the most the permeability of healthy cancellous bone [159]. Another study showed that fluid velocity gets higher in small pores of lattices and with smaller strut diameters [42]. This phenomenon can be observed in Figure F26 on the periphery images for Rhombic Dodecahedron (H) and in Figure F29 for

Rhombicuboctahedron. It also explains why stimulus is higher in those regions, thus favouring bone growth. Finally, it is important to have a good distribution of the stimulation which is mostly affected by the fluid velocity. If there is a heterogeneous distribution of the stimulus, it will result in some regions being highly successful for bone ingrowth and others will remain without any bone.

5.4.3 Lattice Response as Bone Grows

With relative density between 15% and 30% most topologies have 100% of the void space optimal for bone ingrowth. However, it is important for a structure to also perform well at higher relative density. As the bone grows, the structure will stiffen and increase in density. Then, even if the stimulus cells perceive gets lower, it will still be high enough for cells to be stimulated and keep growing. In the situation where the stimulus is optimal but already in the lower range, as soon as bone grows a little bit, cells will not have enough stimulation to deposit bone.

5.4.4 Comparison with previous study

For the canine *in vivo* study, it was shown that octet would be better suited for bone ingrowth compared with tetrahedron. With the replication of their geometrical parameters the model was able to show a similar trend, in Chapter 4. In fact, even with a different unit cell length and pore size, if you look at Figure 41 you would also see that when tetrahedron has a relative density of 45% compared to an octet with a relative density of 30%, octet still shows higher bone growth at any pressure applied. However, if the study would have compared the two topologies with the exact same relative density, they would have found that tetrahedron is slightly better under almost all loads and relative density.

Figure 47. Percentage of the void space having optimal stimulation for bone growth as a function of the relative density for Octet in light blue and Tetrahedron in pink under pressures of 0.5 MPa, 1 MPa, 1.5 MPa and 2 MPa

5.5 - Limitation

Numerous limitations were discussed throughout the Chapters. Following Prendergast et al.,[4] model, bone stimulation is calculated from the fluid velocity and the octahedral shear strain. The measurement of octahedral shear strain implicates both the deformation from tension and compression. However, compression of the cells plays a bigger role than tension, in the bone remodeling and regeneration process [62]. The experimental research leading to the main biophysical stimulus equation was based on cellular differentiation response in canine. The specific stimulus activating human MSCs could vary from the canine as well as between individuals, depending on its genetic. Furthermore, only one model was explored in this thesis, the results could be slightly different using another well-established model such as Cleas et Heigele's method [66].

The omission of growth factors presence, specialised cell diffusions, angiogenesis and the genetic variability implications are some of the biggest factors limiting the reproduction of bone ingrowth *in silico*. Removing the cell diffusion and time iteration processes, as well as neglecting the vascularization and growth factor capacity, affects the estimated location of bone growth within a lattice structure. It would be ideal if the model could implicate the diffusion process of cells without increasing the computation time too much. Currently, the percentage values measured for bone ingrowth can only be used as a reference to compare topologies evaluated with the same method. A comparison of the exact percentages collected with the model to the percentages of bone *in vivo* cannot be made directly. Also, only mature bone stimulation was considered. The evolution of tissue changes between, immature bone, fibrous and cartilage tissue were overlooked in this project, even though they all have the potential to become bone through processes like endochondral and desmoid ossification. In fact, it would be possible to calculate the volume of the elements having BS in the respective range of those tissues. The benefits of collecting such results are unknown.

A compromise between computation time and accuracy was made to enable the collection of sufficient results, in the project timeframe. It is possible and recommended to use more elements in future data collection, especially for comparisons made on more specific features, such as cross-section shape. Additionally, all results were from the evaluation of 4-by-4-by-4 lattices. The effect of a bigger lattice on the trends observed could be further explored. Furthermore, from Chapter 4 it was possible to see that the simulation amplifies the difference between the topology's capacity to grow bone. While the performance of the best topologies could be slightly overestimated, the performance of the worst topologies could also be slightly underestimated. However, the conclusion whereas a topology is better than another under specific parameters remains reliable.

CHAPTER 6 - CONCLUSION

6.1 - Summary

This thesis's goal was to find the best lattice to promote bone ingrowth with a focus on spinal cage load. This was done using a simplified mechano-regulator method to characterize different lattice topologies at different relative densities. The method utilized to quantify bone was supported with both *in silico* and *in vivo* research data. All the data compared with those studies were leading to the same conclusion in terms of topology, cross-section shape and material studied. After the characterization, the following criteria were leading to an optimal lattice structure maximizing the mechanical stimulation for bone ingrowth performance; a structure with a homogenous distribution of the octahedral shear strain; a structure that allows enough flows in the centre of the lattice; a stiffness allowing enough deformation to create fluid movement to stimulate the cells; a topology that performs well under numerous relative density, and a topology that has pore size allowing blood vessel growth and prevent pore clogging. X-Shape, Rhombic Dodecahedron (H), Tesseract and Diamond are lattices filling most of those criteria.

6.2 - Contribution

This thesis presented a 3D mechano-regulatory-based method to estimate the percentage of void space inside a lattice structure that predicts optimal biophysical stimulation for bone to grow. The database collected for each topology, pressure and relative density provides valuable information to improve the design of Ti6Al4V lattice-based spinal cage implants. The method described with the scripts provided in the appendix, can be used to collect more bone estimation for other topologies, cross-sections, anisotropy and materials, under different loads. Lastly, even though this research was focusing on the application of the lattice in LIF cage implant, the results

could also be valuable for other implants such as cervical implants when looking at the data collected at lower pressure.

6.3 - Future Work

From the resulting lattice visuals, a pattern in the velocity and strain could be seen between unit cells. A smaller lattice could be sufficient to evaluate bone growth performance. If the usage of a smaller lattice gives the same results as a 4-by-4-by-4 lattice, it could help reduce the computation time. As for the strain-time and velocity-time response discussed in Chapter 3, it would be useful to conduct a sensitivity analysis for a longer loading time or during cyclic loading, to explore the bone stimulation behaviour during more sedentary daily activities.

Eleven more strut-base topologies are set up for their addition in the database. The performance of TPMS cellular materials could also be interesting to investigate. While lattice with square cross-sections were showing worst performance than a circle, in the polymer study, with the usage of Ti6Al4V the response could be completely different. This leads to a possible study based on shape transformers. Additionally, X-shape, Cubic Center and Octet have higher stiffness in the diagonal direction and Cubic, Tesseract and Vintile have lower stiffness in the diagonal direction. It would be interesting to make a direct comparison with the model and see how much their bone ingrowth potential would change.

The optimal lattice topology suggested were solely based on the mechanical stimulation. While more compliant structures such as Rhombic Dodecahedron (H) and Diamond are better suited for bone ingrowth, before applying them in spinal cage implants mechanical testing should be done to confirm that they are strong enough to support the loads.

Since implants differ in size and patients differ in weight, the pressure applied to the implant can vary significantly from one case to another. If the conception of an implant has the

goal of being used in a variety of patients, a structure that behaves at its best under a wide range of pressure should be chosen. The data collected could also be part of a multi-scale/multi-objective design optimization. In terms of cell seeding, the hypothetical ideal structure would be a structure that has smaller pore size and porosity inside the structure and bigger pore size on the periphery of the implant. However, this would not be good for a LIF implant since the vertebrae itself gets denser as it reaches the periphery [91], [92]. This means that if you want to match the bone stiffness, you will end up with denser structure on the periphery of the implant. Finally, this issue comes back to the need for a multi-scale objective design to find the optimal balance.

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Appendix A - Structure Characteristic

Table A1. Pore size estimation for 1200 microns unit cells, with 5 and 50% relative density

Body Center Cubic (X-Shape)

Cubic
$T = a - D$ $T = 1200 - 184$ Pore Size = 1016	$T = a - D$ $T = 1200 - 680$ Pore Size = 520
Cubic Center (Star)

Diamond

Octet

<p>Circle diameter 0.3908mm Circular perimeter 1.2279mm</p>	<p>Circle diameter 0.2138mm Circular perimeter 0.6717mm</p>
Rhombic Dodecahedron
<p>Circle diameter 0.4306mm Circular perimeter 1.3526mm</p>	<p>Circle diameter 0.1315mm Circular perimeter 0.4131mm</p>
Rhombicuboctahedron
<p>Circle diameter 0.4151mm Circular perimeter 1.3039mm</p>	<p>Circle diameter 0.1731mm Circular perimeter 0.5437mm</p>
<p>Circle diameter 0.2164mm Circular perimeter 0.68mm</p>	<p>0 Pore</p>	

Tesseract

Tetrahedron
$T = (2 - \sqrt{2})(a) - D$ $T = (2 - \sqrt{2})(1200) - 352$ Pore Size = 351	$T = (2 - \sqrt{2})(a) - D$ $T = (2 - \sqrt{2})(1200) - 96$ Pore Size = 607
Truncated Cube
<p>0 Pore</p>

Truncated Cuboctahedron

0 Pore

X-Cross Grid (FBCCXYZ)

Vintile

Appendix B - Convergence Analysis

Figure B1. Element stimulus average within a 2x4x4 Diamond lattice with unit cells of 637 microns and strut diameter of 100 microns, made of pyroclastic polymer strut. Convergence analysis made in function of the number of elements used. Strain/0.0375 and velocity/3 are the scaling factors added together to get the BS

Figure B2. Bone ingrowth within a 2x4x4 Diamond lattice with unit cells of 637 microns and strut diameter of 100 microns, made of pyroclastic polymer strut. Convergence analysis made in function of the number of elements used

Appendix C - *In Silico* Paper Replication

Table C1. Comparison of the strut diameter maximizing the percentage of the total volume that can stimulate bone based on the current simulation with data published by Rodriguez et al. 2018 [14] for the Diamond structure

	Diameter (micron)	Bone Growth (%)	Diameter (micron)	Bone Growth (%)
Pressure (MPa)	Circle Replication	Circle Replication	Circle Rodriguez et al.	Circle Rodriguez et al.
0.05	120	82.11	90	90
0.1	145	75.11	110	85
0.15	160	69.36	131	79
0.25	190	62.86	160	69
0.5	226	50.84	200	52
1	290	33.20	240	33
1.5	305	23.65	260	22
Pressure (MPa)	Square Replication	Square Replication	Square Rodriguez et al.	Square Rodriguez et al.
0.05	77	80.47	72	90
0.1	95	72.25	95	85
0.15	110	66.82	112	79
0.25	126	58.07	133	69
0.5	150	45.87	165	51
1	177	28.16	184	30
1.5	187	16.42	184	14

Table C2. Comparison of the strut diameter maximizing the percentage of the total volume that can stimulate bone based on the current simulation with data published by Rodriguez et al. 2018 [14], for the Truncated Cube structure

	Diameter (micron)	Bone Growth (%)	Diameter (micron)	Bone Growth (%)
Pressure (MPa)	Circle Replication	Circle Replication	Circle Rodriguez et al.	Circle Rodriguez et al.
0.05	100	85.43	70	95
0.1	130	77.77	93	90
0.15	170	71.33	120	85
0.25	170	60.64	152	79
0.5	172	25.90	152	43
1	172	2.07	152	10
1.5	172	0.13	152	0
Pressure (MPa)	Square Replication	Square Replication	Square Rodriguez et al.	Square Rodriguez et al.
0.05	65	85.82	60	95
0.1	87	78.27	82	90
0.15	87	69.76	105	85
0.25	87	46.69	109	72
0.5	87	9.79	109	28
1	87	0.11	109	3
1.5	87	0.01	109	0

Table C3. Comparison of the strut diameter maximizing the percentage of the total volume that can stimulate bone based on the current simulation with data published by Rodriguez et al. 2018 [14] for the Rhombic Dodecahedron with horizontal orientation structure

	Replication	Replication	Rodriguez et al.	Rodriguez et al.
Pressure (MPa)	Diameter (micron)	Bone Growth (%)	Diameter (micron)	Bone Growth (%)
0.05	152	68.39	120	77
0.1	168	61.53	153	68
0.15	184	55.55	168	60
0.25	202	48.70	192	51.5
0.5	232	37.86	231	37
1	258	28.03	262	20
1.5	272	19.45	280	11

Table C4. Comparison of the strut diameter maximizing the percentage of the total volume that can stimulate bone based on the current simulation with data published by Rodriguez et al. 2018 [14], for the Rhombic Dodecahedron with vertical orientation structure

	Replication	Replication	Rodriguez et al.	Rodriguez et al.
Pressure (MPa)	Diameter (micron)	Bone Growth (%)	Diameter (micron)	Bone Growth (%)
0.05	94	86.79	70	92
0.1	110	81.65	88	89
0.15	125	76.59	98	87
0.25	141	71.12	109	81
0.5	170	59.66	132	68
1	214	42.74	175	41

Figure C1. Comparison of the optimized strut diameter maximizing the percentage of the total volume that can stimulate bone based on the current simulation with data published by Rodriguez et al. 2018 [10], the structure made of polymer is tessellated with Diamond, Truncated Cube, Rhombic Dodecahedron (H) and Rhombic Dodecahedron (V) unit cells, with circle cross section, under pressures of 0.05 MPa, 0.1 MPa, 0.15 MPa, 0.5 MPa, 1 MPa and 1.5 MPa

Table C5. Comparison of the strut diameter maximizing the percentage of the total volume that can stimulate bone based on the current simulation with data published by Boccaccio et al. 2018 [15], Rhombicuboctahedron structure made of polymer with different material young's modulus

Young's Modulus of 500MPa				
Pressure (Mpa)	Boccaccio et al. Diameter (micron)	Boccaccio et al. Bone (%)	Replication Diameter (micron)	Replication Bone (%)
0.05	59	90	95	78.10
0.1	82	82	109	68.60
0.5	152	48	170	41.84
1	152	20	170	17.16
1.5	152	6	170	2.38
Young's Modulus of 1000MPa				
Pressure (Mpa)	Boccaccio et al. Diameter (micron)	Boccaccio et al. Bone (%)	Replication Diameter (micron)	Replication Bone (%)
0.05	48	94	75	85.04
0.1	65	89	92	77.65
0.5	125	62	143	53.43
1	152	43	170	41.57
1.5	152	27	170	30.53
Young's Modulus of 1500MPa				
Pressure (Mpa)	Boccaccio et al. Diameter (micron)	Boccaccio et al. Bone (%)	Replication Diameter (micron)	Replication Bone (%)
0.05	40	94	67	88.21
0.1	58	90	85	81.48
0.5	108	70	135	58.79
1	141	53	159	49.80
1.5	152	42	170	39.18

Appendix D - *In Vivo* Paper Replication

Rabbit *in vivo* validation test:

Table D1. Element with optimal environment for bone growth, under 0.4 MPa of pressure

Cubic – Center of lattice (symmetry side)	Cubic – Outside of lattice (pressure = 0)

Diamond – Center of lattice (symmetry side)	Diamond – Outside of lattice (pressure = 0)

Tetrahedron – Center of lattice (symmetry side)	Tetrahedron – Outside of lattice (pressure = 0)

Table D2. Octahedral Shear Strain/0.0375, representing the strain part of the biophysical stimulus, under 0.4 MPa of pressure

Table D3. Octahedral Shear Strain/0.0375, representing the strain part of the biophysical stimulus, under 0.5 MPa of pressure

Table D4. Element with optimal environment for bone growth, under 0.6 MPa of pressure

Cubic – Center of lattice (symmetry side)	Cubic – Outside of lattice (pressure = 0)

Diamond – Center of lattice (symmetry side)	Diamond – Outside of lattice (pressure = 0)

Tetrahedron – Center of lattice (symmetry side)	Tetrahedron – Outside of lattice (pressure = 0)

Table D5. Octahedral Shear Strain/0.0375, representing the strain part of the biophysical stimulus, under 0.6 MPa of pressure

Table D6. Element with optimal environment for bone growth, under 0.7 MPa of pressure

Cubic – Center of lattice (symmetry side)	Cubic – Outside of lattice (pressure = 0)

Diamond – Center of lattice (symmetry side)	Diamond – Outside of lattice (pressure = 0)

Tetrahedron – Center of lattice (symmetry side)	Tetrahedron – Outside of lattice (pressure = 0)

Table D7. Octahedral Shear Strain/0.0375, representing the strain part of the biophysical stimulus, under 0.7 MPa of pressure

Canine *in vivo* validation test:

Table D8. Octahedral Shear Strain/0.0375 representing the strain part of the biophysical stimulus, and element with optimal environment for bone growth, under 0,5 MPa, 0.625 MPa, 0.75 MPa, 0.875 MPa and 1 MPa of pressure, for the Octet and Tetrahedron structures.

Octet under 0.5MPa		Tetrahedron under 0.5MPa	
Optimal for Bone	Strain/0.0375	Optimal for Bone	Strain/0.0375

Octet under 0.625MPa		Tetrahedron under 0.625MPa	
Optimal for Bone	Strain/0.0375	Optimal for Bone	Strain/0.0375

Octet under 0.75MPa		Tetrahedron under 0.75MPa	
Optimal for Bone	Strain/0.0375	Optimal for Bone	Strain/0.0375

Octet under 0.875MPa		Tetrahedron under 0.875MPa	
Optimal for Bone	Strain/0.0375	Optimal for Bone	Strain/0.0375

Octet under 1MPa		Tetrahedron under 1MPa	
Optimal for Bone	Strain/0.0375	Optimal for Bone	Strain/0.0375

Appendix E - Titanium Lattice Characterization Data

Figure E1. Pore size of the topologies evaluated, with 5% and 50% relative density

Figure E2. Strut diameter of the topologies evaluated, with 5% and 50% relative density

Figure E3. Surface Area for a 2x2x4 lattice structure made of 1200 microns unit cells, with 5% and 50% relative density

Figure E4. Young's modulus (GPa) in the z direction with 5% relative density for different topologies

Figure E5. Young's modulus (GPa) in the z direction with 50% relative density for different topologies

Table E1. Percentage of pores with optimal environment to grow bone and structure characteristics under pressures of 0.5 MPa, 1 MPa, 1.5 MPa and 2 MPa

Octet	Radius (micron)	Relative Density (%)	Porosity (%)	Surface Area (mm ²)	Bone growth within pores (%)			
					0.5 MPa	1 MPa	1.5 MPa	2 MPa
0	39	5.1009	94.8991	78.6663	99.548289	98.609574	94.83456	86.01917
10	56	10.0403	89.9597	105.5916	99.399263	99.687104	99.216589	98.98992
15	70	15.0747	84.9253	124.4077	89.669465	99.825824	99.92234	99.54593
20	82	19.9647	80.0353	138.1220	58.132426	98.322117	99.847622	99.96025
25	93	24.8297	75.1703	148.7361	28.087799	88.250583	99.066532	99.80426
30	104	29.9869	70.0131	157.4779	9.98701	57.451348	93.312018	99.12093
35	114	34.8689	65.1311	163.8004	3.944813	34.51129	78.565687	96.04896
40	124	39.8802	60.1198	168.5756	1.838856	20.129693	49.013852	84.84997
45	134	44.9665	55.0335	171.8034	1.337459	8.646519	28.196966	56.77324
50	144	50.0744	49.9256	173.4840	1.115664	5.736921	21.524437	39.32287
Star	Radius (micron)	Relative Density (%)	Porosity (%)	Surface Area (mm ²)	Bone growth within pores (%)			
					0.5 MPa	1 MPa	1.5 MPa	2 MPa
5	50	4.9147	95.0853	60.8206	99.78333	98.82072	96.44395	91.2579
10	73	9.9855	90.0145	82.9786	98.41209	99.87204	99.48239	99.0724
15	91	14.9204	85.0796	97.7618	81.38228	99.595	99.94714	99.73028
20	107	19.8951	80.1049	109.0170	49.86436	95.77682	99.73471	99.935
25	122	24.9706	75.0294	117.9570	28.40478	82.9971	97.83171	99.75823
30	136	29.9939	70.0061	124.8938	14.96874	57.29833	90.49895	98.13387
35	149	34.8469	65.1531	130.1186	5.72717	31.1439	68.78982	91.87885
40	162	39.8273	60.1727	134.1719	3.600173	22.63033	48.71629	78.87162
45	175	44.8822	55.1178	137.0544	1.664966	14.32944	29.40023	53.75206
50	188	49.9590	50.0410	138.7648	1.404551	9.77072	22.15848	37.29609
Vintile	Radius (micron)	Relative Density (%)	Porosity (%)	Surface Area (mm ²)	Bone growth within pores (%)			
					0.5 MPa	1 MPa	1.5 MPa	2 MPa
5	55	5.0189	94.9811	55.2029	99.45327	96.41746	84.48021	57.46622
10	80	10.0599	89.9401	74.6331	99.7322	99.85837	99.3532	98.99956
15	100	15.0203	84.9797	87.6291	86.89859	99.51958	99.95044	99.7344
20	118	20.0391	79.9609	97.3890	42.11802	92.18811	98.96465	99.72906
25	135	25.1476	74.8524	104.9221	17.78479	67.79875	91.93748	98.19633
30	150	29.8677	70.1323	110.2102	7.583868	45.37123	74.31963	91.40999
35	166	35.0398	64.9602	114.4465	2.959518	26.34389	52.95492	72.73347
40	181	39.9425	60.0575	117.1015	1.523394	13.95332	34.1657	53.23227
45	197	45.1480	54.8520	118.5292	1.270468	5.458158	20.25202	34.36959
50	212	49.9311	50.0689	118.5522	99.45327	96.41746	84.48021	57.46622
X Cross Grid	Radius (micron)	Relative Density (%)	Porosity (%)	Surface Area (mm ²)	Bone growth within pores (%)			
					0.5 MPa	1 MPa	1.5 MPa	2 MPa
5	37	4.9995	95.0005	86.8552	99.683822	98.852804	95.21292	87.392504
10	54	10.1599	89.8401	118.3896	99.616578	99.798214	99.443157	99.138653
15	67	15.0645	84.9355	138.9472	86.835822	99.896385	99.960271	99.750502
20	79	20.2047	79.7953	155.1876	50.760476	97.922565	99.929706	99.983194
25	89	24.8617	75.1383	166.7148	24.328753	87.974229	99.317007	99.927139
30	99	29.7951	70.2049	176.4180	9.049765	60.99889	94.444762	99.491364
35	109	34.9455	65.0545	184.2972	4.446996	33.397605	78.920962	96.166231
40	119	40.2537	59.7463	190.3527	2.965776	16.667293	55.48475	86.001797
45	128	45.1172	54.8828	194.2430	2.302729	9.046338	35.408434	69.552726
50	137	50.0176	49.9824	196.6563	1.497718	5.039176	17.898318	46.389031
Cube	Radius (micron)	Relative Density (%)	Porosity (%)	Surface Area (mm ²)	Bone growth within pores (%)			
					0.5 MPa	1 MPa	1.5 MPa	2 MPa
5	92	5.0299	94.9701	45.0091	98.14339	99.91669	99.36909	98.27116
10	133	10.0374	89.9626	61.2338	69.35691	98.18031	99.70986	99.79428

15	166	15.0410	84.9590	72.5758	37.04397	86.45169	98.0082	99.51107
20	195	20.0334	79.9666	81.2789	20.78005	67.72004	89.22743	97.63978
25	222	25.0941	74.9059	88.3187	13.39292	48.17744	80.38917	91.05704
30	247	30.0656	69.9344	93.9231	8.489366	31.68237	63.02251	82.74943
35	271	35.0382	64.9618	98.4765	5.915572	22.41472	46.13414	70.7558
40	294	39.9364	60.0636	102.0802	4.296635	15.80549	31.86039	53.04761
45	317	44.9162	55.0838	104.9401	2.96044	11.91617	22.79067	38.29409
50	340	49.9296	50.0704	107.0561	2.226452	9.162082	16.56621	27.48205
Tesseract	Radius (micron)	Relative Density (%)	Porosity (%)	Surface Area (mm ²)	Bone growth within pores (%)			
					0.5 MPa	1 MPa	1.5 MPa	2 MPa
5	45	5.0282	94.9718	67.1933	99.534831	96.480056	87.627988	74.229971
10	65	10.0474	89.9526	91.2085	99.130956	99.640719	98.771251	97.291184
15	81	15.0517	84.9483	107.8293	94.33425	99.512922	99.797056	99.289003
20	95	20.0418	79.9582	120.4830	81.215904	97.685574	99.481094	99.844855
25	108	25.1066	74.8934	130.6537	59.717689	92.800813	98.090678	99.350111
30	120	30.0890	69.9110	138.6924	37.333584	83.770948	94.995062	98.026582
35	131	34.8673	65.1327	144.9230	22.20715	72.423214	89.44373	95.72652
40	142	39.8051	60.1949	150.0649	14.270527	57.65116	81.129722	91.208675
45	153	44.8597	55.1403	154.1181	8.128566	37.456626	67.26734	82.837133
50	164	49.9897	50.0103	157.0839	6.110425	25.632015	52.284195	72.306434
X-Shape	Radius (micron)	Relative Density (%)	Porosity (%)	Surface Area (mm ²)	Bone growth within pores (%)			
					0.5 MPa	1 MPa	1.5 MPa	2 MPa
5	61	5.1095	94.8905	45.0634	94.21297	43.98525	20.64054	11.51724
10	88	10.1594	89.8406	61.2292	99.97162	98.84097	94.30827	81.22326
15	109	15.0209	84.9791	72.1988	99.54729	99.9513	99.80228	98.7281
20	128	20.0080	79.9920	80.9145	96.40785	99.86882	99.95222	99.88496
25	146	25.1606	74.8394	88.1118	84.66937	99.04136	99.8435	99.93667
30	162	30.0251	69.9749	93.6439	61.3836	96.01136	99.41901	99.8332
35	178	35.0991	64.9009	98.3628	27.41915	85.3244	97.31201	99.32603
40	193	39.9967	60.0033	102.0459	12.14449	66.33331	92.54352	98.14242
45	208	44.9835	55.0165	105.0130	5.376793	44.74114	78.81287	94.05684
50	223	50.0136	49.9864	107.2641	2.620777	23.08135	56.11097	80.70421
Tetrahedron	Radius (micron)	Relative Density (%)	Porosity (%)	Surface Area (mm ²)	Bone growth within pores (%)			
					0.5 MPa	1 MPa	1.5 MPa	2 MPa
5	48	4.9249	95.0751	68.7733	99.785452	98.709826	96.486994	91.823609
10	70	10.0287	89.9713	94.3071	98.645993	99.843743	99.476835	98.963949
15	87	14.9597	85.0403	111.4603	82.217373	99.629642	99.943426	99.682757
20	102	19.9183	80.0817	124.7291	53.421116	97.037203	99.755504	99.965295
25	116	24.9831	75.0169	135.5350	29.015536	86.03285	98.819205	99.828526
30	129	30.0029	69.9971	144.2044	14.268628	67.377275	94.321918	99.239133
35	141	34.8589	65.1411	151.0405	8.415379	47.674143	84.074722	96.989499
40	153	39.8844	60.1156	156.7571	4.686402	28.375624	67.421709	89.805702
45	165	45.0365	54.9635	161.3538	3.225563	18.289034	51.568971	79.137663
50	176	49.8340	50.1660	164.5839	2.236821	12.417672	38.801477	67.483704
Truncated Cube	Radius (micron)	Relative Density (%)	Porosity (%)	Surface Area (mm ²)	Bone growth within pores (%)			
					0.5 MPa	1 MPa	1.5 MPa	2 MPa
5	63	4.9061	95.0939	53.5681	99.946113	98.632253	96.513799	92.939863
10	92	9.6199	90.3801	69.8500	96.410544	99.976033	99.708724	99.159907
15	120	14.9829	85.0171	80.5595	64.288925	97.234833	99.680294	99.837957
20	145	20.0705	79.9295	85.9617	30.738783	81.558331	96.119468	99.253867
25	163	23.7208	76.2792	87.4212	21.660201	68.260597	88.511618	97.365161
Truncated Cuboctahedron	Radius (micron)	Relative Density (%)	Porosity (%)	Surface Area (mm ²)	Bone growth within pores (%)			
					0.5 MPa	1 MPa	1.5 MPa	2 MPa
5	45	4.8874	95.1126	65.3948	98.908473	92.762135	70.489103	35.654923
10	67	10.1279	89.8721	88.8522	99.762868	99.758644	99.21257	98.36433

15	84	15.0613	84.9387	103.1494	92.270172	99.863455	99.877531	99.64133
20	99	19.8693	80.1307	112.9924	56.195304	96.570784	99.830945	99.938593
25	114	24.9519	75.0481	120.2364	26.536291	78.780779	96.285064	99.516875
30	129	30.1645	69.8355	124.8811	13.81791	47.265892	81.299988	95.023321
35	143	35.0186	64.9814	126.8715	5.970419	24.953597	51.340384	76.381517
40	156	39.4119	60.5881	126.6924	98.908473	92.762135	70.489103	35.654923
Rhombicuboctahedron	Radius (micron)	Relative Density (%)	Porosity (%)	Surface Area (mm ²)	Bone growth within pores (%)			
					0.5 MPa	1 MPa	1.5 MPa	2 MPa
5	41	4.8924	95.1076	68.8192	99.829656	98.585773	95.016684	88.856799
10	60	9.9056	90.0944	92.9412	97.665928	99.874233	99.46981	98.927436
15	76	15.1202	84.8798	109.4376	74.461923	98.351269	99.852752	99.863154
20	89	19.8742	80.1258	120.2716	31.897556	88.876744	98.407378	99.719993
25	102	24.9725	75.0275	128.8019	14.900293	54.302508	88.296611	96.792169
30	114	29.8892	70.1108	134.6312	7.202444	27.128567	60.288317	85.800266
35	126	34.9201	65.0799	138.4976	4.45273	17.05064	34.696165	61.218523
40	138	39.9775	60.0225	140.4016	1.928787	10.142353	21.349958	35.345819
45	150	44.9793	55.0207	140.8544	1.510726	7.878792	15.629307	25.483636
50	162	49.8922	50.1078	140.8783	0.865571	4.952041	10.250504	17.347597
Rhombic dodecahedron H	Radius (micron)	Relative Density (%)	Porosity (%)	Surface Area (mm ²)	Bone growth within pores (%)			
					0.5 MPa	1 MPa	1.5 MPa	2 MPa
5	53	4.9560	95.0440	57.8635	1.834554	0.066427	0.014123	0
10	77	9.9581	90.0419	78.5227	85.66114	40.56371	11.07391	3.666956
15	97	15.1384	84.8616	93.0991	99.11134	93.1414	80.20089	62.28938
20	114	20.1285	79.8715	103.6024	99.97665	99.1691	97.25927	93.74993
25	129	24.8926	75.1074	111.4302	99.37169	99.97853	99.37232	98.6561
30	144	29.9207	70.0793	117.9085	90.7958	99.8804	99.98553	99.66179
35	159	35.1399	64.8601	123.0369	53.92864	97.77207	99.83321	99.96445
40	173	40.1212	59.8788	126.6059	30.43445	84.61858	98.6406	99.7486
45	187	45.1488	54.8512	128.9988	13.87917	49.834	88.85175	98.21215
50	201	50.1645	49.8355	130.2158	6.571477	30.00461	57.32158	87.79503
Rhombic dodecahedron V	Radius (micron)	Relative Density (%)	Porosity (%)	Surface Area (mm ²)	Bone growth within pores (%)			
					0.5 MPa	1 MPa	1.5 MPa	2 MPa
5	53	4.9556	95.0444	55.9268	98.25928	81.91517	52.40337	30.50676
10	77	9.9575	90.0425	75.9319	98.11253	99.90446	99.94294	99.56142
15	97	15.1376	84.8624	90.0693	76.63105	97.49816	99.50686	99.82875
20	114	20.1284	79.8716	100.2753	21.7075	83.71176	96.5555	98.83043
25	129	24.8927	75.1073	107.8986	5.912808	52.46568	87.18981	95.85059
30	144	29.9203	70.0797	114.2268	2.232629	21.44252	63.02574	86.46611
35	159	35.1395	64.8605	119.2593	1.418316	11.37592	33.95997	65.31981
40	173	40.1206	59.8794	122.7878	0.86449	4.851653	19.01344	36.71431
45	187	45.1480	54.8520	125.1874	0.711802	3.131749	12.398	25.23644
50	201	50.1641	49.8359	126.4585	0.670365	1.672918	6.364138	15.56695
Diamond	Radius (micron)	Relative Density (%)	Porosity (%)	Surface Area (mm ²)	Bone growth within pores (%)			
					0.5 MPa	1 MPa	1.5 MPa	2 MPa
5	N/A	5.0000	94.9999	45.6746	95.62319	52.93323	13.69501	4.873945
10	N/A	9.9999	90.0000	63.0208	99.85343	99.12466	97.47535	93.75395
15	N/A	14.999	85.0000	75.6003	99.20578	99.84086	99.50634	99.08438
20	N/A	19.999	80.0000	85.6548	93.40137	99.57873	99.82144	99.71608
25	N/A	24.999	75.0000	94.0503	70.96309	98.13238	99.66226	99.82358
30	N/A	29.999	70.0000	101.2270	43.37342	93.8881	98.98394	99.67654
35	N/A	34.999	65.0000	107.4430	19.56972	79.81082	96.60846	99.18234
40	N/A	40.000	59.9999	112.8591	7.80198	54.97346	88.20415	97.19649
45	N/A	44.999	55.0000	117.5830	5.575709	41.66772	76.79636	93.65277
50	N/A	50.000	49.9999	121.6854	3.76099	22.81313	55.44129	80.99934

Table E2. Biophysical stimulus average under pressures of 0.5 MPa, 1 MPa, 1.5 MPa and 2 MPa

Octet	Biophysical Stimulus Average				Tetrahedron	Biophysical Stimulus Average			
	0.5 MPa	1 MPa	1.5 MPa	2 MPa		0.5 MPa	1 MPa	1.5 MPa	2 MPa
0	0.0900	0.1868	0.2865	0.3881	5	0.0724	0.1496	0.2286	0.3111
10	0.0363	0.0748	0.1143	0.1548	10	0.0312	0.0641	0.0984	0.1335
15	0.0204	0.0408	0.0613	0.0846	15	0.0183	0.0367	0.0552	0.0769
20	0.0134	0.0268	0.0403	0.0538	20	0.0128	0.0255	0.0383	0.0527
25	0.0091	0.0183	0.0274	0.0366	25	0.0096	0.0191	0.0287	0.0383
30	0.0065	0.0130	0.0194	0.0259	30	0.0073	0.0146	0.0220	0.0293
35	0.0051	0.0103	0.0155	0.0206	35	0.0059	0.0119	0.0178	0.0238
35	0.0041	0.0081	0.0122	0.0162	40	0.0048	0.0096	0.0144	0.0192
45	0.0032	0.0063	0.0095	0.0127	45	0.0041	0.0081	0.0122	0.0163
50	0.0028	0.0056	0.0084	0.0112	50	0.0036	0.0072	0.0107	0.0143
Star	Biophysical Stimulus Average				Truncated Cube	Biophysical Stimulus Average			
	0.5 MPa	1 MPa	1.5 MPa	2 MPa		0.5 MPa	1 MPa	1.5 MPa	2 MPa
5	0.0730	0.1507	0.2303	0.3129	5	0.0688	0.1409	0.2154	0.2905
10	0.0313	0.0641	0.0981	0.1327	10	0.0274	0.0549	0.0852	0.1151
15	0.0178	0.0356	0.0535	0.0741	15	0.0151	0.0302	0.0453	0.0604
20	0.0123	0.0246	0.0369	0.0493	20	0.0097	0.0194	0.0291	0.0388
25	0.0092	0.0184	0.0277	0.0369	25	0.0080	0.0161	0.0291	0.0322
30	0.0068	0.0137	0.0205	0.0273	30				
35	0.0050	0.0101	0.0151	0.0201	35				
40	0.0043	0.0085	0.0128	0.0171	40				
45	0.0033	0.0067	0.0100	0.0133	45				
50	0.0028	0.0057	0.0085	0.0114	50				
Vintile	Biophysical Stimulus Average				Truncated Cuboctahedron	Biophysical Stimulus Average			
	0.5 MPa	1 MPa	1.5 MPa	2 MPa		0.5 MPa	1 MPa	1.5 MPa	2 MPa
5	0.1325	0.2741	0.4210	0.5725	5	0.1610	0.3329	0.5118	0.6972
10	0.0368	0.0738	0.1141	0.1540	10	0.0429	0.0878	0.1339	0.1808
15	0.0185	0.0370	0.0555	0.0741	15	0.0219	0.0439	0.0670	0.0903
20	0.0108	0.0217	0.0326	0.0434	20	0.0136	0.0273	0.0413	0.0555
25	0.0075	0.0149	0.0224	0.0299	25	0.0092	0.0184	0.0276	0.0371
30	0.0057	0.0115	0.0172	0.0229	30	0.0066	0.0132	0.0199	0.0266
35	0.0043	0.0086	0.0129	0.0172	35	0.0046	0.0091	0.0137	0.0183
40	0.0033	0.0067	0.0100	0.0134	40				
45	0.0026	0.0051	0.0077	0.0102	45				
50					50				
X Cross Grid	Biophysical Stimulus Average				Rhombicuboctahedron	Biophysical Stimulus Average			
	0.5 MPa	1 MPa	1.5 MPa	2 MPa		0.5 MPa	1 MPa	1.5 MPa	2 MPa
5	0.0864	0.1787	0.2733	0.3684	5	0.0820	0.1688	0.2579	0.3487
10	0.0342	0.0709	0.1086	0.1470	10	0.0323	0.0657	0.0999	0.1349
15	0.0185	0.0370	0.0574	0.0779	15	0.0167	0.0333	0.0506	0.0682
20	0.0121	0.0242	0.0364	0.0500	20	0.0105	0.0209	0.0316	0.0424
25	0.0089	0.0179	0.0268	0.0358	25	0.0070	0.0140	0.0210	0.0281
30	0.0066	0.0131	0.0197	0.0263	30	0.0049	0.0099	0.0148	0.0197
35	0.0051	0.0101	0.0152	0.0202	35	0.0038	0.0075	0.0113	0.0150
40	0.0041	0.0083	0.0124	0.0166	40	0.0029	0.0058	0.0087	0.0116
45	0.0035	0.0069	0.0104	0.0138	45	0.0024	0.0048	0.0072	0.0097
50	0.0029	0.0057	0.0086	0.0114	50	0.0019	0.0038	0.0057	0.0076
Cube	Biophysical Stimulus Average				Rhombic dodecahedron H	Biophysical Stimulus Average			
	0.5 MPa	1 MPa	1.5 MPa	2 MPa		0.5 MPa	1 MPa	1.5 MPa	2 MPa
5	0.0322	0.0665	0.1017	0.1374	5	1.8099	3.9762	6.5486	0.0000
10	0.0164	0.0327	0.0491	0.0677	10	0.3606	0.7573	1.1618	1.5754
15	0.0108	0.0217	0.0325	0.0433	15	0.1286	0.2683	0.4122	0.5617

20	0.0079	0.0159	0.0238	0.0318	20	0.0600	0.1249	0.1921	0.2606
25	0.0064	0.0128	0.0193	0.0257	25	0.0339	0.0706	0.1085	0.1474
30	0.0051	0.0103	0.0154	0.0206	30	0.0215	0.0439	0.0671	0.0910
35	0.0043	0.0087	0.0130	0.0173	35	0.0135	0.0270	0.0413	0.0558
40	0.0036	0.0072	0.0107	0.0143	40	0.0096	0.0193	0.0290	0.0393
45	0.0030	0.0060	0.0090	0.0120	45	0.0067	0.0134	0.0200	0.0267
50	0.0025	0.0051	0.0076	0.0102	50	0.0049	0.0098	0.0147	0.0195
Tesseract					Rhombic dodecahedron V				
Biophysical Stimulus Average					Biophysical Stimulus Average				
	0.5 MPa	1 MPa	1.5 MPa	2 MPa		0.5 MPa	1 MPa	1.5 MPa	2 MPa
5	0.1086	0.2259	0.3510	0.4766	5	0.1903	0.3963	0.6068	0.8207
10	0.0468	0.0976	0.1503	0.2045	10	0.0381	0.0759	0.1152	0.1553
15	0.0271	0.0542	0.0850	0.1158	15	0.0153	0.0299	0.0449	0.0599
20	0.0180	0.0360	0.0541	0.0757	20	0.0084	0.0168	0.0251	0.0335
25	0.0128	0.0256	0.0384	0.0529	25	0.0057	0.0115	0.0172	0.0230
30	0.0099	0.0198	0.0297	0.0397	30	0.0043	0.0085	0.0127	0.0170
35	0.0081	0.0161	0.0242	0.0323	35	0.0034	0.0067	0.0101	0.0135
40	0.0066	0.0132	0.0197	0.0263	40	0.0026	0.0053	0.0079	0.0105
45	0.0052	0.0104	0.0156	0.0207	45	0.0022	0.0044	0.0066	0.0088
50	0.0044	0.0087	0.0131	0.0174	50	0.0018	0.0036	0.0054	0.0072
X-Shape					Diamond				
Biophysical Stimulus Average					Biophysical Stimulus Average				
	0.5 MPa	1 MPa	1.5 MPa	2 MPa		0.5 MPa	1 MPa	1.5 MPa	2 MPa
5	0.2837	0.5997	0.9450	1.3211	5	0.2804	0.5848	0.9060	1.2430
10	0.0908	0.1886	0.2913	0.3969	10	0.0716	0.1496	0.2297	0.3109
15	0.0444	0.0892	0.1386	0.1878	15	0.0345	0.0692	0.1077	0.1459
20	0.0276	0.0554	0.0851	0.1151	20	0.0204	0.0408	0.0613	0.0846
25	0.0169	0.0338	0.0508	0.0694	25	0.0140	0.0280	0.0421	0.0562
30	0.0126	0.0252	0.0378	0.0512	30	0.0107	0.0215	0.0322	0.0430
35	0.0086	0.0173	0.0259	0.0346	35	0.0081	0.0162	0.0244	0.0325
40	0.0069	0.0138	0.0206	0.0275	40	0.0062	0.0125	0.0187	0.0249
45	0.0055	0.0109	0.0164	0.0218	45	0.0055	0.0110	0.0165	0.0219
50	0.0042	0.0083	0.0125	0.0167	50	0.0044	0.0087	0.0131	0.0174

Appendix F - Result Plot of All Topology Evaluated

Figure F1. Percentage of the void space that procures an optimal environment for bone growth (%) in function of the relative density, under a pressure of 0.5 MPa, 1 MPa, 1.5 MPa and 2 MPa, for the Octet topology

Figure F2. Percentage of the void space that procures an optimal environment for bone growth (%) in function of the relative density, under a pressure of 0.5 MPa, 1 MPa, 1.5 MPa and 2 MPa, for the Cubic Center topology

Figure F3. Percentage of the void space that procures an optimal environment for bone growth (%) in function of the relative density, under a pressure of 0.5 MPa, 1 MPa, 1.5 MPa and 2 MPa, for the Diamond topology

Figure F4. Percentage of the void space that procures an optimal environment for bone growth (%) in function of the relative density, under a pressure of 0.5 MPa, 1 MPa, 1.5 MPa and 2 MPa, for the Vintile topology

Figure F5. Percentage of the void space that procures an optimal environment for bone growth (%) in function of the relative density, under a pressure of 0.5 MPa, 1 MPa, 1.5 MPa and 2 MPa, for the Cubic topology

Figure F6. Percentage of the void space that procures an optimal environment for bone growth (%) in function of the relative density, under a pressure of 0.5 MPa, 1 MPa, 1.5 MPa and 2 MPa, for the Body Centered (X-Shape) topology

Figure F7. Percentage of the void space that procures an optimal environment for bone growth (%) in function of the relative density, under a pressure of 0.5 MPa, 1 MPa, 1.5 MPa and 2 MPa, for the Rhombic Dodecahedron (vertical) topology

Figure F8. Percentage of the void space that procures an optimal environment for bone growth (%) in function of the relative density, under a pressure of 0.5 MPa, 1 MPa, 1.5 MPa and 2 MPa, for the Rhombic Dodecahedron (Horizontal) topology

Figure F9. Percentage of the void space that procures an optimal environment for bone growth (%) in function of the relative density, under a pressure of 0.5 MPa, 1 MPa, 1.5 MPa and 2 MPa, for the Rhombicuboctahedron topology

Figure F10. Percentage of the void space that procures an optimal environment for bone growth (%) in function of the relative density, under a pressure of 0.5 MPa, 1 MPa, 1.5 MPa and 2 MPa, for the Tetrahedron topology

Figure F11. Percentage of the void space that procures an optimal environment for bone growth (%) in function of the relative density, under a pressure of 0.5 MPa, 1 MPa, 1.5 MPa and 2 MPa, for the Truncatedcuboctahedron topology

Figure F12. Percentage of the void space that procures an optimal environment for bone growth (%) in function of the relative density, under a pressure of 0.5 MPa, 1 MPa, 1.5 MPa and 2 MPa, for the Truncated Cube topology

Figure F13. Percentage of the void space that procures an optimal environment for bone growth (%) in function of the relative density, under a pressure of 0.5 MPa, 1 MPa, 1.5 MPa and 2 MPa, for the Tesseract topology

Figure F14. Percentage of the void space that procures an optimal environment for bone growth (%) in function of the relative density, under a pressure of 0.5 MPa, 1 MPa, 1.5 MPa and 2 MPa, for the X Cross Grid topology

Figure F15. Average biophysical stimulus for 10% relative density structures, under a pressure of 2 MPa

Figure F16. Average biophysical stimulus for 30% relative density structures, under a pressure of 2 MPa

Figure F17. Average biophysical stimulus for 50% relative density structures, under a pressure of 2 MPa

Figure F18. Elements with optimal stimulus for bone growth, within a structure made of Diamond topology, with 10% and 40% relative density, under a pressure of 0.5 MPa and 2 MPa

Figure F19. Distribution of the velocity and the strain part of the biophysical stimulus, the Octet sample at 40% relative density, under a pressure of 2 MPa

Figure F20. Elements with optimal stimulus for bone growth, within a structure made of Octet topology, with 10% and 40% relative density, under a pressure of 0.5 MPa and 2 MPa

Figure F21. Distribution of the velocity and the strain part of the biophysical stimulus, the Cubic Center (Star) sample at 40% relative density, under a pressure of 2 MPa

Figure F22. Elements with optimal stimulus for bone growth, within a structure made of Cubic Center (Star) topology, with 10% and 40% relative density, under a pressure of 0.5 MPa and 2 MPa

Figure F23. Distribution of the velocity and the strain part of the biophysical stimulus, the Vintile sample at 40% relative density, under a pressure of 2 MPa

Figure F24. Elements with optimal stimulus for bone growth, within a structure made of Vintile topology, with 10% and 40% relative density, under a pressure of 0.5 MPa and 2 MPa

Figure F25. Distribution of the velocity and the strain part of the biophysical stimulus, the X-Shape sample at 40% relative density, under a pressure of 2 MPa

Figure F26. Distribution of the velocity and the strain part of the biophysical stimulus, the Rhombic Dodecahedron (Horizontal) sample at 40% relative density, under a pressure of 2 MPa

Figure F27. Distribution of the velocity and the strain part of the biophysical stimulus, the Rhombic Dodecahedron (vertical) sample at 40% relative density, under a pressure of 2 MPa

Figure F28. Elements with optimal stimulus for bone growth, within a structure made of Rhombic Dodecahedron (vertical) topology, with 10% and 40% relative density, under a pressure of 0.5 MPa and 2 MPa

Elements with optimal stimulus for bone growth

Symmetry plane

Periphery

Velocity/3um/s

Symmetry plane

Periphery

Octahedral Shear Strain/0.0375

Symmetry plane

Periphery

Figure F29. Distribution of the velocity and the strain part of the biophysical stimulus, the Rhombicuboctahedron sample at 40% relative density, under a pressure of 2 MPa

Figure F30. Elements with optimal stimulus for bone growth, within a structure made of Rhombicuboctahedron topology, with 10% and 40% relative density, under a pressure of 0.5 MPa and 2 MPa

Figure F30. Distribution of the velocity and the strain part of the biophysical stimulus, the Tetrahedron sample at 40% relative density, under a pressure of 2 MPa

Figure F31. Elements with optimal stimulus for bone growth, within a structure made of Tetrahedron topology, with 10% and 40% relative density, under a pressure of 0.5 MPa and 2 MPa

Figure F32. Distribution of the velocity and the strain part of the biophysical stimulus, the Truncatedcuboctahedron sample at 40% relative density, under a pressure of 2 MPa

Figure F33. Elements with optimal stimulus for bone growth, within a structure made of Truncatedcuboctahedron topology, with 10% and 35% relative density, under a pressure of 0.5 MPa and 2 MPa

Figure F34. Distribution of the velocity and the strain part of the biophysical stimulus, the Truncated Cube sample at 35% relative density, under a pressure of 2 MPa

Figure F35. Elements with optimal stimulus for bone growth, within a structure made of Truncated Cube topology, with 10% and 40% relative density, under a pressure of 0.5 MPa and 2 MPa

Figure F36. Distribution of the velocity and the strain part of the biophysical stimulus, the X-Cross-Grid sample at 40% relative density, under a pressure of 2 MPa

Figure F37. Elements with optimal stimulus for bone growth, within a structure made of X-Cross-Grid topology, with 10% and 40% relative density, under a pressure of 0.5 MPa and 2 MPa

Figure F38. Distribution of the velocity and the strain part of the biophysical stimulus, the Tesseract sample at 40% relative density, under a pressure of 2 MPa

Figure F39. Elements with optimal stimulus for bone growth, within a structure made of Tesseract topology, with 10% and 40% relative density, under a pressure of 0.5 MPa and 2 MPa

Appendix G - Python Script for SpaceClaim Lattice CAD Creation

```
# Python Script, API Version = V19 Beta
#####
#Clear all objects
while GetRootPart().Components.Count >0 :
    GetRootPart().Components[0].Delete()
while GetRootPart().Curves.Count >0 :
    GetRootPart().Curves[0].Delete()
while GetRootPart().DatumPlanes.Count >0 :
    GetRootPart().DatumPlanes[0].Delete()
while GetRootPart().Bodies.Count >0 :
    GetRootPart().Bodies[0].Delete()
while GetRootPart().Beams.Count >0 :
    GetRootPart().Beams[0].Delete()
#####
#Function to calculate unit vector
def UnitVector(point1,point2):
    unitVec = []
    vecMagnitude =0.0
    for i in range(len(point1)):
        unitVec.append(float(point2[i]-point1[i]))
        vecMagnitude = vecMagnitude + (point2[i]-point1[i])** 2
    vecMagnitude = math.sqrt(vecMagnitude)
    for i in range(len(unitVec)):
        unitVec[i] = unitVec[i]/vecMagnitude
    return unitVec
#Function to sweep
def sweep(selection, trajectories):
    options = SweepCommandOptions()
    options.ExtrudeType = ExtrudeType.ForceAdd
    options.Select = True
    result = Sweep.Execute(selection, trajectories, options, None)
    return None
#####
#PARAMETERS and characteristic specific to the unit cell type
#####
#The default setting would be a 2x2x4 lattice (Put 1 to activate the 4X4X4 Lattice)
Four_by_Four = Parameters.Four_by_Four
##### Cross section profil #####
circle_profil = Parameters.circle_profil
square_profil = Parameters.square_profil
triangle_profil = Parameters.triangle_profil
i_beam_profil = Parameters.i_beam_profil
rectangular_profil = Parameters.rectangular_profil
losange_profil = Parameters.losange_profil

##### Unit cell topology #####
Octet = Parameters.Octet
Star = Parameters.Star
Vintile = Parameters.Vintile
X_cross_grid = Parameters.X_cross_grid
Cube = Parameters.Cube
Truncated_cube = Parameters.Truncated_cube
Rhombic_dodecahedron_H = Parameters.Rhombic_dodecahedron_H
Rhombic_dodecahedron_V = Parameters.Rhombic_dodecahedron_V
```



```

X_Shape = Parameters.X_Shape
Tesseract = Parameters.Tesseract
Tetrahedron = Parameters.Tetrahedron
Truncatedcuboctahedron = Parameters.Truncatedcuboctahedron
Rhombicuboctahedron = Parameters.Rhombicuboctahedron
#For the Diamond lattice this script shouldn't be used. That unit cells can't be mirrored
#Diamond can be modelled manually with ANSYS Material Desing Module.
Diamond_1 = Parameters.Diamond_1
Diamond_2 = Parameters.Diamond_2

#mesure pour truncated cuboctahedron
xTC = float(1/(4.0+(2)**(0.5)));
bTC = float(2/(4.0+(2)**(0.5)));
cTC = float(2/(4.0+(2)**(0.5))+(1-(4/(4.0+(2)**(0.5)))));
dTC = float(3/(4.0+(2)**(0.5))+(1-(4/(4.0+(2)**(0.5)))));
#mesure pour rhombicuboctahedron
xR = float(1/(2.0+(2)**(0.5)));
aR = float(1/(2.0+(2)**(0.5))+float(((2)**(0.5))/(2.0+(2)**(0.5))));

#Unit cell parameters
L = Parameters.Unit_Length
Lx = Parameters.Unit_Length_x
Ly = Parameters.Unit_Length_y
Lz = Parameters.Unit_Length_z
r = Parameters.Radius

#Cross section parameters
if i_beam_profil or rectangular_profil:
    H = Parameters.Beam_Height
    W = Parameters.Beam_Width
    HT = Parameters.Beam_Mid_Thickness
    VT = Parameters.Beam_Side_Thickness
    r = (math.sqrt((H/2)**2+(W/2)**2))
elif losange_profil:
    H = Parameters.Losange_Height
    W = Parameters.Losange_width
    r = (H/2)
#####
#Unit Cell Nodes Location and Connection
#####
if Diamond_1 :
    pts_array = [
        [0*Lx,0*Ly,1*Lz], [0*Lx,1*Ly,0*Lz], [1*Lx,1*Ly,1*Lz], [1*Lx,0*Ly,0*Lz],
        [0.5*Lx,0*Ly,0.5*Lz], [0*Lx,0.5*Ly,0.5*Lz], [0.5*Lx,0.5*Ly,0*Lz], [0.5*Lx,1*Ly,0.5*Lz],
        [1*Lx,0.5*Ly,0.5*Lz], [0.5*Lx,0.5*Ly,1*Lz], [0.75*Lx,0.25*Ly,0.25*Lz], [0.75*Lx,0.75*Ly,0.75*Lz],
        [0.25*Lx,0.75*Ly,0.25*Lz], [0.25*Lx,0.25*Ly,0.75*Lz],
    ]
    conn = [
        [0, 13], [12, 1], [11, 2], [3, 10], [4, 10], [13, 4], [13, 5], [5, 12],
        [12, 7], [11, 7], [11, 8], [10, 8], [13, 9], [10, 6], [9, 11], [12, 6]
    ]
    pts_sphere = [
        [0*Lx+r,0*Ly,1*Lz], [0*Lx+r,1*Ly,0*Lz], [1*Lx+r,1*Ly,1*Lz], [1*Lx+r,0*Ly,0*Lz],
        [0.5*Lx+r,0*Ly,0.5*Lz], [0*Lx+r,0.5*Ly,0.5*Lz], [0.5*Lx+r,0.5*Ly,0*Lz], [0.5*Lx+r,1*Ly,0.5*Lz],
        [1*Lx+r,0.5*Ly,0.5*Lz], [0.5*Lx+r,0.5*Ly,1*Lz], [0.75*Lx+r,0.25*Ly,0.25*Lz], [0.75*Lx+r,0.75*Ly,0.75*Lz],
        [0.25*Lx+r,0.75*Ly,0.25*Lz], [0.25*Lx+r,0.25*Ly,0.75*Lz],
    ]

```

```

Unit_name = "Diamond_1"
if Diamond_2 :
    pts_array = [
        [1*Lx,0*Ly,1*Lz], [0.75*Lx,0.25*Ly,0.75*Lz], [0.5*Lx,0*Ly,0.5*Lz], [1*Lx,0.5*Ly,0.5*Lz],
        [0.5*Lx,0.5*Ly,1*Lz], [0.25*Lx,0.25*Ly,0.25*Lz], [0.75*Lx,0.75*Ly,0.25*Lz], [0.25*Lx,0.75*Ly,0.75*Lz],
        [0*Lx,0*Ly,0*Lz], [1*Lx,1*Ly,0*Lz], [0*Lx,1*Ly,1*Lz], [0*Lx,0.5*Ly,0.5*Lz],
        [0.5*Lx,0.5*Ly,0*Lz], [0.5*Lx,1*Ly,0.5*Lz],
    ]
    conn = [
        [0, 1], [2, 1], [1, 4], [3, 1], [2, 5], [3, 6], [4, 7], [5, 8],
        [6, 9], [10, 7], [11, 5], [11, 7], [12, 5], [12, 6], [13, 6], [13, 7]
    ]
    pts_sphere = [
        [1*Lx+r,0*Ly,1*Lz], [0.75*Lx+r,0.25*Ly,0.75*Lz], [0.5*Lx+r,0*Ly,0.5*Lz], [1*Lx+r,0.5*Ly,0.5*Lz],
        [0.5*Lx+r,0.5*Ly,1*Lz], [0.25*Lx+r,0.25*Ly,0.25*Lz], [0.75*Lx+r,0.75*Ly,0.25*Lz], [0.25*Lx+r,0.75*Ly,0.75*Lz],
        [0*Lx+r,0*Ly,0*Lz], [1*Lx+r,1*Ly,0*Lz], [0*Lx+r,1*Ly,1*Lz], [0*Lx+r,0.5*Ly,0.5*Lz],
        [0.5*Lx+r,0.5*Ly,0*Lz], [0.5*Lx+r,1*Ly,0.5*Lz],
    ]
    Unit_name = "Diamond_2"
elif Star :
    pts_array = [
        [0*L,0*L,0*L], [1*L,0*L,0*L], [0*L,1*L,0*L], [0*L,0*L,1*L],
        [1*L,1*L,1*L], [0*L,1*L,1*L], [1*L,0*L,1*L], [1*L,1*L,0*L],
    ]
    conn = [
        [0, 1], [0, 2], [0, 3], [0, 4], [1, 5], [2, 6],
        [7, 3], [3, 6], [3, 5], [2, 7], [2, 5], [5, 4],
        [1, 7], [1, 6], [6, 4], [7, 4],
    ]
    pts_sphere = [
        [0*L+r,0*L,0*L], [1*L+r,0*L,0*L], [0*L+r,1*L,0*L], [0*L+r,0*L,1*L],
        [1*L+r,1*L,1*L], [0*L+r,1*L,1*L], [1*L+r,0*L,1*L], [1*L+r,1*L,0*L],
    ]
    Unit_name = "Star_cubic_center"
elif Rhombic_dodecahedron_H :
    pts_array = [
        [0*Lx,1*Ly,0.5*Lz], [0*Lx,0.5*Ly,0.25*Lz], [0*Lx,0*Ly,0.5*Lz], [0*Lx,0.5*Ly,0.75*Lz],
        [0.5*Lx,1*Ly,0.75*Lz], [0.5*Lx,1*Ly,0.25*Lz], [1*Lx,1*Ly,0.5*Lz],
        [1*Lx,0.5*Ly,0.25*Lz], [1*Lx,0*Ly,0.5*Lz], [1*Lx,0.5*Ly,0.75*Lz],
        [0.5*Lx,0*Ly,0.75*Lz], [0.5*Lx,0*Ly,0.25*Lz], [0.5*Lx,0.5*Ly,0*Lz], [0.5*Lx,0.5*Ly,1*Lz],
    ]
    conn = [
        [0, 1], [1, 2], [2, 3], [0, 3], [0, 5], [5, 6],
        [6, 4], [4, 0], [6, 7], [8, 7], [8, 9], [9, 6],
        [8, 11], [11, 2], [2, 10], [10, 8],
        [12, 1], [12, 5], [12, 7], [12, 11],
        [13, 3], [13, 4], [13, 9], [13, 10],
    ]
    pts_sphere = [
        [0*Lx+r,1*Ly,0.5*Lz], [0*Lx+r,0.5*Ly,0.25*Lz], [0*Lx+r,0*Ly,0.5*Lz], [0*Lx+r,0.5*Ly,0.75*Lz],
        [0.5*Lx+r,1*Ly,0.75*Lz], [0.5*Lx+r,1*Ly,0.25*Lz], [1*Lx+r,1*Ly,0.5*Lz],
        [1*Lx+r,0.5*Ly,0.25*Lz], [1*Lx+r,0*Ly,0.5*Lz], [1*Lx+r,0.5*Ly,0.75*Lz],
        [0.5*Lx+r,0*Ly,0.75*Lz], [0.5*Lx+r,0*Ly,0.25*Lz], [0.5*Lx+r,0.5*Ly,0*Lz], [0.5*Lx+r,0.5*Ly,1*Lz],
    ]
    Unit_name = "Rhombic_H"
elif Rhombic_dodecahedron_V :
    pts_array = [

```

```

[0*Lx,0.5*Ly,1*Lz], [0*Lx,0.25*Ly,0.5*Lz], [0*Lx,0.5*Ly,0*Lz], [0*Lx,0.75*Ly,0.5*Lz],
[0.5*Lx,0.75*Ly,1*Lz], [0.5*Lx,0.25*Ly,1*Lz], [1*Lx,0.5*Ly,1*Lz],
[1*Lx,0.25*Ly,0.5*Lz], [1*Lx,0.5*Ly,0*Lz], [1*Lx,0.75*Ly,0.5*Lz],
[0.5*Lx,0.75*Ly,0*Lz], [0.5*Lx,0.25*Ly,0*Lz], [0.5*Lx,0*Lx,0.5*Lz], [0.5*Lx,1*Ly,0.5*Lz],
]
conn = [
[0, 1], [1, 2], [2, 3], [0, 3], [0, 5], [5, 6],
[6, 4], [4, 0], [6, 7], [8, 7], [8, 9], [9, 6],
[8, 11], [11, 2], [2, 10], [10, 8],
[12, 1], [12, 5], [12, 7], [12, 11],
[13, 3], [13, 4], [13, 9], [13, 10],
]
pts_sphere = [
[0*Lx+r,0.5*Ly,1*Lz], [0*Lx+r,0.25*Ly,0.5*Lz], [0*Lx+r,0.5*Ly,0*Lz], [0*Lx+r,0.75*Ly,0.5*Lz],
[0.5*Lx+r,0.75*Ly,1*Lz], [0.5*Lx+r,0.25*Ly,1*Lz], [1*Lx+r,0.5*Ly,1*Lz],
[1*Lx+r,0.25*Ly,0.5*Lz], [1*Lx+r,0.5*Ly,0*Lz], [1*Lx+r,0.75*Ly,0.5*Lz],
[0.5*Lx+r,0.75*Ly,0*Lz], [0.5*Lx+r,0.25*Ly,0*Lz], [0.5*Lx+r,0*Lx,0.5*Lz], [0.5*Lx+r,1*Ly,0.5*Lz],
]
Unit_name = "Rhombic_V"
elif Truncated_cube :
pts_array = [
[1*Lx/3,0*Ly,0*Lz], [0*Lx,1*Ly/3,0*Lz], [0*Lx,0*Ly,1*Lz/3],
[0*Lx,2*Ly/3,0*Lz], [1*Lx/3,1*Ly,0*Lz], [0*Lx,1*Ly,1*Lz/3],
[0*Lx,0*Ly,2*Lz/3], [1*Lx/3,0*Ly,1*Lz], [0*Lx,1*Ly/3,1*Lz],
[2*Lx/3,0*Ly,0*Lz], [1*Lx,1*Ly/3,0*Lz], [1*Lx,0*Ly,1*Lz/3],
[1*Lx,1*Ly,1*Lz/3], [2*Lx/3,1*Ly,0*Lz], [1*Lx,2*Ly/3,0*Lz],
[1*Lx/3,1*Ly,1*Lz], [0*Lx,1*Ly,2*Lz/3], [0*Lx,2*Ly/3,1*Lz],
[2*Lx/3,1*Ly,1*Lz], [1*Lx,2*Ly/3,1*Lz], [1*Lx,1*Ly,2*Lz/3],
[1*Lx,1*Ly/3,1*Lz], [2*Lx/3,0*Ly,1*Lz], [1*Lx,0*Ly,2*Lz/3],
]
conn = [
[0, 1], [2, 1], [0, 2],
[8, 7], [7, 6], [8, 6],
[15,16], [16, 17], [15, 17],
[5, 4], [4, 3], [3, 5],
[18, 20], [20, 19], [19, 18],
[12,14], [12, 13], [13, 14],
[10,11], [11, 9], [9, 10],
[21, 23], [23, 22], [22, 21],
[15, 18], [8, 17], [7, 22], [19, 21],
[4, 13], [14, 10], [0, 9], [1, 3],
[20, 12], [11, 23], [16, 5], [6, 2],
]
pts_sphere = [
[1*Lx/3+r,0*Ly,0*Lz], [0*Lx+r,1*Ly/3,0*Lz], [0*Lx+r,0*Ly,1*Lz/3],
[0*Lx+r,2*Ly/3,0*Lz], [1*Lx/3+r,1*Ly,0*Lz], [0*Lx+r,1*Ly,1*Lz/3],
[0*Lx+r,0*Ly,2*Lz/3], [1*Lx/3+r,0*Ly,1*Lz], [0*Lx+r,1*Ly/3,1*Lz],
[2*Lx/3+r,0*Ly,0*Lz], [1*Lx+r,1*Ly/3,0*Lz], [1*Lx+r,0*Ly,1*Lz/3],
[1*Lx+r,1*Ly,1*Lz/3], [2*Lx/3+r,1*Ly,0*Lz], [1*Lx+r,2*Ly/3,0*Lz],
[1*Lx/3+r,1*Ly,1*Lz], [0*Lx+r,1*Ly,2*Lz/3], [0*Lx+r,2*Ly/3,1*Lz],
[2*Lx/3+r,1*Ly,1*Lz], [1*Lx+r,2*Ly/3,1*Lz], [1*Lx+r,1*Ly,2*Lz/3],
[1*Lx+r,1*Ly/3,1*Lz], [2*Lx/3+r,0*Ly,1*Lz], [1*Lx+r,0*Ly,2*Lz/3],
]
Unit_name = "Truncated_Cube"
elif Octet :
pts_array = [
[0.5*Lx,0.5*Ly,0*Lz], [0*Lx,0*Ly,0*Lz], [1*Lx,0*Ly,0*Lz], [1*Lx,1*Ly,0*Lz],

```

```

[0*Lx,1*Ly,0*Lz], [1*Lx,0.5*Ly,0.5*Lz], [0.5*Lx,1*Ly,0.5*Lz], [0*Lx,0.5*Ly,0.5*Lz],
[0.5*Lx,0*Ly,0.5*Lz], [0.5*Lx,0.5*Ly,1*Lz], [0*Lx,0*Ly,1*Lz], [0*Lx,1*Ly,1*Lz],
[1*Lx,0*Ly,1*Lz], [1*Lx,1*Ly,1*Lz],
]
conn = [
[0, 1], [0, 2], [0, 3], [0, 4], [0, 5], [0, 6], [0, 7], [0, 8],
[9, 5], [9, 6], [9, 7], [9, 8], [7, 1], [7, 4], [7, 10], [7, 11],
[6, 5], [6, 7], [8, 1], [8, 2], [8, 10], [8, 12], [8, 5], [8, 7],
[9, 10], [9, 12], [9, 13], [9, 11], [6, 4], [6, 3], [6, 11], [6, 13],
[5, 2], [5, 3], [5, 12], [5, 13],
]
pts_sphere = [
[0.5*Lx+r,0.5*Ly,0*Lz], [0*Lx+r,0*Ly,0*Lz], [1*Lx+r,0*Ly,0*Lz], [1*Lx+r,1*Ly,0*Lz],
[0*Lx+r,1*Ly,0*Lz], [1*Lx+r,0.5*Ly,0.5*Lz], [0.5*Lx+r,1*Ly,0.5*Lz], [0*Lx+r,0.5*Ly,0.5*Lz],
[0.5*Lx+r,0*Ly,0.5*Lz], [0.5*Lx+r,0.5*Ly,1*Lz], [0*Lx+r,0*Ly,1*Lz], [0*Lx+r,1*Ly,1*Lz],
[1*Lx+r,0*Ly,1*Lz], [1*Lx+r,1*Ly,1*Lz],
]
Unit_name = "Octet"
elif Vintile:
pts_array = [
[0*Lx,0.5*Ly,0.25*Lz], [0*Lx,0.25*Ly,0*Lz], [0*Lx,0.75*Ly,0*Lz], [0.25*Lx,0.5*Ly,0.5*Lz],
[0.5*Lx,1*Ly,0.25*Lz], [0.5*Lx,0.75*Ly,0.5*Lz], [1*Lx,0.5*Ly,0.25*Lz], [0.75*Lx,0.5*Ly,0.5*Lz],
[0.5*Lx,0*Ly,0.25*Lz], [0.75*Lx,0*Ly,0*Lz], [0.25*Lx,0*Ly,0*Lz], [0.5*Lx,0.25*Ly,0.5*Lz],
[0*Lx,0.5*Ly,0.75*Lz], [0*Lx,0.25*Ly,1*Lz], [0*Lx,0.75*Ly,1*Lz], [0.5*Lx,1*Ly,0.75*Lz],
[1*Lx,0.5*Ly,0.75*Lz], [0.5*Lx,0*Ly,0.75*Lz], [0.75*Lx,0*Ly,1*Lz], [0.25*Lx,0*Ly,1*Lz],
[0.25*Lx,1*Ly,0*Lz], [0.75*Lx,1*Ly,0*Lz], [1*Lx,0.75*Ly,0*Lz], [1*Lx,0.25*Ly,0*Lz],
[0.25*Lx,1*Ly,1*Lz], [0.75*Lx,1*Ly,1*Lz], [1*Lx,0.75*Ly,1*Lz], [1*Lx,0.25*Ly,1*Lz],
]
conn = [
[0, 1], [0, 2], [0, 3], [4, 5], [6, 7], [8, 9], [8, 10], [8, 11],
[12, 13], [12, 14], [12, 3], [15, 5], [16, 7], [17, 18], [17, 19], [17, 11],
[2, 20], [21, 22], [23, 9], [3, 11], [7, 11], [3, 5], [7, 5], [1, 10],
[14, 24], [25, 26], [27, 18], [13, 19], [4, 21], [4, 20], [15, 25], [15, 24],
[6, 23], [6, 22], [16, 27], [16, 26],
]
pts_sphere = [
[0*Lx+r,0.5*Ly,0.25*Lz], [0*Lx+r,0.25*Ly,0*Lz], [0*Lx+r,0.75*Ly,0*Lz], [0.25*Lx+r,0.5*Ly,0.5*Lz],
[0.5*Lx+r,1*Ly,0.25*Lz], [0.5*Lx+r,0.75*Ly,0.5*Lz], [1*Lx+r,0.5*Ly,0.25*Lz], [0.75*Lx+r,0.5*Ly,0.5*Lz],
[0.5*Lx+r,0*Ly,0.25*Lz], [0.75*Lx+r,0*Ly,0*Lz], [0.25*Lx+r,0*Ly,0*Lz], [0.5*Lx+r,0.25*Ly,0.5*Lz],
[0*Lx+r,0.5*Ly,0.75*Lz], [0*Lx+r,0.25*Ly,1*Lz], [0*Lx+r,0.75*Ly,1*Lz], [0.5*Lx+r,1*Ly,0.75*Lz],
[1*Lx+r,0.5*Ly,0.75*Lz], [0.5*Lx+r,0*Ly,0.75*Lz], [0.75*Lx+r,0*Ly,1*Lz], [0.25*Lx+r,0*Ly,1*Lz],
[0.25*Lx+r,1*Ly,0*Lz], [0.75*Lx+r,1*Ly,0*Lz], [1*Lx+r,0.75*Ly,0*Lz], [1*Lx+r,0.25*Ly,0*Lz],
[0.25*Lx+r,1*Ly,1*Lz], [0.75*Lx+r,1*Ly,1*Lz], [1*Lx+r,0.75*Ly,1*Lz], [1*Lx+r,0.25*Ly,1*Lz],
]
Unit_name = "Vintile"
elif Cube:
pts_array = [
[0*Lx,0*Ly,0*Lz], [1*Lx,0*Ly,0*Lz], [0*Lx,1*Ly,0*Lz], [0*Lx,0*Ly,1*Lz],
[1*Lx,0*Ly,1*Lz], [0*Lx,1*Ly,1*Lz], [1*Lx,1*Ly,0*Lz], [1*Lx,1*Ly,1*Lz],
]
conn = [
[0, 1], [0, 2], [0, 3], [3, 4], [3, 5], [2, 6], [2, 5], [5, 7],
[1, 6], [1, 4], [4, 7], [6, 7],
]
pts_sphere = [
[0*Lx+r,0*Ly,0*Lz], [1*Lx+r,0*Ly,0*Lz], [0*Lx+r,1*Ly,0*Lz], [0*Lx+r,0*Ly,1*Lz],
[1*Lx+r,0*Ly,1*Lz], [0*Lx+r,1*Ly,1*Lz], [1*Lx+r,1*Ly,0*Lz], [1*Lx+r,1*Ly,1*Lz],
]

```

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    ]
    Unit_name = "Cube"
elif X_cross_grid:
    pts_array = [
        [0*Lx,0*Ly,0*Lz], [1*Lx,0*Ly,0*Lz], [0*Lx,1*Ly,0*Lz], [0*Lx,0*Ly,1*Lz],
        [1*Lx,0*Ly,1*Lz], [0*Lx,1*Ly,1*Lz], [1*Lx,1*Ly,0*Lz], [1*Lx,1*Ly,1*Lz],
    ]
    conn = [
        [0, 1], [0, 2], [0, 3], [3, 4], [3, 5], [2, 6], [2, 5], [5, 7],
        [1, 6], [1, 4], [4, 7], [6, 7], [0, 4], [0, 5], [0, 6], [0, 7],
        [1, 2], [1, 3], [1, 5], [1, 7], [2, 3], [2, 4], [3, 6], [3, 7],
        [4, 5], [4, 6], [5, 6], [2, 7]
    ]
    pts_sphere = [
        [0*Lx+r,0*Ly,0*Lz], [1*Lx+r,0*Ly,0*Lz], [0*Lx+r,1*Ly,0*Lz], [0*Lx+r,0*Ly,1*Lz],
        [1*Lx+r,0*Ly,1*Lz], [0*Lx+r,1*Ly,1*Lz], [1*Lx+r,1*Ly,0*Lz], [1*Lx+r,1*Ly,1*Lz],
    ]
    Unit_name = "X_X_grid"
elif Truncatedcuboctahedron:
    pts_array = [
        [bTC*Lx,0*Ly,xTC*Lz],
        [xTC*Lx,bTC*Ly,1*Lz], [bTC*Lx,xTC*Ly,1*Lz], [cTC*Lx,xTC*Ly,1*Lz], [dTC*Lx,bTC*Ly,1*Lz],
        [dTC*Lx,cTC*Ly,1*Lz], [cTC*Lx,dTC*Ly,1*Lz], [bTC*Lx,dTC*Ly,1*Lz], [xTC*Lx,cTC*Ly,1*Lz],
        [0*Lx,cTC*Ly,dTC*Lz], [0*Lx,bTC*Ly,dTC*Lz], [0*Lx,xTC*Ly,cTC*Lz], [0*Lx,xTC*Ly,bTC*Lz],
        [0*Lx,bTC*Ly,xTC*Lz], [0*Lx,cTC*Ly,xTC*Lz], [0*Lx,dTC*Ly,bTC*Lz], [0*Lx,dTC*Ly,cTC*Lz],
        [xTC*Lx,0*Ly,bTC*Lz], [xTC*Lx,0*Ly,cTC*Lz], [bTC*Lx,0*Ly,dTC*Lz],
        [cTC*Lx,0*Ly,dTC*Lz], [dTC*Lx,0*Ly,cTC*Lz], [dTC*Lx,0*Ly,bTC*Lz], [cTC*Lx,0*Ly,xTC*Lz],
        [bTC*Lx,1*Ly,xTC*Lz], [xTC*Lx,1*Ly,bTC*Lz], [xTC*Lx,1*Ly,cTC*Lz], [bTC*Lx,1*Ly,dTC*Lz],
        [cTC*Lx,1*Ly,dTC*Lz], [dTC*Lx,1*Ly,cTC*Lz], [dTC*Lx,1*Ly,bTC*Lz], [cTC*Lx,1*Ly,xTC*Lz],
        [xTC*Lx,bTC*Ly,0*Lz], [bTC*Lx,xTC*Ly,0*Lz], [cTC*Lx,xTC*Ly,0*Lz], [dTC*Lx,bTC*Ly,0*Lz],
        [dTC*Lx,cTC*Ly,0*Lz], [cTC*Lx,dTC*Ly,0*Lz], [bTC*Lx,dTC*Ly,0*Lz], [xTC*Lx,cTC*Ly,0*Lz],
        [1*Lx,cTC*Ly,dTC*Lz], [1*Lx,bTC*Ly,dTC*Lz], [1*Lx,xTC*Ly,cTC*Lz], [1*Lx,xTC*Ly,bTC*Lz],
        [1*Lx,bTC*Ly,xTC*Lz], [1*Lx,cTC*Ly,xTC*Lz], [1*Lx,dTC*Ly,bTC*Lz], [1*Lx,dTC*Ly,cTC*Lz],
    ]
    conn = [
        [0, 23], [23, 22], [21, 20], [20, 19], [19, 18], [18, 17], [17, 0], [22, 21],
        [1, 2], [2, 3], [3, 4], [4, 5], [5, 6], [6, 7], [7, 8], [8, 1],
        [10, 11], [11, 12], [12, 13], [13, 14], [14, 15], [15, 16], [16, 9], [9, 10],
        [24, 25], [25, 26], [26, 27], [27, 28], [28, 29], [29, 30], [30, 31], [31, 24],
        [32, 33], [33, 34], [34, 35], [35, 36], [37, 38], [36, 37], [38, 39], [39, 32],
        [40, 41], [41, 42], [42, 43], [43, 44], [44, 45], [45, 46], [46, 47], [47, 40],
        [2, 19], [3, 20], [21, 42], [22, 43], [23, 34], [0, 33], [17, 12], [18, 11],
        [1, 10], [8, 9], [7, 27], [6, 28], [5, 40], [4, 41], [16, 26], [15, 25],
        [14, 39], [13, 32], [47, 29], [30, 46], [45, 36], [44, 35], [31, 37], [24, 38],
    ]
    pts_sphere = [
        [bTC*Lx+r,0*Ly,xTC*Lz],
        [xTC*Lx+r,bTC*Ly,1*Lz], [bTC*Lx+r,xTC*Ly,1*Lz], [cTC*Lx+r,xTC*Ly,1*Lz], [dTC*Lx+r,bTC*Ly,1*Lz],
        [dTC*Lx+r,cTC*Ly,1*Lz], [cTC*Lx+r,dTC*Ly,1*Lz], [bTC*Lx+r,dTC*Ly,1*Lz], [xTC*Lx+r,cTC*Ly,1*Lz],
        [0*Lx+r,cTC*Ly,dTC*Lz], [0*Lx+r,bTC*Ly,dTC*Lz], [0*Lx+r,xTC*Ly,cTC*Lz], [0*Lx+r,xTC*Ly,bTC*Lz],
        [0*Lx+r,bTC*Ly,xTC*Lz], [0*Lx+r,cTC*Ly,xTC*Lz], [0*Lx+r,dTC*Ly,bTC*Lz], [0*Lx+r,dTC*Ly,cTC*Lz],
        [xTC*Lx+r,0*Ly,bTC*Lz], [xTC*Lx+r,0*Ly,cTC*Lz], [bTC*Lx+r,0*Ly,dTC*Lz],
        [cTC*Lx+r,0*Ly,dTC*Lz], [dTC*Lx+r,0*Ly,cTC*Lz], [dTC*Lx+r,0*Ly,bTC*Lz], [cTC*Lx+r,0*Ly,xTC*Lz],
        [bTC*Lx+r,1*Ly,xTC*Lz], [xTC*Lx+r,1*Ly,bTC*Lz], [xTC*Lx+r,1*Ly,cTC*Lz], [bTC*Lx+r,1*Ly,dTC*Lz],
        [cTC*Lx+r,1*Ly,dTC*Lz], [dTC*Lx+r,1*Ly,cTC*Lz], [dTC*Lx+r,1*Ly,bTC*Lz], [cTC*Lx+r,1*Ly,xTC*Lz],
        [xTC*Lx+r,bTC*Ly,0*Lz], [bTC*Lx+r,xTC*Ly,0*Lz], [cTC*Lx+r,xTC*Ly,0*Lz], [dTC*Lx+r,bTC*Ly,0*Lz],
        [dTC*Lx+r,cTC*Ly,0*Lz], [cTC*Lx+r,dTC*Ly,0*Lz], [bTC*Lx+r,dTC*Ly,0*Lz], [xTC*Lx+r,cTC*Ly,0*Lz],
    ]

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[1*Lx+r,cTC*Ly,dTC*Lz], [1*Lx+r,bTC*Ly,dTC*Lz], [1*Lx+r,xTC*Ly,cTC*Lz], [1*Lx+r,xTC*Ly,bTC*Lz],
[1*Lx+r,bTC*Ly,xTC*Lz], [1*Lx+r,cTC*Ly,xTC*Lz], [1*Lx+r,dTC*Ly,bTC*Lz], [1*Lx+r,dTC*Ly,cTC*Lz],
]
Unit_name = "Truncatedcuboctahedron"
elif Rhombicuboctahedron:
pts_array = [
[xR*Lx,0*Ly,xR*Lz], [aR*Lx,0*Ly,xR*Lz], [1*Lx,xR*Ly,xR*Lz], [1*Lx,aR*Ly,xR*Lz],
[aR*Lx,1*Ly,xR*Lz], [xR*Lx,1*Ly,xR*Lz], [0*Lx,aR*Ly,xR*Lz], [0*Lx,xR*Ly,xR*Lz],
[xR*Lx,0*Ly,aR*Lz], [aR*Lx,0*Ly,aR*Lz], [1*Lx,xR*Ly,aR*Lz], [1*Lx,aR*Ly,aR*Lz],
[aR*Lx,1*Ly,aR*Lz], [xR*Lx,1*Ly,aR*Lz], [0*Lx,aR*Ly,aR*Lz], [0*Lx,xR*Ly,aR*Lz],
[xR*Lx,aR*Ly,0*Lz], [xR*Lx,xR*Ly,0*Lz], [xR*Lx,xR*Ly,1*Lz], [xR*Lx,aR*Ly,1*Lz],
[aR*Lx,aR*Ly,0*Lz], [aR*Lx,xR*Ly,0*Lz], [aR*Lx,xR*Ly,1*Lz], [aR*Lx,aR*Ly,1*Lz],
]
conn = [
[0, 7], [0, 1], [1, 2], [2, 3], [3, 4], [4, 5], [6, 5], [6, 7],
[8, 15], [8, 9], [9, 10], [10, 11], [11, 12], [12, 13], [13, 14], [14, 15],
[0, 8], [1, 9], [2, 10], [11, 3], [12, 4], [13, 5], [6, 14],
[7, 15], [5, 16], [16, 17], [0, 17], [8, 18], [18, 19], [19, 13], [4, 20],
[20, 21], [1, 21], [9, 22], [22, 23], [23, 12], [16, 20], [17, 21], [18, 22],
[19, 23], [2, 21], [3, 20], [6, 16], [7, 17], [14, 19], [15, 18], [11, 23], [10, 22],
]
pts_sphere = [
[xR*Lx+r,0*Ly,xR*Lz], [aR*Lx+r,0*Ly,xR*Lz], [1*Lx+r,xR*Ly,xR*Lz], [1*Lx+r,aR*Ly,xR*Lz],
[aR*Lx+r,1*Ly,xR*Lz], [xR*Lx+r,1*Ly,xR*Lz], [0*Lx+r,aR*Ly,xR*Lz], [0*Lx+r,xR*Ly,xR*Lz],
[xR*Lx+r,0*Ly,aR*Lz], [aR*Lx+r,0*Ly,aR*Lz], [1*Lx+r,xR*Ly,aR*Lz], [1*Lx+r,aR*Ly,aR*Lz],
[aR*Lx+r,1*Ly,aR*Lz], [xR*Lx+r,1*Ly,aR*Lz], [0*Lx+r,aR*Ly,aR*Lz], [0*Lx+r,xR*Ly,aR*Lz],
[xR*Lx+r,aR*Ly,0*Lz], [xR*Lx+r,xR*Ly,0*Lz], [xR*Lx+r,xR*Ly,1*Lz], [xR*Lx+r,aR*Ly,1*Lz],
[aR*Lx+r,aR*Ly,0*Lz], [aR*Lx+r,xR*Ly,0*Lz], [aR*Lx+r,xR*Ly,1*Lz], [aR*Lx+r,aR*Ly,1*Lz],
]
Unit_name = "Rhombicuboctahedron"
elif X_Shape:
pts_array = [
[0*Lx,0*Ly,0*Lz], [1*Lx,0*Ly,0*Lz], [0*Lx,1*Ly,0*Lz], [0*Lx,0*Ly,1*Lz],
[1*Lx,0*Ly,1*Lz], [0*Lx,1*Ly,1*Lz], [1*Lx,1*Ly,0*Lz], [1*Lx,1*Ly,1*Lz],
]
conn = [
[6, 3], [0, 7], [1, 5], [2, 4],
]
pts_sphere = [
[0*Lx+r,0*Ly,0*Lz], [1*Lx+r,0*Ly,0*Lz], [0*Lx+r,1*Ly,0*Lz], [0*Lx+r,0*Ly,1*Lz],
[1*Lx+r,0*Ly,1*Lz], [0*Lx+r,1*Ly,1*Lz], [1*Lx+r,1*Ly,0*Lz], [1*Lx+r,1*Ly,1*Lz],
]
Unit_name = "X_Shape"
elif Tesseract :
pts_array = [
[0*Lx,0*Ly,0*Lz], [1*Lx,0*Ly,0*Lz], [0.25*Lx,0.25*Ly,0.25*Lz], [0.75*Lx,0.25*Ly,0.25*Lz],
[0*Lx,1*Ly,0*Lz], [0.25*Lx,0.75*Ly,0.25*Lz], [0*Lx,0*Ly,1*Lz], [0.25*Lx,0.25*Ly,0.75*Lz],
[0.75*Lx,0.75*Ly,0.25*Lz], [0.75*Lx,0.75*Ly,0.75*Lz], [0.75*Lx,0.25*Ly,0.75*Lz], [0.25*Lx,0.75*Ly,0.75*Lz],
[1*Lx,1*Ly,0*Lz], [1*Lx,0*Ly,1*Lz], [1*Lx,1*Ly,1*Lz], [0*Lx,1*Ly,1*Lz],
]
conn = [
[0, 1], [2, 3], [0, 4], [2, 5], [0, 6], [2, 7], [8, 3], [8, 5],
[8, 9], [10, 3], [10, 7], [10, 9], [11, 5], [11, 7], [11, 9], [0, 2],
[1, 3], [12, 8], [4, 5], [6, 7], [13, 10], [14, 9], [15, 11], [6, 13],
[6, 15], [4, 12], [4, 15], [15, 14], [1, 12], [1, 13], [13, 14], [12, 14],
]
pts_sphere = [

```

```

[0*Lx+r,0*Ly,0*Lz], [1*Lx+r,0*Ly,0*Lz], [0.25*Lx+r,0.25*Ly,0.25*Lz], [0.75*Lx+r,0.25*Ly,0.25*Lz],
[0*Lx+r,1*Ly,0*Lz], [0.25*Lx+r,0.75*Ly,0.25*Lz], [0*Lx+r,0*Ly,1*Lz], [0.25*Lx+r,0.25*Ly,0.75*Lz],
[0.75*Lx+r,0.75*Ly,0.25*Lz], [0.75*Lx+r,0.75*Ly,0.75*Lz], [0.75*Lx+r,0.25*Ly,0.75*Lz], [0.25*Lx+r,0.75*Ly,0.75*Lz],
[1*Lx+r,1*Ly,0*Lz], [1*Lx+r,0*Ly,1*Lz], [1*Lx+r,1*Ly,1*Lz], [0*Lx+r,1*Ly,1*Lz],
]
Unit_name = "Tesseract"
elif Tetrahedron:
pts_array = [
[0*Lx,0*Ly,0*Lz], [1*Lx,0*Ly,0*Lz], [0*Lx,1*Ly,0*Lz], [0*Lx,0*Ly,1*Lz],
[1*Lx,0*Ly,1*Lz], [0*Lx,1*Ly,1*Lz], [1*Lx,1*Ly,0*Lz], [1*Lx,1*Ly,1*Lz],
]
conn = [
[0, 1], [0, 2], [0, 3], [3, 4], [3, 5], [2, 6], [2, 5], [5, 7],
[1, 6], [1, 4], [4, 7], [6, 7], [2, 3], [6, 4], [2, 1], [5, 4],
[3, 1], [5, 6], [5, 1], [7, 0]
]
pts_sphere = [
[0*Lx+r,0*Ly,0*Lz], [1*Lx+r,0*Ly,0*Lz], [0*Lx+r,1*Ly,0*Lz], [0*Lx+r,0*Ly,1*Lz],
[1*Lx+r,0*Ly,1*Lz], [0*Lx+r,1*Ly,1*Lz], [1*Lx+r,1*Ly,0*Lz], [1*Lx+r,1*Ly,1*Lz],
]
Unit_name = "Tetrahedron"
#####
#Point and line Creation (Making the unit cell skeleton)
#####
PTS = []
for i in pts_array:
PTS.append(Point.Create(*i))
PTS2 = []
for i in pts_sphere:
PTS2.append(Point.Create(*i))
for i in range(len(PTS)):
SphereBody.Create(PTS[i], PTS2[i], ExtrudeType.None)
#####
#Cross Section Creation and extrusion
#####
ComponentsIndex = 0
for i in conn:
selection = Selection.Create(GetRootPart())
result = ComponentHelper.CreateNewComponent(selection, None)
SketchLine.Create(PTS[i[1]], PTS[i[0]])
unitVec = UnitVector(pts_array[i[0]], pts_array[i[1]])
direc = Direction.Create(*unitVec)
PointForDirection = (Point.Create(*pts_array[i[0]]))
frame = Frame.Create(PointForDirection, direc)
plane = Plane.Create(frame)
ViewHelper.SetSketchPlane(plane)
#Circle
if circle_profil:
origin = Point2D.Create(0,0)
SketchCircle.Create(origin,r)
#Square
elif square_profil:
point1 = Point2D.Create(M(r*0.7071),M(-r*0.7071))
point2 = Point2D.Create(M(-r*0.7071),M(-r*0.7071))
point3 = Point2D.Create(M(-r*0.7071),M(r*0.7071))
result = SketchRectangle.Create(point1, point2, point3)
#Triangle

```

```

elif triangle_profil:
    result = SketchLine.Create((Point2D.Create(M(r*-1), M(r*0))), (Point2D.Create(M(r*0.500), M(r*-0.866))))
    result = SketchLine.Create((Point2D.Create(M(r*-1), M(r*0))), (Point2D.Create(M(r*0.500), M(r*0.866))))
    result = SketchLine.Create((Point2D.Create(M(r*0.5), M(r*-0.866))), (Point2D.Create(M(r*0.500), M(r*0.866))))
#I Beam
elif i_beam_profil:
    result = SketchLine.Create((Point2D.Create(M(-W/2), M(H/2))), (Point2D.Create(M(W/2), M(H/2))))
    result = SketchLine.Create((Point2D.Create(M(W/2), M(H/2))), (Point2D.Create(M(W/2), M(H/2-VT))))
    result = SketchLine.Create((Point2D.Create(M(W/2), M(H/2-VT))), (Point2D.Create(M(HT/2), M(H/2-VT))))
    result = SketchLine.Create((Point2D.Create(M(HT/2), M(H/2-VT))), (Point2D.Create(M(HT/2), M(-H/2+VT))))
    result = SketchLine.Create((Point2D.Create(M(HT/2), M(-H/2+VT))), (Point2D.Create(M(W/2), M(-H/2+VT))))
    result = SketchLine.Create((Point2D.Create(M(W/2), M(-H/2+VT))), (Point2D.Create(M(W/2), M(-H/2))))
    result = SketchLine.Create((Point2D.Create(M(W/2), M(-H/2))), (Point2D.Create(M(-W/2), M(-H/2))))
    result = SketchLine.Create((Point2D.Create(M(-W/2), M(-H/2))), (Point2D.Create(M(-W/2), M(-H/2+VT))))
    result = SketchLine.Create((Point2D.Create(M(-W/2), M(-H/2+VT))), (Point2D.Create(M(-HT/2), M(-H/2+VT))))
    result = SketchLine.Create((Point2D.Create(M(-HT/2), M(-H/2+VT))), (Point2D.Create(M(-HT/2), M(H/2-VT))))
    result = SketchLine.Create((Point2D.Create(M(-HT/2), M(H/2-VT))), (Point2D.Create(M(-W/2), M(H/2-VT))))
    result = SketchLine.Create((Point2D.Create(M(-W/2), M(H/2-VT))), (Point2D.Create(M(-W/2), M(H/2))))
#Losange
elif losange_profil:
    result = SketchLine.Create((Point2D.Create(M(0), M(H/2))), (Point2D.Create(M(W/2), M(0))))
    result = SketchLine.Create((Point2D.Create(M(W/2), M(0))), (Point2D.Create(M(0), M(-H/2))))
    result = SketchLine.Create((Point2D.Create(M(0), M(-H/2))), (Point2D.Create(M(-W/2), M(0))))
    result = SketchLine.Create((Point2D.Create(M(-W/2), M(0))), (Point2D.Create(M(0), M(H/2))))
#Losange
elif rectangular_profil:
    result = SketchLine.Create((Point2D.Create(M(-W/2), M(H/2))), (Point2D.Create(M(W/2), M(H/2))))
    result = SketchLine.Create((Point2D.Create(M(W/2), M(H/2))), (Point2D.Create(M(W/2), M(-H/2))))
    result = SketchLine.Create((Point2D.Create(M(W/2), M(-H/2))), (Point2D.Create(M(-W/2), M(-H/2))))
    result = SketchLine.Create((Point2D.Create(M(-W/2), M(-H/2))), (Point2D.Create(M(-W/2), M(H/2))))
mode = InteractionMode.Solid
result = ViewHelper.SetViewMode(mode, None)
selection = Selection.Create(GetRootPart().Components[ComponentsIndex].Content.Bodies[0].Faces[0])
trajectories = Selection.Create(GetRootPart().Components[ComponentsIndex].Content.Curves[0])
sweep_strut = sweep(selection, trajectories)
ComponentsIndex += 1
#####
#Creating one component
#####
#Making everything active
result = ComponentHelper.SetRootActive(None)
#Selecting the strut
merge_bodies = []
for i in range(0, GetRootPart().Components.Count):
    compo = GetRootPart().Components[i]
    bodies = compo.GetBodies()
    for j in range(0, bodies.Count):
        merge_bodies.append(bodies[j])
#Selecting the spheres
for i in range(0, GetRootPart().Bodies.Count):
    merge_bodies.append(GetRootPart().Bodies[i])
#Merging everything together
targets = Selection.Create(merge_bodies)
result = Combine.Merge(targets)
#Deleting useless components, solids and the last curve in component 1
while GetRootPart().Components.Count > 1 :
    GetRootPart().Components[1].Delete()

```



```

while GetRootPart().Curves.Count > 0 :
    GetRootPart().Curves[0].Delete()
selection = Selection.Create(GetRootPart().Components[0].Content.Curves[0])
result = Delete.Execute(selection)
#####
#Extruding a cube
#####
pts_cube = [[0*Lx,0*Ly,0*Lz], [1*Lx,0*Ly,0*Lz],]
conn_cube = [[0, 1]]
PTS_Cube = []
for i in pts_cube:
    PTS_Cube.append(Point.Create(*i))
for i in conn_cube:
    selection = Selection.Create(GetRootPart())
    result = ComponentHelper.CreateNewComponent(selection, None)
    SketchLine.Create(PTS_Cube[i[1]], PTS_Cube[i[0]])
    unitVec = UnitVector(pts_cube[i[0]], pts_cube[i[1]])
    direc = Direction.Create(*unitVec)
    PointForDirection = (Point.Create(*pts_cube[i[0]]))
    frame = Frame.Create(PointForDirection, direc)
    plane = Plane.Create(frame)
    ViewHelper.SetSketchPlane(plane)
    point1 = Point2D.Create(M(0),M(0))
    point2 = Point2D.Create(M(0),M(Lz))
    point3 = Point2D.Create(M(Ly),M(0))
    result = SketchRectangle.Create(point1, point2, point3)
    mode = InteractionMode.Solid
    result = ViewHelper.SetViewMode(mode, None)
    selection = Selection.Create(GetRootPart().Components[1].Content.Bodies[0].Faces[0])
    trajectories = Selection.Create(GetRootPart().Components[1].Content.Curves[0])
    options = SweepCommandOptions()
    options.ExtrudeType = ExtrudeType.ForceIndependent
    options.Select = True
    result = Sweep.Execute(selection, trajectories, options, None)
#####
#Deleting extra line and Create Fluid Tissue and Solid Phase Components
#####
selection = Selection.Create(GetRootPart().Components[1].Content.Curves[0])
result = Delete.Execute(selection)
#Boolean function
targets = Selection.Create(GetRootPart().Components[1].Content.Bodies[0])
tools = Selection.Create(GetRootPart().Components[0].Content.Bodies[0])
options = MakeSolidsOptions()
result = Combine.Intersect(targets, tools, options)
# Delete everything outside the cube
selection = Selection.Create(GetRootPart().Components[0])
result = Delete.Execute(selection)
# Create New Component to put the solid domain
selection = Selection.Create(GetRootPart())
result = ComponentHelper.CreateNewComponent(selection, None)
# Move the solid domain into the new componant
selections = Selection.Create(GetRootPart().Components[0].Content.Bodies[1])
component = Selection.Create(GetRootPart().Components[1])
result = ComponentHelper.MoveBodiesToComponent(selections, component, False, None)
# Set Color
selection = Selection.Create(GetRootPart().Components[1].Content.Bodies[0])
options = SetColorOptions()

```

```

options.FaceColorTarget = FaceColorTarget.Body
ColorHelper.SetColor(selection, options, ColorHelper.Azure)
# Rename Object
selection = Selection.Create(GetRootPart().Components[1].Content.Bodies[0])
result = RenameObject.Execute(selection,"solid_domain")
# Rename 'Solid' to 'fluid_domain'
selection = Selection.Create(GetRootPart().Components[0].Content.Bodies[0])
result = RenameObject.Execute(selection,"fluid_domain")
# Rename 'Component2' to 'FD'
selection = Selection.Create(GetRootPart().Components[0])
result = RenameObject.Execute(selection,"FD")
# Rename 'Component3' to 'SD'
selection = Selection.Create(GetRootPart().Components[1])
result = RenameObject.Execute(selection,"SD")

# Modify Color
selection = Selection.Create(GetRootPart().Components[0].Content.Bodies[0])
options = SetColorOptions()
options.FaceColorTarget = FaceColorTarget.Body
ColorHelper.SetColor(selection, options, Color.FromArgb(100, 0, 128, 255))
#####
#Create a 2-by-2-by-4 Lattice
#####
#Create a component to put the plane
selection = Selection.Create(GetRootPart())
result = ComponentHelper.CreateNewComponent(selection, None)
#Create plane A
PointForDirection = (Point.Create(0*Lx,0*Ly,0*Lz))
direc = Direction.Create(0*Lx,0*Ly,1*Lz)
DatumPlaneCreator.Create(PointForDirection, direc)
#Create plane B
PointForDirection = (Point.Create(0*Lx,0*Ly,-Lz))
direc = Direction.Create(0*Lx,0*Ly,1*Lz)
DatumPlaneCreator.Create(PointForDirection, direc)
#Create plane C
PointForDirection = (Point.Create(0*Lx,0*Ly,0*Lz))
direc = Direction.Create(1*Lx,0*Ly,0*Lz)
DatumPlaneCreator.Create(PointForDirection, direc)
#Create plane D
PointForDirection = (Point.Create(0*Lx,0*Ly,0*Lz))
direc = Direction.Create(0*Lx,1*Ly,0*Lz)
DatumPlaneCreator.Create(PointForDirection, direc)
# Mirror Plane A
selection = Selection.Create(GetRootPart().Components[1].Content.Bodies[0])
ss = GetRootPart().Components[2].GetAllDatumPlanes()[0]
mirrorPlane = Selection.Create(ss)
options = MirrorOptions()
result = Mirror.Execute(selection, mirrorPlane, options, None)
selection = Selection.Create(GetRootPart().Components[0].Content.Bodies[0])
options = MirrorOptions()
result = Mirror.Execute(selection, mirrorPlane, options, None)
# Mirror Plane B
selection = Selection.Create(GetRootPart().Components[1].Content.Bodies[0])
ss = GetRootPart().Components[2].GetAllDatumPlanes()[1]
mirrorPlane = Selection.Create(ss)
options = MirrorOptions()
result = Mirror.Execute(selection, mirrorPlane, options, None)

```

```

selection = Selection.Create(GetRootPart().Components[0].Content.Bodies[0])
options = MirrorOptions()
result = Mirror.Execute(selection, mirrorPlane, options, None)
# Mirror Plane C
selection = Selection.Create(GetRootPart().Components[1].Content.Bodies[0])
ss = GetRootPart().Components[2].GetAllDatumPlanes()[2]
mirrorPlane = Selection.Create(ss)
options = MirrorOptions()
result = Mirror.Execute(selection, mirrorPlane, options, None)
selection = Selection.Create(GetRootPart().Components[0].Content.Bodies[0])
options = MirrorOptions()
result = Mirror.Execute(selection, mirrorPlane, options, None)
# Mirror Plane D
selection = Selection.Create(GetRootPart().Components[1].Content.Bodies[0])
ss = GetRootPart().Components[2].GetAllDatumPlanes()[3]
mirrorPlane = Selection.Create(ss)
options = MirrorOptions()
result = Mirror.Execute(selection, mirrorPlane, options, None)
selection = Selection.Create(GetRootPart().Components[0].Content.Bodies[0])
options = MirrorOptions()
result = Mirror.Execute(selection, mirrorPlane, options, None)
# Delete Objects
selection = Selection.Create(GetRootPart().Components[2])
result = Delete.Execute(selection)
# Activate Part: Component2
selection = Selection.Create(GetRootPart().Components[0])
result = ComponentHelper.SetActive(selection, None)
#####
#Create a Plate
#####
#Create a component to put the plane
selection = Selection.Create(GetRootPart())
result = ComponentHelper.CreateNewComponent(selection, None)
# Set Sketch Plane 1
sectionPlane = Plane.Create(Frame.Create(Point.Create((Lx),(Ly),(Lz)),
    -Direction.DirX,
    Direction.DirY))
result = ViewHelper.SetSketchPlane(sectionPlane, None)
# Set Sketch Plane 2
sectionPlane = Plane.Create(Frame.Create(Point.Create(MM(0), MM(0),(Lz)),
    -Direction.DirY,
    Direction.DirX))
result = ViewHelper.SetSketchPlane(sectionPlane, None)
# Sketch Rectangle
point1 = Point2D.Create((Ly),(-Lx))
point2 = Point2D.Create((-Ly),(-Lx))
point3 = Point2D.Create((-Ly),(Lx))
result = SketchRectangle.Create(point1, point2, point3)
# Solidify Sketch
mode = InteractionMode.Solid
result = ViewHelper.SetViewMode(mode, None)
# Extrude 1 Face
selection = Selection.Create(GetRootPart().Components[2].Content.Bodies[0].Faces[0])
options = ExtrudeFaceOptions()
options.ExtrudeType = ExtrudeType.Add
result = ExtrudeFaces.Execute(selection, (Lz/3), options)
#####

```

```

#Create a 4X4X4 Lattice
#####
if Four_by_Four:
    # Python Script, API Version = V19
    #Create a component to put the plane for the 4 by 4
    selection = Selection.Create(GetRootPart())
    result = ComponentHelper.CreateNewComponent(selection, None)
    #Create plane E
    PointForDirection = (Point.Create(1*Lx,0*Ly,0*Lz))
    direc = Direction.Create(1*Lx,0*Ly,0*Lz)
    DatumPlaneCreator.Create(PointForDirection, direc)
    #Create plane F
    PointForDirection = (Point.Create(0*Lx,1*Ly,0*Lz))
    direc = Direction.Create(0*Lx,1*Ly,0*Lz)
    DatumPlaneCreator.Create(PointForDirection, direc)
    # Mirror Plane E
    selection = Selection.Create(GetRootPart().Components[1].Content.Bodies[0])
    ss = GetRootPart().Components[3].GetAllDatumPlanes()[0]
    mirrorPlane = Selection.Create(ss)
    options = MirrorOptions()
    result = Mirror.Execute(selection, mirrorPlane, options, None)
    selection = Selection.Create(GetRootPart().Components[0].Content.Bodies[0])
    options = MirrorOptions()
    result = Mirror.Execute(selection, mirrorPlane, options, None)
    selection = Selection.Create(GetRootPart().Components[2].Content.Bodies[0])
    options = MirrorOptions()
    result = Mirror.Execute(selection, mirrorPlane, options, None)
    # Mirror Plane F
    selection = Selection.Create(GetRootPart().Components[1].Content.Bodies[0])
    ss = GetRootPart().Components[3].GetAllDatumPlanes()[1]
    mirrorPlane = Selection.Create(ss)
    options = MirrorOptions()
    result = Mirror.Execute(selection, mirrorPlane, options, None)
    selection = Selection.Create(GetRootPart().Components[0].Content.Bodies[0])
    options = MirrorOptions()
    result = Mirror.Execute(selection, mirrorPlane, options, None)
    selection = Selection.Create(GetRootPart().Components[2].Content.Bodies[0])
    options = MirrorOptions()
    result = Mirror.Execute(selection, mirrorPlane, options, None)
    # Delete Plane
    selection = Selection.Create(GetRootPart().Components[3])
    result = Delete.Execute(selection)
    #####
    # Activate Part: Component2
    selection = Selection.Create(GetRootPart().Components[0])
    result = ComponentHelper.SetActive(selection, None)
    #####
    # Create Named Selection Group For Material Properties Assignment
    #####
    primarySelection = Selection.Create(GetRootPart().Components[0].Content.Bodies[0])
    secondarySelection = Selection()
    result = NamedSelection.Create(primarySelection, secondarySelection)
    result = NamedSelection.Rename("Group1", "Granulation_tissue")
    primarySelection = Selection.Create(GetRootPart().Components[1].Content.Bodies[0])
    secondarySelection = Selection()
    result = NamedSelection.Create(primarySelection, secondarySelection)
    result = NamedSelection.Rename("Group1", "Scaffold")

```

```

#####STRUCTURE COMPLETED#####
#####
# Output Lattice Characteristic in Text File
#####
# Measure volume, Surface Area, Relative Density and Porosity
area = GetRootPart().Components[1].Content.Bodies[0].Shape.SurfaceArea
area_mm2 = area*1000000
volume_fluid = GetRootPart().Components[0].Content.Bodies[0].Shape.Volume
volume_strut = GetRootPart().Components[1].Content.Bodies[0].Shape.Volume
Relative_density = repr((volume_strut/(volume_strut+volume_fluid))*100)
Porosity = repr(100 - (volume_strut/(volume_strut+volume_fluid))*100)
# Write unit cell info to a file
print(area_mm2)
unit_info = open("file_destination.txt", "a")
str_area_mm2 = repr(area_mm2)
str_radius = repr((r*1000000))
str_lenght = repr((L*1000000))
unit_info.write( Unit_name +
    "\t" + "Lenght_um = " + "\t" + str_lenght +
    "\t" + "radius_um = " + "\t" + str_radius +
    "\t" + "area_mm2 = " + "\t" + str_area_mm2 +
    "\t" + "Relative_density = " + "\t" + Relative_density +
    "\t" + "Porosity = " + "\t" + Porosity +
    "\n")
unit_info.close()
#####

```

Appendix H - ACT-Python Script for Material Assignment

```
#Fluid_Domain (granulation tissue - 09a3bc4f-f372-421e-8632-08ee7545e064)
with Transaction():
    _Object = Model.Children[0].Children[1]
    _Object.Assignment = "09a3bc4f-f372-421e-8632-08ee7545e064"
#Solid_Domain (scaffold (polymer) - 75c04f31-19c3-4ceb-973d-364541cc0819)
with Transaction():
    _Object = Model.Children[0].Children[0]
    _Object.Assignment = "75c04f31-19c3-4ceb-973d-364541cc0819"
#Steel plate (steel - aaf2c75d-7151-4602-bf0e-6d886b3a73e4)
with Transaction():
    _Object = Model.Children[0].Children[2]
    _Object.Assignment = "aaf2c75d-7151-4602-bf0e-6d886b3a73e4"

i = Model.Children[0].Children[1].Children[0].ObjectId
gran = DataModel.GetObjectById(i)
cs = gran.AddCommandSnippet()
i = Model.Children[0].Children[0].Children[0].ObjectId
gran = DataModel.GetObjectById(i)
cs = gran.AddCommandSnippet()

gran = Model.Children[0].Children[1].Children[0].Children[0].ObjectId
scaffold = Model.Children[0].Children[0].Children[0].Children[0].ObjectId

DataModel.GetObjectById(gran).Input = r""""!Commands will be executed after material definitions in /PREP7.
! Active UNIT system in Workbench when this object was created: Metric (um, kg, uN, s, V, mA)
E=0.2          !Young's modulus x
NU=0.167       !Poisson's ratio
KX=10000       !Permeability X
KY=10000       !Permeability Y
KZ=10000       !Permeability Z
BMS=2300       !Bulk Modulus of solid
BMF=2300       !Bulk Modulus of fluid
PR=0.8         !Porosity
/PREP7
ET,1,CPT217    !Choose the element type (Coupled-pore-pressure, tetrahedral elements, poroelastic,fully direct coupling)
KEYOPT,1,12,1  !element type number 1, then knum number 12 Structural-pore-fluid-diffusion
MP,EX,1,E      !Assing the Young's modulus
MP,EY,1,E
MP,EZ,1,E
MP,NUXY,1,NU   !Assing the Poisson's ratio
MP,NUYZ,1,NU
MP,NUXZ,1,NU
TB,PM,1,,,PERM !Assing permeability
TB,DATA,1,KX,KY,KZ
TB,PM,1,,,SP   !Assing Bulk modulus solid
TB,DATA,1,BMS
TB,PM,1,,,FP   !Assing Bulk modulus and porosity for Fluid
TB,DATA,1,BMF,,PR
SHPP,ON
VMESH,ALL
""""

DataModel.GetObjectById(scaffold).Input = r""""!Commands will be executed after material definitions in /PREP7.
! Active UNIT system in Workbench when this object was created: Metric (um, kg, uN, s, V, mA)
E_S=1000       !Young's modulus
```

```

NU_S=0.3    !Poisson's ratio
KX_S=10000  !Permeability X
KY_S=10000  !Permeability Y
KZ_S=10000  !Permeability Z
BMS_S=13920 !Bulk Modulus of solid
BMF_S=2300  !Bulk Modulus of fluid
PR_S=0.5    !Porosity
/PREP7
ET,2,CPT217 !Choose the element type (Coupled-pore-pressure, tetrahedral elements, poroelastic,fully direct coupling)
KEYOPT,2,12,1 !element type number 2, then knum number 12 Structural-pore-fluid-diffusion
MP,EX,2,E_S  !Assing the Young's modulus
MP,EY,2,E_S
MP,EZ,2,E_S
MP,NUXY,2,NU_S !Assing the Poisson's ratio
MP,NUYZ,2,NU_S
MP,NUXZ,2,NU_S
TB,PM,2,,,PERM !Assing permeability
TB,DATA,1,KX_S,KY_S,KZ_S
TB,PM,2,,,SP !Assing Bulk modulus solid
TB,DATA,1,BMS_S
TB,PM,2,,,FP !Assing Bulk modulus and porosity for Fluid
TB,DATA,1,BMF_S,,PR_S
SHPP,ON""""

```


Appendix I - APDL Script for Boundary Conditions

```
/PREP7
CMSEL,S,wall_1f_press !Select the surface "name selection"
D,ALL,PRES,0 !Boundary condition (fluid pressure = 0)
ALLSEL,ALL !reselect everything for the next steps
CMSEL,S,wall_2f_press
D,ALL,PRES,0
ALLSEL,ALL
CMSEL,S,wall_3f_sym !Symmetry
DSYM,SYMM,Y
ALLSEL
CMSEL,S,wall_4f_sym !Symmetry
DSYM,SYMM,X
ALLSEL
OUTRES,ALL,ALL !activate solution result data
/SOLU
```

Appendix J - APDL Script for Post-Processing Solutions

```
/POST1
SET, LAST
ETABLE, E1, EPEL, 1          !Principal strain 1 = E1
ETABLE, E2, EPEL, 2          !Principal strain 2 = E2
ETABLE, E3, EPEL, 3          !Principal strain 3 = E3
!Octahedral Shear Strain calculation
SADD, E12, E1, E2, 1, -1
SADD, E23, E2, E3, 1, -1
SADD, E13, E1, E3, 1, -1
SEXP, E12_2, E12, , 2
SEXP, E23_2, E23, , 2
SEXP, E13_2, E13, , 2
SADD, E12_E23, E12_2, E23_2
SADD, E123_2, E13_2, E12_E23
SEXP, E123_SQRT, E123_2, , 1/2
SMULT, OSS, E123_SQRT, , 2/3
SMULT, BSS, OSS, , 1/0.0375
!
ALLSEL
ETABLE, FLOW, FLOW          !Fluid Flow Sum of Element Nodal Values
ETABLE, FLUX_X, FFLX, X      !Fluid Flow Flux X in poromechanics
ETABLE, FLUX_Y, FFLX, Y      !Fluid Flow Flux Y in poromechanics
ETABLE, FLUX_Z, FFLX, Z      !Fluid Flow Flux Z in poromechanics
ETABLE, VOLUME, VOLU

/OUTPUT, pretab_test4
PRETAB, E1, E2, E3, BSS, FLOW, FLUX_X, FLUX_Y, FLUX_Z, VOLUME
/OUT
```

Appendix K - MATLAB Script for Biophysical Stimulus Calculation

```

fclose('all')
clc
clear all
%%
DP_results=xlsread('Ansys_Data_Point_Output_CSV_File');
%create new row to get radius in microns(Strut Diameter(m) x 200000)
DP_results(:,50) = DP_results(:,3).*2000000;
dp=(DP_results(:,1)).';%row 1 = data point number
radius=(DP_results(:,50)).';;%row 3 et 50 = radius in um;
pressure=(DP_results(:,27)).';%row 27 = Pression MPa

for i_A_results=1:length(dp)
    dpstring=num2str(dp(i_A_results));
    ds=strcat('Workbench_projects_location\dp',dpstring,'\SYS-2\MECH\ds.dat');
    pretab=strcat('Workbench_projects_location\dp',dpstring,'\SYS-2\MECH\pretab_test4');
    geometry_trial=[dp(i_A_results); radius(i_A_results); pressure(i_A_results)];
%Erase the following workspace table
evalin('base','clear BONE_RESULTS')
evalin('base','clear e_stats')
evalin('base','clear elements_fluid_D')
evalin('base','clear elements_solid_D')
evalin('base','clear Fluid_elements')
evalin('base','clear Fluid_info_Array')
evalin('base','clear indices')
evalin('base','clear mat')
evalin('base','clear nblock_details')
evalin('base','clear nodes')
evalin('base','clear node_count')
evalin('base','clear nblock_length')
evalin('base','clear tline')
evalin('base','clear Solid_elements')
evalin('base','clear Solid_info_Array')
evalin('base','clear VelocityArray')
evalin('base','clear Element_removal')
evalin('base','clear Element_strut')
evalin('base','clear Element_velocity')
tic %Start the timer to time the MATLAB part of the computation process
fid = fopen(ds); %open "ds" file from ANSYS folder
tline = fgetl(fid); %Read from the specific line
is_nblock = false
nblock_length = 0
node_count = 1
while ischar(tline) %When tline is a character array, repeat procedure
    if contains(tline, 'nblock') %If tline start with "nblock", head to the next step and is_nblock becomes true
        is_nblock = true
        nblock_details = strsplit(tline, ',') %The first following line get split at each commas
        nblock_length = nblock_details(3) %The third number is stored
        nblock_length = cell2mat(nblock_length) %Next two step switch that number from string to number
        nblock_length = str2num(nblock_length)
        tline = fgetl(fid);
        continue
    end
    if string(tline) == '-1' %If tline contains -1 is_nblock becomes false and the code stop
        is_nblock = false

```

```

end
if is_nblock %If is_block is still true the loop continues
    if tline(1) == '(' %Ignore the line right after nblock
        tline = fgetl(fid);
        continue
    end
    C = strsplit(tline); %split tline with spaces
    for r = 2:length(C)
        nodes(node_count, r-1) = str2num(cell2mat(C(r))) ;
    end
    node_count = node_count + 1;
end
tline = fgetl(fid);
end
fclose(fid);
%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%% Body 2 - SOLID DOMAIN %%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%
fid = fopen(ds); %open file
tline = fgetl(fid); %Read lines of the file
is_eblock = false
eblock_length = 0
element_count = 1
while ischar(tline) %When tline is a character array, repeat procedure
    if contains(tline, 'Elements for Body 2') %If tline start with "Elements for Body 2", is_nblock becomes true
        is_eblock = true
        tline = fgetl(fid);
        continue
    end
    if string(tline) == '-1' %If tline contains -1 is_eblock becomes false and the code stop
        is_eblock = false
    end
    if is_eblock %If is_block is still true the loop continues
        if tline(1) == '(' %Ignore the line right after nblock
            tline = fgetl(fid);
            continue
        end
        C = strsplit(tline); %split tline with spaces
        for r = 2:length(C)
            elements_solid_D(element_count, r-1) = str2num(cell2mat(C(r))) ;
        end
        element_count = element_count + 1;
    end
    tline = fgetl(fid);
end
fclose(fid);
%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%% BODY 1 - FLUID BODY %%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%
fid = fopen(ds); %open file
tline = fgetl(fid); %Read lines of the file
is_eblock = false
eblock_length = 0
element_count = 1
while ischar(tline) %When tline is a character array, repeat procedure
    if contains(tline, 'Elements for Body 1') %If tline start with "Elements for Body 1", is_nblock becomes true
        is_eblock = true
        tline = fgetl(fid);
        continue
    end
    if string(tline) == '-1' %If tline contains -1 is_eblock becomes false and the code stop

```

```

        is_eblock = false
    end
    if is_eblock %If is_block is still true the loop continues
        if tline(1) == '(' %Ignore the line right after nblock
            tline = fgetl(fid);
            continue
        end
        C = strsplit(tline);
        for r = 2:length(C)
            elements_fluid_D(element_count, r-1) = str2num(cell2mat(C(r))) ;
        end
        element_count = element_count + 1;
    end
    tline = fgetl(fid);
end
fclose(fid);
%%%%Remove zeros
for i=3:2:length(elements_fluid_D)
    Fluid_elements(i,:) = elements_fluid_D(i,11:15);
end
indices=find(Fluid_elements(:,1)==0);
Fluid_elements(indices,:)=[];

for i=3:2:length(elements_solid_D)
    Solid_elements(i,:) = elements_solid_D(i,11:15);
end
indices=find(Solid_elements(:,1)==0);
Solid_elements(indices,:)=[];
%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%% GET THE RESULTS FROM THE PRETAB %%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%
fid = fopen(pretab);
tline = fgetl(fid);
is_e_stats = false
e_stats_count = 1
e_stats = zeros(1,10);
while ischar(tline) %Remove 11 line after the word ANSYS
    if contains(tline, '*** ANSYS')
        for r = 1:12
            tline = fgetl(fid);
        end
    end
    if contains(tline, 'FLUX_X')%If the line contains "flux_x" the code run the next line
        is_e_stats = true
        tline = fgetl(fid);
        continue
    end
    if contains(tline, 'MINIMUM VALUES')%If the line contains "Minimum Values" the code stop
        is_e_stats = false
    end
    if is_e_stats %Seperates the characters
        if length(tline) == 0
            tline = fgetl(fid);
            continue
        end
        [mat,tok,ext] = regexp(tline, '(\{9\})(\{13\})(\{13\})(\{13\})(\{13\})(\{13\})(\{13\})(\{13\})(\{13\})(\{13\})(.+)', 'match', 'tokens');
        if length(tok) == 0
            tline = fgetl(fid);
            continue
        end
    end
end

```



```

%% Column 5 = BS
%% Column 6 and 7 = bone or no bone
%% Column 8 = Store bone volume
%% Column 9 = Strain bone or no bone
%% Column 10 = Strain bone or no bone
%% Column 11 = Velocity bone or no bone
%% Column 12 = Velocity bone or no bone
%% Column 13 = strain Store bone volume
%% Column 14 = Velocity Store bone volume
%% %%%%%%%%%%%%%%% For Solid_indo_Array %%%%%%%%%%%%%%%
%% Column 1 = Element number
%% Column 2 = Volume
Fluid_info_Array(:,4)=(VelocityArray(:,1))/3 %had the velocity to the fluid_info_array
%Calculate the BS in a 5th column
Fluid_info_Array(:,5) = Fluid_info_Array(:,2) + Fluid_info_Array(:,4);
%Evalute if it's bone or not in column 6 and 7
Fluid_info_Array(:,6)= 0.01>=Fluid_info_Array(:,5)<=0.53;
Fluid_info_Array(:,7)= Fluid_info_Array(:,5)<=0.53;
%Store bone volume in 8th column
Fluid_info_Array(:,8) = Fluid_info_Array(:,3).*Fluid_info_Array(:,6).*Fluid_info_Array(:,7);
%calculate bone ingrowth percentage
Bone_volume = sum(Fluid_info_Array(:,8));
%%
%Official bone ingrowth percentage
Total_solid_fluid_Volume = sum(Fluid_info_Array(:,3)) + sum(Solid_info_Array(:,2));
Total_fluid_Volume = sum(Fluid_info_Array(:,3));
Total_bone_Volume = sum(Fluid_info_Array(:,8));
Fluid_Velocity_Average = mean(Fluid_info_Array(:,4))
Strain_Average = mean(Fluid_info_Array(:,2))
BS_Average = mean(Fluid_info_Array(:,5))
%% % % % % ifjustvelocity_percentage = (Bone_volume_ifjustvelocity/Total_solid_fluid_Volume)*100
%% % % % % ifjuststrain_percentage = (Bone_volume_ifjuststrain/Total_solid_fluid_Volume)*100
Bone_ingrowth_percentage_Fluid = (Total_bone_Volume/Total_fluid_Volume)*100
Bone_ingrowth_percentage = (Total_bone_Volume/Total_solid_fluid_Volume)*100
Relative_Density = ((sum(Solid_info_Array(:,2)))/Total_solid_fluid_Volume)*100;
Structure_porosity = (Total_fluid_Volume/Total_solid_fluid_Volume)*100;
%%
%%Store the element numbers that contains optimal stimulation for bone
Fluid_info_Array(:,15)= Fluid_info_Array(:,8)>1
Fluid_info_Array(:,16)= Fluid_info_Array(:,15).*Fluid_info_Array(:,1);
Element_removal(:,1) = Fluid_info_Array(:,16);
indices=find(Element_removal(:,1)==0); %remove all the zeroes
Element_removal(indices,:)=[]; %#ok<SAGROW>
writematrix(Element_removal,'File_location\Element_bone.csv')
%Create a file with the element forming the strut
Element_strut(:,1) = Solid_elements(:,1);
writematrix(Element_strut,'File_location \Element_strut.csv')
Element_velocity(:,1) = Fluid_info_Array(:,1);
Element_velocity(:,2) = Fluid_info_Array(:,4);
writematrix(Element_velocity,'File_location \Element_velocity.csv')
%%
source=strcat('File_location\Element_bone.csv');
destination=strcat('Workbench_projects_location\dp',dpstring,'\Element_bone.csv');
status = copyfile(source, destination);
source2=strcat('File_location\Element_strut.csv');
destination2=strcat('Workbench_projects_location\dp',dpstring,'\Element_strut.csv');
status2 = copyfile(source2, destination2);

```

```

source3=strcat('File_location\Element_velocity.csv');
destination3=strcat('Workbench_projects_location\dp',dpstring,'\Element_velocity.csv');
status3 = copyfile(source3, destination3);
%%
Time_A = toc/60 %Stop Computation time timer
BONE_RESULTS{i_A_results}(1,1)= dp(i_A_results);
BONE_RESULTS{i_A_results}(1,2)= radius(i_A_results);
BONE_RESULTS{i_A_results}(1,3)= pressure(i_A_results);
BONE_RESULTS{i_A_results}(1,4)= Bone_ingrowth_percentage;
BONE_RESULTS{i_A_results}(1,5)= Bone_ingrowth_percentage_Fluid;
BONE_RESULTS{i_A_results}(1,8)= BS_Average;
BONE_RESULTS{i_A_results}(1,9)= Strain_Average;
BONE_RESULTS{i_A_results}(1,10)= Fluid_Velocity_Average;
BONE_RESULTS{i_A_results}(1,11)= Time_A;
BONE_RESULTS{i_A_results}(1,12)= Relative_Density;
BONE_RESULTS{i_A_results}(1,13)= Structure_porosity;

fid = fopen('File_location \Results_Automatic','a');
fprintf(fid,'\n',BONE_RESULTS{i_A_results});
fprintf(fid,'\t%f',BONE_RESULTS{i_A_results});
fclose(fid);
%%
end

```

Appendix L - ACT-Python Script for Element Optimal for Bone

```
items = []
import csv
with open(r"File_Location\\Element_delete.csv") as csvDataFile:
    csvReader = csv.reader(csvDataFile)
    for row in csvReader:
        items.append(int(row[0]))

with open(r" File_Location\\Element_strut.csv") as csvDataFile:
    csvReader = csv.reader(csvDataFile)
    for row in csvReader:
        items.append(int(row[0]))

NS = Model.Children[6].ObjectId
NS_id = DataModel.GetObjectById(NS)
NS_List = NS_id.AddNamedSelection()

##Assing the name selection to a worksheet
List_ID = Model.Children[6].Children[8].ObjectId

#region Set current scoping selection
currentSelection =
ExtAPI.SelectionManager.CreateSelectionInfo(Ansys.ACT.Interfaces.Common.SelectionTypeEnum.GeometryEntities)
currentSelection.Ids = []
ExtAPI.SelectionManager.NewSelection(currentSelection)

#region Set current scoping selection
currentSelection =
ExtAPI.SelectionManager.CreateSelectionInfo(Ansys.ACT.Interfaces.Common.SelectionTypeEnum.MeshElements)
currentSelection.Ids = items
ExtAPI.SelectionManager.NewSelection(currentSelection)
```