

Computational Assessment of 3D Printed Vertebral Surrogates

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INTRODUCTION: Additive manufacturing (AM) is rapidly expanding the capabilities for custom design and fabrication. While Human Body Modeling (HBM) is an established field with many physical and computational models readily available, AM is continuing to reduce the manufacturing limitations that result in differences between physical surrogates, their computational counterparts, and the human body. The anisotropic nature of trabecular bone produces distinctly unique properties that have heretofore been exceedingly difficult to replicate in manufacturable materials. Increasing work is being completed to generate uniquely complex structures to model the properties of trabecular bone, applying functional grading as a method of inducing anisotropic properties in AM materials. However, the micro-mechanical basis for these models is often neglected. As such, methods are needed for computationally comparing these structures to those that have been validated for use in existing HBMs. Furthermore, these HBMs may benefit from the introduction of a geometry that can better represent stress concentrations. Previous work completed using Global HBM Consortium (GHBMC) models on the biomechanical simulation of vertebral loading was utilized as the basis for the prescribed loads in the following work. The interaction between L4 and L5 at rest and in a representative automotive impact was modeled, with differences in structural response being assessed. The GHBMC M50 L5 is used, given the difference in moduli between the trabecular region and the cortical region. Because material properties influence geometric features of fabricated components, shown in **Figure 1**, it is hypothesized that if a material is too rigid or elastic, it will perform poorly in biomimicry.

METHODS: Five materials common in biomedical AM are tested in simulation by applying reported material properties (**Table 1**) to the finite element (FE) models. The popular surface gyroid is modeled using a FE shell mesh, reducing mesh complexity as per computational constraints. The same shell mesh is used across multiple material simulations, only changing shell thickness as this is the single parameter to change when switching to materials of greater or lesser stiffness while keeping effective microstructural elasticity constant. The required volume fraction, and thus the required structural thickness, is calculated based on the desired elastic modulus of the trabecular structure, a constant among all simulations based on the GHBMC model assuming linear behavior (0.38 GPa). The desired volume fraction of the gyroid structure is calculated using equations from Gibson et. al., relating the volume fraction of a fine cell-based structure to its properties relative to the bulk material. From this, the desired structural thickness is calculated using an empirically derived relationship between gyroid volume fraction and thickness from MATLAB and nTopology, the software used to create the initial shell mesh and perform all previously mentioned calculations. Once these steps have been completed, the result is the gyroid shell model depicted in **Figure 2A**. Cortical faces are tied to trabecular nodes with a 0.1 mm tolerance, and pressure is applied to the superior cortical mesh, reflecting loading at rest and in an automotive accident in a 50th percentile male. Loading of L5 was simulated in sitting conditions, and in crash conditions based on: (1) the loads acting on lumbar spine while sitting: $F_{L4L5, \text{sitting}} \approx 84.94 \text{ N}$; (2) the biomechanics of lumbar segments in compression: $F_{L4L5, \text{accident}} \approx 10.94 \text{ kN}$ (130x seated), at 2.24 MPH. Implicit and modal simulations were performed for both states. **Figure 2B** shows the GHBMC L5, with the superior cortical elements loaded normally and the inferior nodes being fixed. The same procedure for tying nodes is used for the solid element GHBMC model and the shell element gyroid model.

RESULTS: In the GHBMC M50 model, the trabecular region has a 0.38 GPa elastic modulus, and the cortical region has an 18.44 GPa elastic modulus. The properties of the printable materials are shown in **Table 1**, including elastic modulus, elongation at break, and ultimate strength. 95th percentile maximum strain, 95th percentile maximum Von Mises Stress (VMS), and failure outcomes relative to max elongation and/or maximum strength are reported for the static analyses. Additionally, for the modal analyses, the minimum buckling load factor as well as presence of buckling failure are reported. In both the accident and seated scenarios, Fortify DT has 95th percentile max principal strain and VMS most like that in the GHBMC model. In terms of VMS, the response has % difference (PD) of 1.38% and 0.62% in the seated and accident scenarios. Otherwise, average PD is about 80% for the other materials. In modal response, average PD is over 140% for the other materials, again with the GHBMC as the baseline and PD being computed per material/condition. The buckling load factor PD in an accident scenario with Fortify DT is 17.7%, with PD in the seated scenario being on par with the average material PD.

DISCUSSION: When printing complex pieces, it is common to push the capabilities of the printer by utilizing a structural thickness related to printer resolution. However, in simulation, the limiting constraint becomes the number of elements. As unit cell size decreases, the number of elements increases significantly with edge length decreasing and the number of unit cells increasing. This influences mesh size drastically. As this work is computational, replicates are not used. Via logistic regression, as rigidity increases, so do the odds of material failure in an accident ($p=0.0095$). Neither modal nor material failure are observed when sitting. Supporting the hypothesis, an optimal balance of thickness and mechanics yields analogous properties.

SIGNIFICANCE AND CLINICAL RELEVANCE: This work yields insight into the effect that material has on structure, and how those structural changes influence the biomimetic efficacy of a design. While these outcomes are found based on investigation into vertebral surrogates, this work has the potential to model other biomechanical surrogates, including that of the femur. The significance of this work is that it shows how analysis of evolving AM materials/structures can be completed relative to unconventional biological structure, advancing the HBM and biomimicry fields. Furthermore, it highlights a fatal flaw in the use of Titanium for such stiffness related biomimetic designs. In general, structures need to be extremely thin to accurately attain the effective elastic properties of trabecular bone. This investigation may influence implant design in the future, bearing clinical relevance.

Table 1. Tabulated results of FEA:

Material	Input			Condition	Thickness (mm)	Output				
	Young's Modulus (GPa)	Elongation at Break (%)	Ultimate Strength (MPa)			95th Percentile Max Principal Strain (%)	95th Percentile Von-Mises Stress (MPa)	Failure in Compression?	Min. Buckling Load Factor	Failure in Buckling?
Ti-6Al-4V	116	10	1000	Sitting	0.0458	0.01	12.50	NO	18.440	NO
				Accident	1.60	1663.17	18.44	YES	0.902	YES
Fortify DT	7.35	1.8	90	Sitting	0.5146	0.01	0.72	NO	2108.914	NO
				Accident	1.43	92.62	18.44	YES	16.375	NO
CE 221	3.9	3	85	Sitting	0.8972	0.01	0.39	NO	3028.044	NO
				Accident	1.45	50.02	18.44	NO	27.217	NO
IN605	1.5	120	42	Sitting	2.0744	0.01	0.15	NO	1001.443	NO
				Accident	1.40	18.79	18.44	NO	38.813	NO
UMMA 90	1.4	30	30	Sitting	2.3038	0.01	0.14	NO	1014.433	NO
				Accident	1.39	17.47	18.44	NO	40.021	NO
GHBMC	0.38	0.5	3.9 Trils 1460 Com	Sitting		0.01	0.72	NO	403.486	NO
				Accident		0.76	93.40	YES	13.703	NO

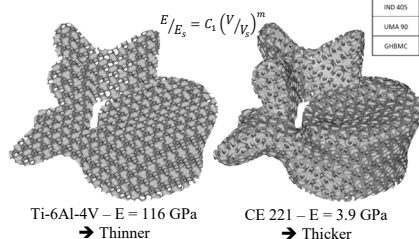


Figure 1. Influence of material variation on geometry of a unit-cell-stiffness-matched L5 surrogate.

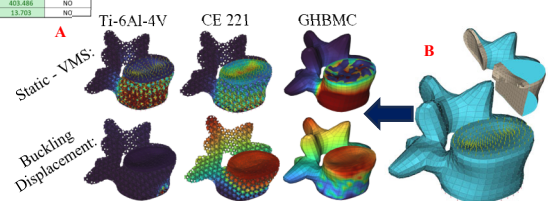


Figure 2. (A) GHBMC L5 model cross-section and distributed load.; (B) VMS & buckling displacement maps in L5. Red = critical stress/load.