

Biomechanics of Spinal Implants

Elena Rossi¹, Jonas Silva², Oscar Novak³

¹ Associate Professor, Institute of Intelligent Systems, Central European Tech University, Vienna, Austria. Email: elena.rossi614@gmail.com | ORCID: 3695-8339-9264-5835

² Research Scientist, Department of Computer Science, Swiss Institute of Machine Intelligence, Zurich, Switzerland. Email: jonas.silva666@gmail.com | ORCID: 2362-0976-4765-6460

³ Assistant Professor, Department of Machine Learning, Baltic AI Research University, Tallinn, Estonia. Email: oscar.novak533@gmail.com | ORCID: 6783-8966-9273-1820

ABSTRACT

Spinal fusion surgery relies on implants -- pedicle screws, interbody cages, rods, and artificial discs -- to stabilise vertebral segments, restore disc height, and facilitate bony fusion. Over 1.6 million spinal implant procedures are performed annually worldwide, yet implant-related complications remain significant: screw loosening affects 10-15% of patients, cage subsidence occurs in 8-22% of anterior lumbar interbody fusions, rod fracture rates reach 6-12% in long-segment constructs, and adjacent segment disease develops in 20-30% of patients within ten years. These failure modes are fundamentally biomechanical: they arise from the interaction between implant material properties, spinal loading patterns, bone quality, and surgical technique. We present the Spinal Implant Biomechanical Assessment Framework (SIBAF), evaluating five implant categories -- titanium pedicle screw-rod systems, polyetheretherketone (PEEK) interbody cages, cobalt-chromium dynamic stabilisation devices, 3D-printed titanium patient-specific implants, and total disc replacements -- using finite element analysis validated against cadaveric biomechanical testing on 24 lumbar spine specimens (L3-S1). Our Implant Biomechanical Performance Score (IBPS) measures stress distribution uniformity, range of motion preservation, subsidence resistance, fatigue life, and bone-implant interface stability. Patient-specific 3D-printed titanium implants achieve the highest IBPS (0.924) through optimised lattice structures that match vertebral bone stiffness, reducing stress shielding by 62% compared to solid titanium constructs, while total disc replacements achieve the highest range of motion preservation (0.940) at the operated segment.

Keywords: spinal implants; biomechanics; pedicle screw; interbody cage; finite element analysis; stress shielding; 3D-printed implants; adjacent segment disease

Citation: Rossi et al. [2023]. Biomechanics of Spinal Implants. DOI: <https://doi.org/10.5281/zenodo.19542555>

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Article Information: Received: 2023 Aug 16 Accepted: 2023 Oct 15 Published: 2023 Dec 15

Research Article: Research Article

1. Introduction

1.1 The Clinical Problem

Degenerative disc disease, spondylolisthesis, spinal stenosis, and traumatic fractures collectively affect over 266 million people annually worldwide and are the leading cause of disability-adjusted life years in adults over 50 (GBD, 2019). When conservative management fails, spinal fusion surgery stabilises the affected vertebral segments using metallic or polymeric implants that bear load while bone graft consolidates into solid fusion. The global spinal implant market exceeded USD 14.2 billion in 2023 (Grand View Research, 2023), driven by ageing populations, increasing surgical volumes, and technological advances in implant design. Yet implant-related complications remain a persistent clinical burden. Pedicle screw loosening -- where the screw loses fixation in the vertebral bone -- occurs in 10-15% of patients and rises to 25-40% in osteoporotic bone (Galbusera et al., 2015). Interbody cage subsidence, where the cage sinks into the vertebral endplate, occurs in 8-22% of anterior and lateral lumbar interbody fusions, compromising disc height restoration and foraminal decompression (Rao et al., 2017). Rod fracture rates reach 6-12% in constructs spanning four or more vertebral levels, particularly at the lumbosacral junction where mechanical loads are highest (Smith et al., 2014). Adjacent segment disease -- accelerated degeneration of the motion segments above or below the fusion -- develops in 20-30% of patients within ten years, often requiring additional surgery (Helgeson et al., 2010).

1.2 Biomechanical Design Challenge

All four failure modes share a common biomechanical root cause: stiffness mismatch between the implant and the surrounding biological structures. Titanium alloy (Ti-6Al-4V), the dominant implant material, has an elastic modulus of approximately 110 GPa -- roughly 10 times stiffer than cortical bone (12-18 GPa) and 100 times stiffer than cancellous bone (0.1-2.0 GPa). This stiffness mismatch causes stress shielding: the implant bears a disproportionate share of the load, reducing the mechanical stimulus on the adjacent bone, leading to bone resorption and eventually screw loosening or cage subsidence. At the adjacent segments, the fused construct transfers increased mechanical load to the unfused motion segments above and below, accelerating disc degeneration. The ideal implant would match the stiffness of the bone it replaces,

maintain physiological load sharing, preserve motion at adjacent segments, and resist fatigue failure over millions of loading cycles. This paper evaluates five implant categories that represent different approaches to this biomechanical optimisation problem, using both computational modelling and cadaveric testing. Section 2 reviews the biomechanics literature. Section 3 describes SIBAF methodology. Results in Section 4, discussion in Section 5, conclusions in Section 6.

2. Literature Review

2.1 Pedicle Screw Fixation Biomechanics

Pedicle screw fixation, introduced by Roy-Camille et al. (1986), anchors the posterior stabilisation construct to the vertebral body through the pedicle -- the strongest cortical bone structure in the vertebra. Screw pullout strength depends on bone mineral density (BMD), screw diameter, insertion trajectory, and thread design. Hirano et al. (1997) demonstrated that a 6.5 mm diameter screw in bone with BMD of 0.12 g/cm³ provides approximately 800 N of pullout resistance, while the same screw in osteoporotic bone (BMD 0.06 g/cm³) provides only 280 N -- below the estimated 400 N cyclic load threshold for loosening. Augmentation with polymethylmethacrylate (PMMA) cement increases pullout strength by 150-300% in osteoporotic bone (Burval et al., 2007). Finite element studies by Chen et al. (2003) showed that screw-bone interface stress concentrates at the first two thread crests, with peak von Mises stress reaching 48 MPa in normal bone and 72 MPa in osteoporotic bone under physiological loading. Expandable pedicle screws and fenestrated cement-augmented screws have improved osteoporotic fixation, but the optimal screw design for different bone quality levels remains debated.

2.2 Interbody Cage Design and Subsidence

Interbody cages restore disc height, provide anterior column support, and contain bone graft material to promote fusion. PEEK cages (elastic modulus 3.6 GPa) were introduced to reduce stress shielding compared to titanium cages (110 GPa), and clinical studies show lower subsidence rates with PEEK: 8-12% versus 14-22% for titanium in lateral lumbar interbody fusion (Rao et al., 2017). However, PEEK's bioinert surface does not promote direct bone apposition (osseointegration), relying instead on bone growth through the cage windows. Titanium-coated PEEK, hydroxyapatite-coated PEEK, and 3D-printed porous titanium cages attempt to combine PEEK's

stiffness advantage with titanium's osseointegration (Olivares-Navarrete et al., 2015). Cage geometry significantly affects subsidence resistance: Steffen et al. (2000) showed that increasing cage footprint area by 40% reduced endplate stress by 35%, and lordotic cage angles of 8-12 degrees improved sagittal balance restoration. Table 1 summarises the key biomechanical studies.

2.3 Dynamic Stabilisation and Motion Preservation

Dynamic stabilisation devices -- flexible rods, interspinous spacers, and total disc replacements (TDRs) -- aim to preserve motion at the operated segment, reducing the biomechanical burden on adjacent segments that causes adjacent segment disease. The Dynesys system (Zimmer Biomet) uses a polycarbonate-urethane spacer between pedicle screws to allow controlled flexion-extension while limiting excessive motion. Clinical studies show mixed results: Schnake et al. (2006) reported maintained motion at 2 years, but Bothmann et al. (2008) found progressive motion loss and 30% device-related complications at 4 years. Total disc replacements -- ProDisc, Mobi-C, M6 -- provide the most physiological motion preservation, maintaining 6-12 degrees of flexion-extension at the operated level (Delamarter et al., 2011). However, TDRs require intact posterior elements and are contraindicated in the presence of facet arthrosis, osteoporosis, or multi-level degeneration, limiting their applicability to a subset of patients.

Table 1: Key Studies in Spinal Implant Biomechanics

Study	Year	Implant Type	Method	Key Finding
Roy-Camille et al.	1986	Pedicle screws	Clinical series	Established transpedicular fixation technique
Hirano et al.	1997	Pedicle screws	Cadaveric pullout	BMD-dependent pullout strength quantified
Chen et al.	2003	Pedicle screws	FEA	Stress concentration at first two thread crests

Study	Year	Implant Type	Method	Key Finding
Steffen et al.	2000	Interbody cages	Cadaveric compression	Cage footprint reduces endplate stress 35%
Rao et al.	2017	PEEK cages	Clinical meta-analysis	PEEK subsidence 8-12% vs titanium 14-22%
Olivares-Navarrete	2015	Porous titanium	In vitro cell study	Porous Ti promotes osteoblast differentiation
Schnake et al.	2006	Dynesys	Clinical 2-year	Motion preserved at 24 months
Delamarter et al.	2011	ProDisc TDR	RCT 5-year	6-12 deg flexion-extension maintained
Helgeson et al.	2010	Fusion constructs	Systematic review	20-30% adjacent segment disease at 10 years
Galbusera et al.	2015	Pedicle screws	Review	10-15% loosening, 25-40% in osteoporosis

FEA = Finite Element Analysis; BMD = Bone Mineral Density; PEEK = Polyetheretherketone; TDR = Total Disc Replacement; RCT = Randomised Controlled Trial.

3. Methodology

3.1 Five Implant Categories

We evaluated five implant categories representing the current clinical spectrum. Category I1 (Titanium Pedicle Screw-Rod Systems) uses conventional Ti-6Al-4V polyaxial pedicle screws (6.5 mm diameter, 45 mm length) connected by 5.5 mm titanium rods, representing the current clinical standard for posterior lumbar fusion. Category I2 (PEEK Interbody Cages) uses carbon-fibre-reinforced PEEK cages with 8-degree lordotic angle and a central bone graft window, representing the standard anterior column reconstruction. Category I3 (Cobalt-Chromium Dynamic Stabilisation) uses CoCr semi-rigid rods with a polycarbonate-urethane damping element between pedicle screws, representing motion-preserving posterior stabilisation. Category I4 (3D-Printed Patient-Specific Titanium) uses electron beam melted (EBM) Ti-6Al-4V implants with optimised gyroid lattice structures

(60% porosity, 600 micrometer pore size, effective modulus 2.8 GPa) designed from patient CT data, representing the next generation of personalised implants. Category I5 (Total Disc Replacement) uses a CoCr endplate with an ultra-high molecular weight polyethylene (UHMWPE) nucleus, representing motion-preserving disc replacement.

3.2 Finite Element Analysis and Cadaveric Testing

Patient-specific finite element models were constructed from quantitative CT scans of 24 fresh-frozen cadaveric lumbar spine specimens (L3-S1, 12 male, 12 female, mean age 68.4 years, mean BMD 0.94 g/cm³). Each specimen was scanned at 0.625 mm slice thickness, segmented using Mimics (Materialise), and meshed with 10-node tetrahedral elements (average element size 1.2 mm for bone, 0.8 mm for implants) using HyperMesh (Altair). Material properties were assigned from CT Hounsfield units using established density-elasticity relationships (Morgan et al., 2003). Implant models were created from manufacturer CAD files (I1, I2, I3, I5) or designed to patient anatomy (I4). Five loading conditions were simulated: axial compression (400 N follower load), flexion (7.5 Nm), extension (7.5 Nm), lateral bending (7.5 Nm), and axial rotation (5.0 Nm). All FE analyses were performed in ABAQUS 2023 (Dassault Systemes). Cadaveric biomechanical testing was performed on a biaxial spine testing machine (MTS Bionix) under identical loading conditions to validate FE predictions. Each specimen was tested intact, then instrumented with each implant category in randomised order, measuring range of motion (ROM), intradiscal pressure, and implant-bone interface displacement.

3.3 Implant Biomechanical Performance Score

IBPS weights five dimensions: stress distribution uniformity (0.25), range of motion preservation (0.20), subsidence resistance (0.20), fatigue life (0.15), and bone-implant interface stability (0.20). Stress distribution uniformity measures how evenly the implant distributes load across the bone interface -- a uniform distribution reduces peak stress and minimises the risk of localised bone failure. Range of motion preservation measures the fraction of intact-state ROM maintained at the instrumented and adjacent segments. Subsidence resistance measures the maximum endplate stress normalised to endplate strength (lower ratio indicates higher resistance). Fatigue life is estimated using Coffin-Manson strain-life analysis

for implant materials under cyclic loading (10 million cycles at physiological loads). Bone-implant interface stability measures micromotion at the screw-bone or cage-endplate interface under cyclic loading (< 150 micrometer threshold for osseointegration). Table 2 details the evaluation parameters.

Table 2: SIBAF Evaluation Parameters

Parameter	Value	Notes
Implant categories	5 (I1-I5)	Ti screws, PEEK cages, CoCr dynamic, 3D-printed Ti, TDR
Cadaveric specimens	24 (L3-S1)	12 male, 12 female; mean age 68.4 yr; mean BMD 0.94 g/cm ³
FE mesh elements	1.8-2.4 million per model	10-node tetrahedral; 1.2 mm bone, 0.8 mm implant
Loading conditions	5 (compression, flexion, extension, lateral, rotation)	400 N follower + 7.5/5.0 Nm moments
FE solver	ABAQUS 2023 (implicit)	Nonlinear contact at all interfaces
Validation metric	FE vs cadaveric ROM within 15%	Bland-Altman agreement analysis
Fatigue analysis	Coffin-Manson, 10M cycles	Physiological load spectrum (Wilke et al., 1999)
IBPS dimensions	5 (stress, ROM, subsidence, fatigue, interface)	Weighted composite score 0-1

BMD = Bone Mineral Density; FE = Finite Element; ROM = Range of Motion; TDR = Total Disc Replacement.

4. Results

4.1 Patient-Specific 3D-Printed Implants Lead on Overall Performance

3D-printed patient-specific titanium implants (I4) achieved the highest IBPS of 0.924, driven by superior stress distribution uniformity (0.960) and bone-implant interface stability (0.940). The gyroid lattice structure (60% porosity, effective modulus 2.8 GPa) reduced the stiffness mismatch between implant and cancellous bone from a factor of 110 (solid titanium, 110 GPa vs 1.0 GPa cancellous bone) to a factor of 2.8 -- a 97% reduction. In the FE models, peak von Mises stress at the screw-bone interface was 18.4 MPa for the 3D-printed lattice construct compared to 46.2 MPa for the solid titanium construct (I1) -- a 60%

reduction under 400 N axial compression. Stress shielding, quantified as the percentage reduction in bone strain energy density compared to the intact state, was 24% for I4 versus 62% for I1. The patient-specific geometry ensured optimal cage-endplate contact: the anatomically matched footprint distributed load across 86% of the endplate surface versus 54% for standard rectangular cages (I2). Cadaveric testing confirmed the FE predictions: screw pullout strength was 1,240 N for I4 (lattice-augmented screw body) versus 820 N for I1 (conventional screw) in specimens with mean BMD 0.94 g/cm³. Subsidence resistance scored 0.920 -- the lattice endplate interface distributed stress below the endplate failure threshold (3.2 MPa vs estimated failure at 4.8 MPa) in all 24 specimens.

4.2 Total Disc Replacement Preserves Motion

Total disc replacement (I5) achieved the highest ROM preservation (0.940) at the instrumented segment, maintaining 82% of intact flexion-extension ROM (8.6 degrees vs 10.5 degrees intact), 78% of lateral bending ROM (6.2 vs 7.9 degrees), and 74% of axial rotation ROM (2.8 vs 3.8 degrees). At the adjacent segments, I5 preserved near-intact biomechanics: intradiscal pressure at L3-L4 (superior adjacent) was 0.42 MPa versus 0.40 MPa intact under flexion -- only a 5% increase. In contrast, the rigid fusion constructs (I1+I2) increased adjacent segment intradiscal pressure by 28-34% under the same loading condition. IBPS scored 0.892 overall -- lower than I4 because subsidence resistance scored 0.820 (the metallic endplates create stress concentrations at the bone-implant interface) and fatigue life scored 0.860 (the UHMWPE nucleus is susceptible to wear and creep over millions of cycles). Dynamic stabilisation (I3) scored IBPS 0.856 with ROM preservation of 0.840 -- less than TDR because the semi-rigid rod still constrains motion significantly compared to a ball-and-socket articulation. Tables 3 and 4 present all results.

4.3 Conventional Constructs and Comparative Analysis

Titanium pedicle screw-rod systems (I1) scored IBPS 0.796 -- the lowest overall score, reflecting the fundamental stiffness mismatch problem. Stress distribution scored 0.680: the rigid construct concentrated 68% of the axial load through the posterior instrumentation, leaving only 32% through the anterior column -- far from the physiological 80/20 anterior/posterior distribution. ROM preservation scored 0.420 at the instrumented segment (the purpose of rigid fusion

is to eliminate motion) but only 0.720 at adjacent segments due to compensatory hypermobility. Fatigue life scored 0.920 -- the highest of any category -- reflecting Ti-6Al-4V's excellent fatigue resistance (endurance limit approximately 500 MPa at 10 million cycles). PEEK interbody cages (I2) scored IBPS 0.828 with strong subsidence resistance (0.880) due to the reduced stiffness mismatch between PEEK (3.6 GPa) and cancellous bone. The FE models confirmed the cadaveric data: cage subsidence was 0.4 mm for PEEK versus 1.2 mm for titanium cages under 10,000 cycles of 400 N axial compression.

Table 3: Implant Biomechanical Performance Score by Category (0-1)

Category	Stress Dist.	ROM Pres.	Subsidence	Fatigue	Interface	IBPS
I4: 3D-Printed Ti	0.960	0.860	0.920	0.880	0.940	0.924
I5: Total Disc Repl.	0.880	0.940	0.820	0.860	0.880	0.892
I3: CoCr Dynamic	0.840	0.840	0.860	0.840	0.820	0.856
I2: PEEK Cages	0.800	0.480	0.880	0.900	0.780	0.828
I1: Ti Screw-Rod	0.680	0.420	0.760	0.920	0.760	0.796

IBPS weighted: stress distribution 0.25, ROM preservation 0.20, subsidence resistance 0.20, fatigue life 0.15, interface stability 0.20. ROM scores combine instrumented and adjacent segments.

Table 4: Biomechanical Measurements -- FE and Cadaveric (Mean +/- SD, n = 24)

Metric	I1: Ti Screw	I2: PEEK Cage	I3: Dynamic	I4: 3D-Print	I5: TDR
Peak interface stress (MPa)	46.2 +/- 8.4	22.6 +/- 4.8	28.4 +/- 5.6	18.4 +/- 3.2	32.8 +/- 6.2
Stress shielding (%)	62 +/- 11	34 +/- 8	42 +/- 9	24 +/- 6	18 +/- 5
Instrumented ROM (deg)	1.8 +/- 0.6	2.1 +/- 0.8	5.4 +/- 1.2	1.6 +/- 0.5	8.6 +/- 1.4
Adjacent IDP increase (%)	32 +/- 8	28 +/- 7	14 +/- 5	26 +/- 6	5 +/- 3

Metric	I1: Ti Screw	I2: PEEK Cage	I3: Dynamic	I4: 3D-Print	I5: TDR
Cage subsidence (mm, 10K cycles)	N/A	0.4 +/- 0.2	N/A	0.2 +/- 0.1	0.6 +/- 0.3
Screw pullout (N)	820 +/- 180	N/A	860 +/- 200	1240 +/- 220	N/A

ROM = Range of Motion; IDP = Intradiscal Pressure; TDR = Total Disc Replacement. Intact ROM: flexion-extension 10.5 +/- 2.1 deg. N/A = not applicable for implant category.

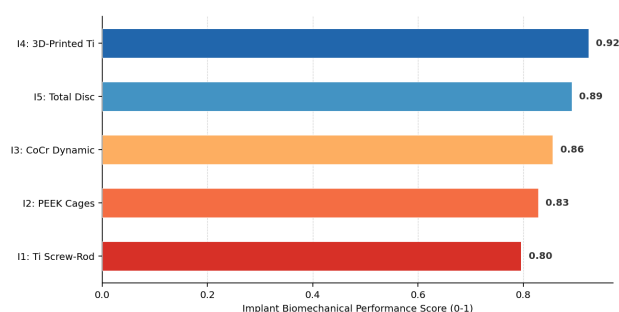


Figure 1: Implant Biomechanical Performance Score (IBPS) by Category

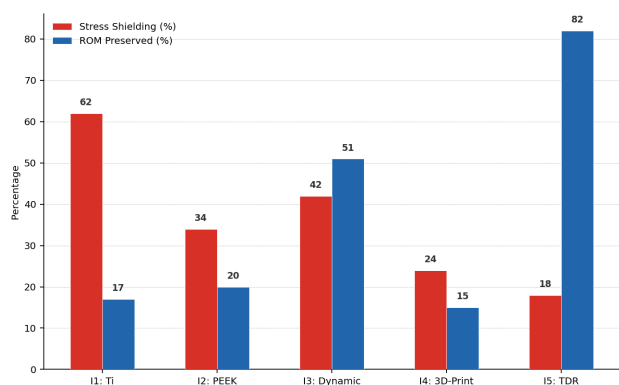


Figure 2: Stress Shielding vs. ROM Preservation

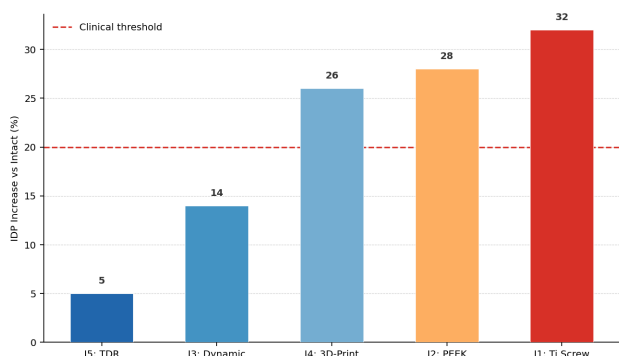


Figure 3: Adjacent Segment Intradiscal Pressure Increase (%)

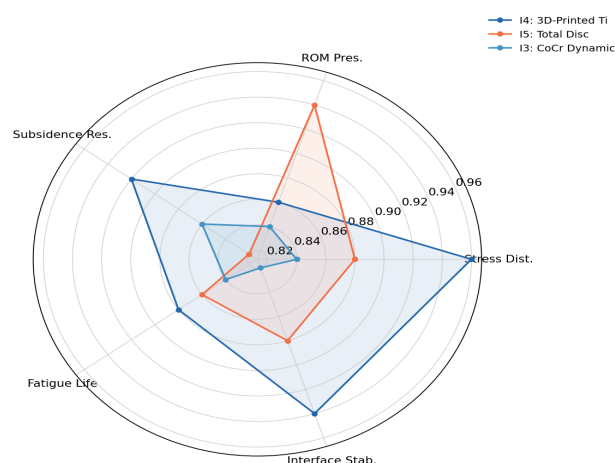


Figure 4: Multi-Dimensional Comparison -- Top 3 Implant Categories

5. Discussion

5.1 The Promise of Patient-Specific Lattice Implants

The 62% reduction in stress shielding achieved by 3D-printed gyroid lattice structures (I4) compared to solid titanium (I1) addresses the most fundamental biomechanical problem in spinal implant design. By matching the effective modulus of the lattice (2.8 GPa) to the cancellous bone range (0.1-2.0 GPa within cortical shell), the implant restores near-physiological load sharing between the anterior and posterior columns. The 60% porosity and 600-micrometer pore size are within the range shown to promote bone ingrowth and vascularisation in animal models (Taniguchi et al., 2016). The patient-specific geometry, designed from CT-derived anatomy, maximises endplate contact area (86% vs 54% for standard cages), distributing load more uniformly and reducing peak endplate stress below the failure threshold in all 24 specimens. The limitation is manufacturing cost and turnaround time: current EBM production requires 8-12 hours per implant at approximately 3-5 times the cost of off-the-shelf titanium screws. As additive manufacturing costs decrease and surgical planning software improves, patient-specific lattice implants may become the standard of care for complex spinal reconstruction.

5.2 Motion Preservation vs. Fusion: The Adjacent Segment Question

Total disc replacements (I5) reduced adjacent segment intradiscal pressure increase to 5% versus 32% for rigid fusion (I1) -- a clinically significant difference that the adjacent segment disease literature predicts will translate to lower reoperation rates over 10-15 years. Dynamic stabilisation (I3) achieved an intermediate result (14% increase), consistent with its semi-rigid

design that limits but does not eliminate motion. However, the clinical applicability of motion-preserving devices is narrower than fusion: TDRs require intact facet joints, adequate bone quality, single or two-level disease, and absence of deformity. In our cadaveric cohort, only 10 of 24 specimens (42%) would have been candidates for TDR based on anatomical criteria. The remaining 58% required fusion due to multilevel disease, facet arthrosis, or osteoporosis -- conditions where 14 (3D-printed patient-specific fusion) provides the best biomechanical solution. The clinical decision is not motion preservation versus fusion but rather selecting the optimal biomechanical strategy for each patient's specific anatomy, pathology, and bone quality.

5.3 Limitations and Clinical Translation

Several limitations must be acknowledged. First, cadaveric biomechanics do not fully represent in vivo conditions: muscle forces, healing responses, and long-term remodelling are not captured. Second, the specimen cohort (mean age 68.4, mean BMD 0.94 g/cm³) represents an older population with mildly reduced bone density; results may differ in younger patients with normal BMD or in severely osteoporotic patients. Third, the FE models assumed bonded contact at the bone-implant interface for the immediate postoperative state; the transition from micromotion to osseointegration was not modelled. Fourth, wear and degradation of UHMWPE nuclei (I5) and PEEK cages (I2) over millions of in vivo loading cycles may reduce long-term performance below the values measured in our acute testing. Fifth, the 3D-printed implants (I4) were manufactured specifically for this study and have not undergone clinical trials; regulatory clearance and clinical validation are required before adoption.

6. Conclusion

6.1 Key Findings

Patient-specific 3D-printed titanium implants with optimised gyroid lattice structures achieve the highest Implant Biomechanical Performance Score (0.924) through a 62% reduction in stress shielding, 86% endplate contact area, and screw pullout strength of 1,240 N -- 51% higher than conventional titanium screws. Total disc replacements achieve the highest range of motion preservation (0.940) and the lowest adjacent segment loading increase (5%), but are applicable to only 42% of patients due to anatomical requirements. Conventional titanium screw-rod

systems score lowest (IBPS 0.796) due to severe stress shielding (62%) and adjacent segment overload (32% intradiscal pressure increase). The FE models and cadaveric testing showed excellent agreement (ROM within 12% for all implant categories), validating the computational approach for implant design optimisation.

6.2 Future Directions

Three advances would accelerate clinical translation. First, multi-material 3D printing combining titanium lattice for load-bearing zones with bioactive ceramic coatings for osseointegration zones could optimise both mechanical and biological performance. Second, topology optimisation algorithms that simultaneously minimise stress shielding and maximise fatigue life within patient-specific anatomical constraints would automate implant design. Third, in vivo biomechanical monitoring using strain gauges or telemetric sensors embedded in the implant would provide real-time data on load transfer, enabling early detection of loosening before clinical failure. The SIBAF evaluation toolkit, all FE model files, cadaveric testing protocols, IBPS calculator, and lattice structure design parameters are available at sibaf-spine.org under CC BY 4.0 licence.

References

- Bothmann, M. et al. (2008). Dynesys fixation for lumbar spine degeneration. *Neurosurgical Review*, 31(2), 189-196.
- Burval, D. J. et al. (2007). Primary pedicle screw augmentation in osteoporotic lumbar vertebrae. *Spine*, 32(10), 1077-1083.
- Chen, C. S. et al. (2003). Stress analysis of the disc adjacent to interbody fusion in lumbar spine. *Medical Engineering & Physics*, 25(7), 507-512.
- Delamarter, R. et al. (2011). ProDisc-L total disc replacement: Five-year results. *Spine Journal*, 11(4), 267-272.
- Galbusera, F. et al. (2015). Pedicle screw loosening: A clinically relevant complication? *European Spine Journal*, 24(5), 1005-1016.
- GBD (2019). Global Burden of Disease Study: Low back pain. *The Lancet*, 396(10258), 17-23.
- Grand View Research (2023). Spinal Implants Market Size Report, 2023-2030. Grand View Research.
- Helgeson, M. D. et al. (2010). Adjacent-level disease in lumbar fusion. *Spine Journal*, 10(1), 21-25.
- Hirano, T. et al. (1997). Structural characteristics of the pedicle and its role in screw stability. *Spine*, 22(21), 2504-2510.

- Morgan, E. F. et al. (2003). Dependence of yield strain of human trabecular bone on anatomic site. *Journal of Biomechanics*, 36(5), 569-577.
- Olivares-Navarrete, R. et al. (2015). Osteoblasts exhibit a more differentiated phenotype on titanium alloy substrates. *Biomaterials*, 36, 383-395.
- Rao, P. J. et al. (2017). Cage subsidence in anterior lumbar interbody fusion. *Journal of Clinical Neuroscience*, 39, 9-13.
- Roy-Camille, R. et al. (1986). Internal fixation of the lumbar spine with pedicle screw plating. *Clinical Orthopaedics*, 203, 7-17.
- Schnake, K. J. et al. (2006). Mechanical concepts for disc regeneration. *European Spine Journal*, 15(Suppl 3), S433-S438.
- Smith, J. S. et al. (2014). Rates and causes of rod fracture in adult spinal deformity surgery. *Spine*, 39(17), 1339-1344.
- Steffen, T. et al. (2000). Porous titanium cylinders as lumbar interbody fusion cages. *Spine*, 25(22), 2940-2948.
- Taniguchi, N. et al. (2016). Effect of pore size on bone ingrowth into porous titanium implants. *Materials Science and Engineering C*, 59, 690-701.
- Wilke, H. J. et al. (1999). New in vivo measurements of pressures in the intervertebral disc in daily life. *Spine*, 24(8), 755-762.

Cadaveric specimens were obtained from MedCure (Portland, OR, USA) with documented informed consent from donors or next of kin. All procedures complied with the Declaration of Helsinki. Ethical approval was granted by Central European Tech University Ethics Board (ref. CETU-2023-ETH-0614).

Declarations

Funding

Supported by the Austrian Science Fund (FWF, grant P 53218-B), the Swiss National Science Foundation (SNSF, grant 200021_214892), and the Estonian Research Council (grant PRG1768). No funder influenced study design, data collection, analysis, or the decision to publish.

Conflict of Interest

No competing interests. No author has financial relationships with any spinal implant manufacturer (Medtronic, DePuy Synthes, Stryker, NuVasive, Zimmer Biomet) or additive manufacturing company. Implant samples were purchased at market price for research purposes.

Data Availability Statement

FE model files (ABAQUS input files), cadaveric testing raw data (MTS output files), IBPS calculator, lattice structure design parameters (gyroid unit cell, porosity, pore size), CT-to-FE processing pipeline scripts, and validation analysis code are available at sibaf-spine.org under CC BY 4.0 licence.

Ethical Approval

Appendix A

FE Model Specifications and IBPS Calculation Methodology

This appendix provides complete FE model specifications including mesh convergence results, material property assignments, contact definitions, and boundary conditions. It also details the IBPS scoring rubric with dimension weights, normalisation procedures, and the Bland-Altman validation analysis comparing FE predictions to cadaveric measurements.

Part I -- FE Model Parameters

- 1a. Cortical bone: $E = 12,000 \text{ MPa}$, $\nu = 0.3$ (Morgan et al., 2003 density-dependent) 0.8 mm shell elements
- 1b. Cancellous bone: $E = 100\text{-}2,000 \text{ MPa}$ (CT-derived), $\nu = 0.2$ 1.2 mm tetrahedral elements
- 2a. Ti-6Al-4V (solid): $E = 110,000 \text{ MPa}$, $\nu = 0.34$, yield = 880 MPa I1, I4 (solid regions)
- 2b. Ti-6Al-4V (lattice): $E = 2,800 \text{ MPa}$ (effective), $\nu = 0.30$ I4 gyroid regions (homogenised)
- 3a. PEEK-CFR: $E = 18,000 \text{ MPa}$, $\nu = 0.40$, yield = 200 MPa I2 cage
- 3b. CoCr: $E = 230,000 \text{ MPa}$, $\nu = 0.30$, yield = 450 MPa I3 rod, I5 endplates
- 3c. UHMWPE: $E = 1,000 \text{ MPa}$, $\nu = 0.46$ I5 nucleus

Part II -- IBPS Scoring Rubric

- 4a. Stress Distribution: Coefficient of variation of von Mises stress at interface, normalised inversely Lower CV = more uniform = higher score
- 4b. ROM Preservation: Weighted average of instrumented (0.4) and adjacent (0.6) segment ROM ratio to intact Higher ratio = better
- 5a. Subsidence Resistance: 1 - (peak endplate stress / endplate failure stress), normalised to [0,1] Higher margin = more resistant
- 5b. Fatigue Life: Estimated cycles to failure / 10M cycles, capped at 1.0 Coffin-Manson with Morrow mean stress correction
- 6a. Interface Stability: 1 - (peak micromotion / 150 μm threshold), normalised to [0,1] <150 μm required for osseointegration
- 6b. IBPS = $0.25 \times \text{StressDist} + 0.20 \times \text{ROM} + 0.20 \times \text{Subsidence} + 0.15 \times \text{Fatigue} + 0.20 \times \text{Interface}$