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Enhancing Surgical Outcomes in Pars Interarticularis Fracture: An Investigation of Treatment Modalities and Variability in Spondylolysis Surgical Approaches

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Abstract

Numerous studies have failed to definitively establish the most stable repair construct for spondylolysis. We seek to clarify this issue through a combination of physical experimentation using SawBones and corresponding finite element analyses (FEA). Using twenty-five SawBones lumbar vertebral body models, we tested four repair techniques: Buck's intralaminar screws, pedicle-screw and hook construct, pars interarticularis plating, and pedicle-screw intralaminar screw construct. Each Sawbones repair construct was secured with a C-clamp to the base of an MTS 858 Bionix test system. The testing frame was programmed to apply axial load to replicate an extension moment on the pars interarticularis and gradually increased until catastrophic failure, gapping of the pars defect exceeded 5 mm, or an obvious yield point was identified on stress-strain curves. The FEA involved creating an L4-L5 vertebrae model from CT scans, simulating fractures, and analyzing stress using SolidWorks and Ansys software. Results showed that while no repair method fully replicated the original stiffness of the intact pars, Buck's technique emerges as the closest approximation. Repair by pedicle screw hook, intralaminar screws, and plates demonstrates enhanced peak load to failure and elastic displacement. In FEA, the Buck technique exhibits superior reliability in comparison to other methods when assessing von Mises stresses, and the novel Implants 1 and 2 showcase the lowest average displacements. While the clinical significance of maintaining greater peak load or elastic displacement remains uncertain, our findings contribute to understanding spondylolysis repair. Additionally, utilizing FEA for novel designs offers promise in enhancing surgical outcomes.

Keywords: Spondylolysis; Pars fracture; Sawbones; Finite element analysis; Pedicle screw; Buck's

Abbreviations: FEA (Finite Element Analysis)

Introduction

Spondylolysis, an anatomical defect or fracture of the pars interarticularis of vertebrae (most commonly seen at L5 and L4), develops after birth and occurs in up to 6% of the population, with most patients being asymptomatic [1]. A much higher incidence is seen in young athletic patients participating in football, gymnastics, and weightlifting, with rates as high as 46% reported [1,2]. Symptomatic patients frequently respond well to conservative treatments, including activity modifications, physical therapy, lumbosacral orthosis bracing, anti-inflammatory medications, and avoidance of sporting activities.

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Surgical treatment is recommended for patients who remain symptomatic after 6 months to 1 year of nonsurgical management. Surgical treatment options for spondylolysis vary with patient age, functional status, instability, and presence of degenerative disc changes. Posterolateral or interbody fusion is the preferred treatment for patients greater than 30 years of age, with the presence of dysplastic lamina, degenerative disc changes on magnetic resonance imaging, and associated spondylolisthesis higher than Meyerding grade 2. However, direct repair of the pars defect can be successful in younger patients without degenerative discs or spondylolisthesis.

The first direct pars repair technique was described in 1968 by Kimura, which entailed bone grafting the defect without internal fixation, followed by prolonged bed rest and bracing [3]. In 1970, Buck introduced the intralaminar screw technique, which entails bone grafting the pars defect and then lagging the defect with 3.5-4.5 mm AO screws [4]. This technique has been widely studied clinically, with a high rate of excellent to good outcomes, but has been reported to be technically challenging [2] [5] [6]. Morscher modified this concept by placing a screw into the base of the superior articular process coupled to a hook that applies compression to the pars defect via the inferior laminar border, with excellent clinical results reported [7]. Gillet and Petit first described using a pedicle screw-rod construct, popularized by the ease of instrumentation and excellent clinical outcomes in several case series [8-10]. Additionally, a similar method involving the placement of pedicle screws and laminar hooks to compress the pars defect has been widely adopted and studied, which has yielded similarly good results. Songer and Rovin further augmented 11-15 pedical screw-based constructs by tensioning the spinous process with cables connected to pedicle screws that exert a compressive force across the pars defect, with a success rate of 71% reported [16].

Several biomechanical studies have attempted to determine which repair constructs have the most stable repair technique on calf cadaveric or fresh human cadaveric spines. Most compared Scott's wiring, Buck's screws, pedicle screw rod, and pedicle screw hook constructs. These studies have concluded that all repair techniques improve the mechanical stability of a spondylolysis model. Still, Scott's wiring is significantly weaker than Buck's screws and pedicle screw constructs [10] [12] [17-19]. However, biomechanical superiority has not been established between Buck's technique, the pedicle screw rod, and the pedicle screw hook methods. Some studies have suggested that Buck's may have higher fixation strength than screw-based constructs [20]. whereas others have suggested the contrary [12] [18]. Although none of these are statistically significant. Further, novel constructs have been tested by Patel et al. and Roberto et al. that consisted of a pedicle screw intralaminar screw and laminar plating that did not afford biomechanical superiority to conventional techniques, but direct comparisons have not been made [17] [19].

In contemporary biomechanical research, finite element analysis (FEA) is a prominent method for conducting comparative investigations. In its most basic form, FEA is a computational tool employed for predictive models of real-world stress-strain, including but not limited to mechanical forces, vibrational and thermal fluctuations, and other pertinent physical behaviors under specific boundary conditions. This analytical paradigm enables the assessment of a product's structural integrity, wear resistance, and fidelity.

This study's intended design purpose is two-fold. The first part was to compare, using a Sawbone model, commonly utilized spondylolysis repair techniques consisting of Buck's screws, pedicle screw-intralaminar hook construct, pedicle-screw intralaminar screws, and pars interarticularis plating. The secondary part of this study resides in computational simulation, where state-of-the-art surgical techniques intersect with a novel design paradigm conceptualized by the Biomechanics Research Laboratory for addressing fractures of the same nature. The overarching ambition here is to discern the potential enhancements and advantages offered by the innovative design relative to extant practices.

We predicted that the pedicle screw-based constructs would provide the greatest biomechanical strength compared to the others, as demonstrated in several previous studies. Additionally, we predicted that the findings from the proposed new constructs to tackle this type of fracture will signify potential avenues for future clinical use.

Materials and Methods

SawBone Experimentation

A total of 25 SawBones lumbar vertebral body models were used to test four repair techniques. Five vertebral body models were used for each repair technique, and a control group represented an intact pars interarticularis. The test constructs included Buck's intralaminar screws, pedicle-screw and hook construct, pars interarticularis plating, and pedicle-screw intralaminar screw construct (Figure 1). To spondylolysis in the Sawbones model, an osteotomy was performed at the pars interarticularis with a Stryker (Kalamazoo, MI) Precision Offset 9.0 x 0.51 x 25 mm blade.

Repairing the pars defect with Buck's screw technique involved predrilling the screw trajectories perpendicular to the osteotomy and placing 3.5 x 30 mm cortical screws (Synthes; Warsaw, IN) with a lag technique. The plating technique involved using 3.5 mm reconstruction plates (Synthes; Warsaw, IN) with locking screws. One screw was placed into the pedicle, and two screws were placed into the lamina with this construct. The pedicle screw and hook were

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Figure 1: Intact (a) and Fractured Sawbones with fixations for pars fractures: b) Bucks' Screw Model, c) Pedicle Screw Laminar Hook Model, d) Pars Interarticularis Plating Model

constructed by placing a 6.5 mm pedicle screw (Spinecraft LLC; Westmont, IL) attached to a laminar hook. The pedicle screw intralaminar screw construct consisted of 6.5 mm Expedium lumbar pedicle screws (DePuy; Warsaw, IN) and Mountaineer 3.5 mm x 10 mm cervical spine screws (DePuy; Warsaw, IN) linked by a 3.5-5.5 transition rod. All pedicle screws were placed at the intersection of the mid-transverse process line and the lateral border of the superior articular process. All set screws on the pedicle screw constructs were torqued to 80 in-lbs. using the standard torque handle.

Each Sawbones repair construct was secured with a C-clamp to the base of an MTS 858 Bionix test system (Minneapolis, MN) (Figure 2). The testing frame was programmed to apply axial load in such a manner as to replicate an extension moment on the pars interarticularis. This was gradually increased until catastrophic failure, gapping of the pars defect exceeded 5 mm, or an obvious yield point was identified on stress-strain curves. Catastrophic failure was defined as a fracture of the lamina or posterior elements, hardware breakage, hardware deforming, and hardware pullout.

Data measurements of peak load to yield point, modulus of elasticity, and percent strain at yield point. Data were recorded on an Excel spreadsheet (Microsoft; Redmond, WA). Independent samples t-test was used to make comparisons between repair techniques. ANOVA was used to detect statistically significant differences across all repair techniques. A significance level of 0.05 was used.

FEA Experimentation

For the FEA portion of the study, we created an L4-L5 vertebrae model using Mimics Medical 22.0, starting from the CT scans of a 68-year-old white female cadaver specimen. Once our model was developed in Mimics, we imported it into the SolidWorks 2019 Academic version (Dassault Systèmes, Concord, MA) for fracture creation and finally into Ansys 19.2 Academic Version (Canonsburg, PA) for the static analysis and simulation.

Mimics is an acronym denoting Materialise Interactive Medical Image Control System, an image processing software for 3D design and modeling developed by Materialise NV, Belgium. Starting from the 2D images, Mimics can create a 3D model to be processed and imported into other software for further analysis. Both L4-L5 assemblies were used to understand the area where the stresses concentrate the most. Still, we only used the L5 vertebra to perform all the surgical techniques and the crack since our purpose was to study spondylolysis, not the more severe spondylolisthesis.

The transformation from two-dimensional (2D) CT scans to a comprehensive 3D model necessitated several vital stages. The first task was to create a mask to differentiate between bone and the other tissue types. The threshold procedure set the value expressed in Hounsfield unit (HU), from which bone was differentiated from other structures. We set the range from 110 to 3071 HU. The adoption of this conservative lower HU threshold stemmed from the deliberate inclusion of adjacent structures, amenable to subsequent fine-tuning via the Multiple Slice Edit tool while ensuring the comprehensive capture of the region of interest. The subsequent phase involved the isolation of the L5 vertebra from the contiguous spinal structures. This was effectuated by applying the Split Mask tool, nested within the Segment section of the software. Then, the last part before importing everything in 3Matics was to smooth the model with a smooth factor of 0.7 and 3 iterations. These values were chosen mainly due to a trial-and-error process that tried to limit the smooth factor to avoid distorted or too-smoothed models. The same procedure was repeated for the L4 segment, hence obtaining the two levels in the exact position of the patient's CT scan. Because we took the images from CT scans, all the patient's ligaments and intervertebral disc could not be modeled using the Mimics mask. Hence, we had to create the ligaments and intervertebral discs in SolidWorks. The parts created were the following: intervertebral disc, supraspinous ligaments, interspinous ligaments, and intertransverse ligaments. Each component was modeled singularly, and the final result was assembled in one assembly module (Figure 3).

We made a small gap in the SolidWorks model to create the fractured version. We first analyzed the completed L4-L5 model to prove that the stress concentration was higher in the pars interarticularis region. The most important thing

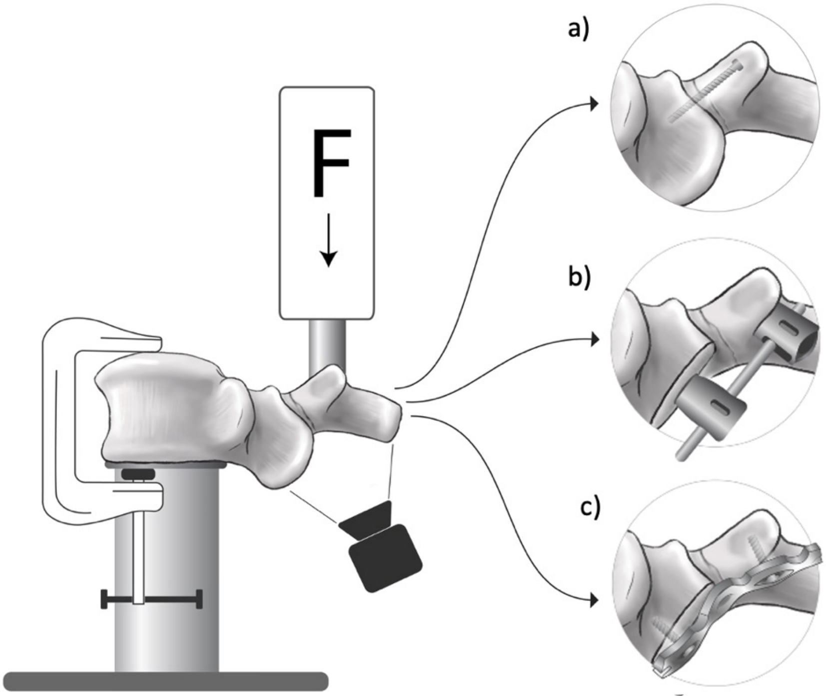

Figure 2: Diagram of SawBone Testing Frame. a) Bucks Screw Model, b) Pedicle Screw Laminar Hook Model, c) Pars Interarticularis Plating Model

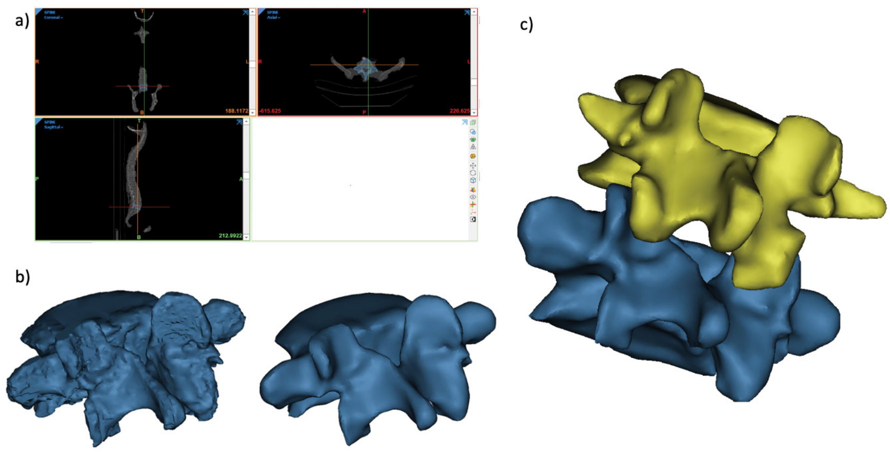

Figure 3: L4-L5 vertebrae modeling. a) Mimics view, b) L5 vertebra model in Mimics before and after smoothing and wrapping operation, c) L4-L5 final vertebra model

was to position the crack where we found the higher stresses in the neural arch and where all the literature indicates the fracture usually occurs. Given that the width of the fracture is not well defined in the literature and can range from 0.1 up to 1mm [22,23]. We decided to make the fracture 0.5mm. As for the depth, we made it through the whole vertebrae. The reasoning behind this is that the surgical operation of the pars, and hence the utilization of an implant, is done only when the pars is completely fractured (regardless of side), not when there is a stress fracture barely visible on the plane radiographs.

The final step was to make simulations on the intact and fractured L5 vertebra only and see the changes in stress distribution. In the Static Structural module of Ansys Workbench, we assigned boundary conditions such that both the upper and lower surfaces of the vertebral body were fixed with no rotation or displacement allowed—this simulated fixation with a clamp that can be easily replicated in a laboratory with appropriate machines. As for the load, we applied a weight of 500N to the region that would be pressed during a possible experiment. The stress and deformation values were measured for later comparison.

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Next, we chose to take two of the constructs that yielded high construct stiffness when previously tested on SawBones (Buck's and Pedicle Screw Intralaminar Hook) and modeled them using FEA to validate our findings further (Figures 4 & 5).

We also tested and compared these to 2 novel designs to look for a potential alternative to the current repair techniques (Figure 6). For all the methods tested (Buck, Hook-Screw, 2 novel), the tool Cavity was used to combine the vertebra and the screw once it was positioned correctly. This feature in SolidWorks allowed us to create a hole that perfectly resembled the object's shape inside the main one selected (i.e., in this case, the cavity will have the same shape as the screw). Additionally, since surgeons usually close the gap of the fracture once the screw is in place, we reduced the fracture size from 0.5 mm to 0.1 mm to effectively mimic the gap closure procedure.

Lastly, the models were imported into Ansys Workbench, where the exact boundary and load conditions were applied at the same faces of the vertebra. The techniques' subsequent stresses (maximum and minimum) and the displacement (maximum, minimum, and average) were measured and compared.

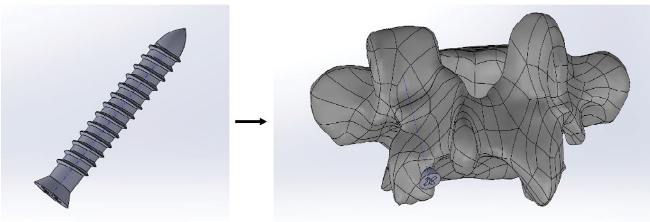

Figure 4: FEA Models of Buck's Screw and insertion into fractured vertebral model

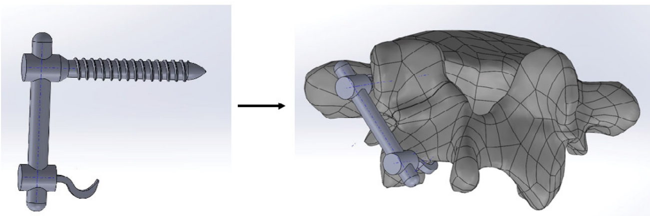

Figure 5: FEA Models of Hook-Screw and insertion into fractured vertebral model

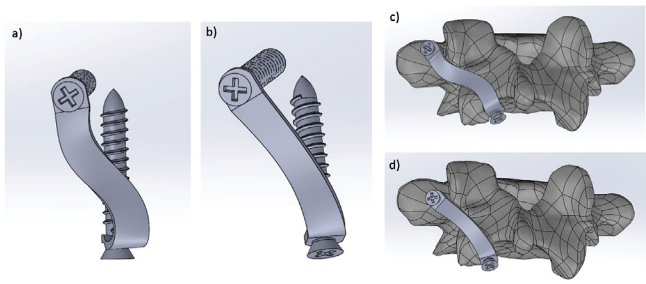

Figure 6: FEA models of the novel design paradigms conceptualized by the Biomechanical Research Laboratory. a&c) Implant 1, b&d) Implant 2

Concerning the design of the Buck technique in the SolidWorks environment, we took the fractured vertebra model of 0.5mm fracture as the starting point of our assembly. We reduced the fracture distance, as discussed above. The new vertebra was coupled with a screw to replicate the Buck screw. The length and width of the screw (30mm and 4mm, respectively) were chosen as they correspond to the standard length used in the Buck technique.

Concerning the design of the Hook-Screw technique, we first designed the Hook-Screw system to fit perfectly on the vertebra. The rod connecting the screw and the hook had a diameter of 4.5 mm and a length of 41 mm. The screw had a length of 31 mm and a diameter of 4 mm. The screw length was made to pass through the entire pedicle to guarantee the best support and attachment.

The novel design studied in this paper was conceived and proposed in the UIC Biomechanics Research Laboratory by Professor Farid Amirouche. The main idea was to keep the two entry points of the previous methods, the lower point of entry from the Buck technique and the upper entry point from the Hook-Screw technique, and a plate was used to connect them. The first plate was designed to be more curved and follow the geometry and shape of the vertebra to avoid contact with many surrounding muscles or tendons. The second plate had a similar shape to the previous one, but the curvature was reduced. In the SolidWorks environment, to connect the two screws, a rectangular section of width 1.5 mm was assigned, and then with the Feature Boundary Boss, we were able to create a solid from the contours we had defined. The only holes in the plate were made in the points mentioned above since the curvature of the plate won't allow other holes to contribute to the stability of the plate.

Results

Load-to-failure testing of the Sawbones with an intact pars articularis demonstrated a peak load to failure of 200.322 N (95% CI; 1 45.367 – 255.277 N). The load to failure for models with a pars defect repaired with Buck's Screw technique was 276.128 N (95% CI; 184.928 – 367.328 N), which was not statistically different from the intact pars ($p = 0.212$). Repair constructs with pedicle screw-hook and pedicle screw-intralaminar screws demonstrated peak loads to failure of 514.891 N (95% CI; 441.386 – 588.396 N) and 522.915 N (95% CI; 505.820 – 540.010), respectively, with no statistical difference between these constructs ($p = 0.842$). Using laminar plates to stabilize the pars defect demonstrated a peak load to failure of 554.750 N (95% CI; 501.503 – 607.997 N). There was no statistically significant difference in the peak load of the plate, pedicle screws, intralaminar

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screw, and pedicle screw hook constructs (p = 0.569). There was a statistically significant difference in peak load strength between all constructs, with the pedicle screw and plate constructs sustaining the highest loads (p < 0.0001). (Figure 7)

Elastic displacement of the intact pars Sawbones models was 0.61 mm (95% CI; 0.36 – 0.86 mm). This was less than Buck's screw construct, measuring 1.26 mm (95% CI; 0.59 – 1.91 mm) but was insignificant (p = 0.118). The pedicle screw intralaminar screw construct underwent the most critical displacement while maintaining elastic properties measuring at 5.34 mm (95% CI; 2.93 – 7.77 mm) but was not significantly different than laminar plating constructs measured at 4.41 mm (95% C I; 2.71 – 6.12 mm) (p = 0560). Pedicle screw hook constructs underwent 2.60 mm of elastic displacement (95% CI; 1.52 – 3.66 mm) but were insignificant compared to the plating and pedicle screw intralaminar screw constructs (p = 0.128 and p = 0.881, respectively). Statistically, there was a substantial difference in elastic measurements across all groups (p = 0.002). (Figure 8)

Construct stiffness was measured as the slope of the elastic region of the load versus displacement curve, with a calculated stiffness of 151.00 N/mm (95% CI; 120.30 – 168.80 N/mm) for the intact pars. This was greater than the stiffness of Buck's screw repair construct, which was measured to be 97.67 N/mm (95% CI; 74.00– 137.00N/mm), but this was not significant (p = 0.195). There was no significant difference in construct stiffness between the pedicle screw intralaminar screw, pedicle screw hook, and plate constructs (p = 0.355), with measurements being 71.34 N/mm (95% CI; 53.21 – 89.47 N/mm), 67.12 N/mm (95% CI; 45.76 – 88.48 N/mm), and 59.44 N/mm (95% CI; 41.01 – 77.87 N/mm) respectively. Statistical analysis with one-way ANOVA demonstrated a significant difference in construct stiffness between all constructs (p = 0.0005). (Figure 9)

Table 1 compares the FEA values for equivalent stresses (maximum and minimum, i.e., von Mises) and Table 2 compares the displacement (maximum, minimum, and average) of the intact vertebra, the new designs, and the fracture.

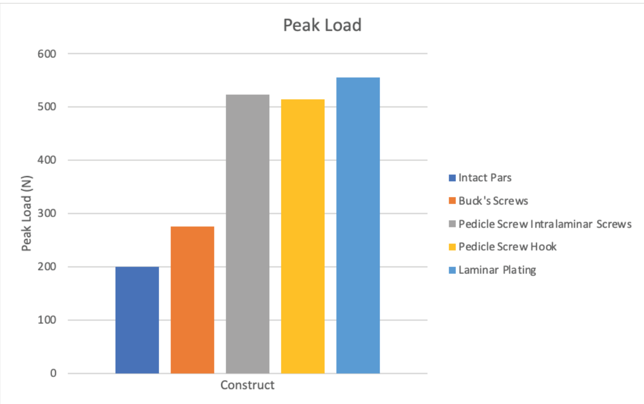

Figure 7: Load-to-Failure Testing Results

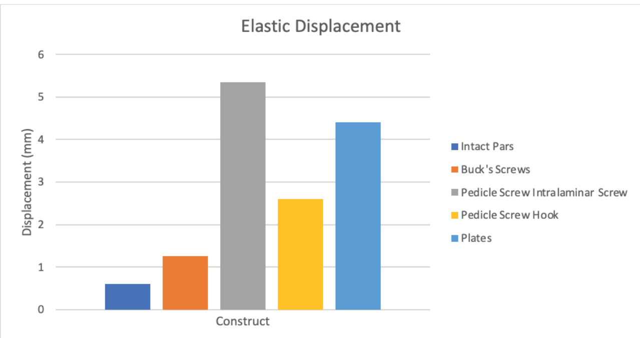

Figure 8: Elastic Displacement Results

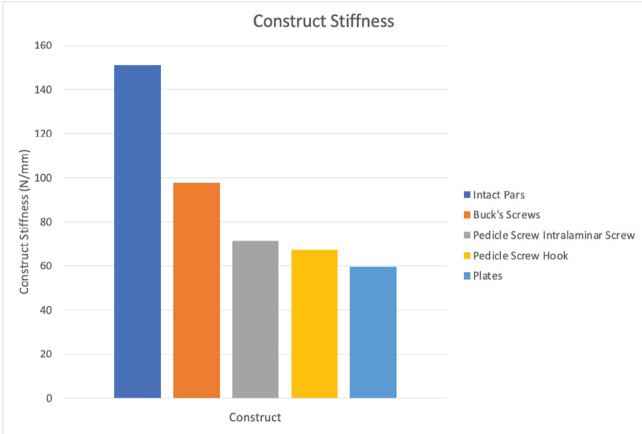

Figure 9: Construct Stiffness Measurements

Table 1: Comparison of maximum and minimum stresses (expressed in megapascals [MPa]) experienced by the vertebra under different conditions

Condition	Maximum [MPa]	Minimum [MPa]
Intact	35.71	2.32 e-002
Buck	51.14	1.81 e-002
Implant 1	69.49	3.32 e-002
Fractured	78.42	2.7 e-005
Hook-Screw	80.94	1.02 e-002
Implant 2	116.57	1.03 e-002

Table 2: Maximum, minimum, and average deformations in millimeters (mm) for various conditions

Condition	Maximum [mm]	Minimum [mm]	Average [mm]
Intact	0.24	0	3.01 e-002
Implant 1	0.46	0	5.07 e-002
Implant 2	0.45	0	5.33 e-002
Buck	0.45	0	6.32 e-002
Hook Screw	0.62	0	5.08 e-002
Fractured	1.01	0	0.14

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Discussion

Repair of the pars interarticularis allowed for increased construct elasticity but did not quite recreate the stiffness of the intact pars Sawbone model. This corroborates with previous studies that have examined the stability and characterized motion of the pars defect after commonly used repair techniques. In a study that created bilateral pars defect in calf cadaveric L2-6 vertebral bodies and repaired with either cable plates, pedicle screw hook, rod constructs, or a pars plate, the intact pars maintained greater stability in mean aggregate rotation [19]. However, compared to the flexion and extension stability, the primary endpoint was the amount of displacement rather than the calculated construct stiffness, which they found no significant difference between the three tested repair constructs. We reported construct stiffness as a function of the slope of the elastic region of the load versus displacement curve. In comparison, the Roberto et al. plate construct demonstrated significantly increased lateral bending stiffness of the plate construct; we found no significant difference in construct stiffness between the plate, pedicle screw hook, and pedicle screw intralaminar screw constructs but with a higher peak load to failure, albeit not significantly. Interestingly, the pedicle screws intralaminar and plate constructs allowed for the increased elastic displacement of 4.412 mm, measured as the displacement at the pars defect while maintaining elastic properties. This finding was significant compared to the pedicle screw hook construct, which has a well-proven track record in multiple clinical and biomechanical studies [11] [12] [21]. Stiffness measurements of the plate constructs could have been affected by the lack of screw fixation points on both sides of the defect. In our study, there was one screw followed by two screws, which could have been attributed to the deviation of our findings from those reported by Roberto et al., as the failure mechanism appeared to come from the plate-screw interface. Future studies should include plates with anatomic designs that maximize screw placement on both sides of the pars defect.

Utilization of pedicle screw intralaminar screw constructs has seldom been reported in the literature, with only one study describing its use [17]. In that study, Patel et al. analyzed CT scans to assess the anatomy of L4 and L5 laminae in patients with spondylolysis and controls to accept intralaminar screws, which were inserted much like Buck's technique. They determined that anatomically the laminae would accommodate the 4.5 mm by 2.5 mm screws for this fixation technique. Further, a biomechanical comparison of the pedicle screw intralaminar screw to the more conventional pedicle screw hook construct yielded no significant difference in axial, rotational, and bending motion, consistent with our findings. However, our technique of the pedicle screw intralaminar screw method was slightly different than that described by Patel et al., where we used 6 x 40 mm pedicle screws connected to two 3.5 mm x 10 mm screws inserted on each side of the laminae and connected with a 3.5-5.5 mm transition rod. Despite these design differences, both pedicle screw intralaminar screw constructs performed similarly between our study and Patel et al.

This is the first study to compare the pedicle screw intralaminar screw construct to other well-established repair techniques biomechanically. This relatively novel construct proved to have a very high peak load to failure, 522.915 N, and tolerate a large amount of elastic displacement, 5.345 mm. It appears to have performed comparatively to the pedicle screw hook construct, demonstrating a 514.891 N peak load and 2.598 mm of elastic displacement. It has a good track record in the literature. It is currently unclear if these reported metrics are of significance. However, in theory, constructs with greater elasticity can tolerate more significant amounts of minor displacement forces and still maintain the reduction of the pars defect. This could play an essential role during the early phases of the healing process when early weight bearing begins.

In both the SawBone and FEA analyses, Buck's screw technique most closely stabilized the pars defect to the intact stiffness measurements. A combined clinical and biomechanical study by Fan et al. compared Buck's technique with the Texas Scottish Rite Hook, Scott's wiring technique, and pedicle screw U-link and found no significant difference between Buck's screw technique to the pedicle screw hook and U-link constructs with respect to degrees of motion in flexion-extension, lateral flexion, and rotation [12]. They then reported on 11 patients that were repaired using the TSRH construct with 10/11 healing on CT scans with good clinical reported outcomes in patients with radiographic nonunion. Another biomechanical study by Mihara et al. recorded the change in motion across the L3-4 and L4-5 motion segments after creating and repairing a pars defect at L4 in nine calf lumbar spine models [20]. The Buck's technique appeared to establish spinal motion and stability across both segments to the intact pars compared to the pedicle screw hook construct. Although they reported metrics different than our study, percent range of motion rather than construct stiffness, their findings are substantiated by the current study that Buck's technique most closely recreates construct stiffness to that of the intact pars. The reasoning for this is still unclear but may stem from the lag effect of the repair construct generating more cortical contact between defects, which confers greater inherent stability, rather than the other tested constructs that approximate the two fragments more or less.

Per Table 1, Buck's technique and Implant 1 had a maximum von Mises stress value lower than the fractured vertebra. The Hook-Screw technique was comparable to the fractured vertebra, while Implant 2 reached a maximum

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value that is considerably higher in comparison to the Buck technique and Implant 1. Therefore, from a pure von Mises stress point of view, the Buck technique seems the most reliable compared to the other methods, even if Implant 1 presents similar results. Regarding the displacement analysis, Implants 1 and 2 had the lowest average displacements. The new implant designs improved the deformation distributions overall, decreasing the deformations by 15% for Implant 2 and slightly more for Implant 1 concerning the Buck technique. Given the results of our FEA approach, Implant 1's design could be a promising avenue for pars interarticularis fracture repair. One potential next step for this project would be to 3D print and test the implants on sawbones to validate our results. The designs themselves could be changed and improved even though the plate's curvature is the main changing parameter since the insertion points are fixed.

Limitations

There were several limitations in the current study. SawBone models were used rather than cadaveric specimens. We believed this would permit a more uniform testing model than cadaveric specimens. However, these measurements may not fully reflect the proper stability of these repair constructs as they lack the muscle and ligamentous structures often included in biomechanical tests. 3.5 mm reconstruction plates were used when evaluating the plating constructs, which may have affected our results, which are not designed for spinal lamina. Mainly, due to the anatomy of the pars and lamina and the geometry of the reconstruction plates, obtaining at least two points of screw fixation on either side of the defect was sometimes impossible. This may have led to decreased construct stiffness. Future studies should evaluate anatomically designed plates that can accommodate more screw fixation points. Lastly, the ligaments in FEA were treated as isotropic elastic parts for simplicity, even though they do not behave in an elastic way but viscoelastic.

Conclusion

The intricacies associated with pars interarticularis fractures, characterized by their formidable difficulty and minute scale, contribute to the persisting lack of consensus regarding the optimal approach. The diverse array of plates and screws available further exacerbates the variability in surgeon approaches, precluding the establishment of a definitive standard. The absence of a discernible superiority among existing methodologies underscores the imperative for continued exploration into minimally invasive modalities. Such endeavors are envisioned to yield advancements capable of ameliorating postoperative discomfort and reducing hospitalization duration after spondylolysis surgery.

In the current two-fold study, repairing the pars defect did not fully recreate the inherent stiffness of the intact pars. Still, while we originally predicted the pedicle screw constructs would provide the greatest biomechanical strength given previous literature, Buck's technique appeared to approximate this most closely. Repairing with pedicle screw hook, intralaminar screws, and plates enabled a more significant peak load to failure and elastic displacement than the intact pars and Buck's technique. Although it is unclear whether it is clinically essential to maintain a more significant peak load to failure or elastic displacement, this study is relevant for better understanding spondylolysis repair. It may also be a metric of interest given the good track record of the pedicle screw hook construct in the literature. The use of FEA for novel designs, such as the two implants discussed in this paper, also serves as a promising approach to pars fracture repairment, potentially improving surgical outcomes. This nuanced exploration is a foundational step in advancing the discourse surrounding spondylolysis repair, offering prospects for meaningful innovation in orthopaedic implant design and surgical approaches.

Conflicts of interest: None to disclose

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