

Research Paper

The effect of onlay cortical fibula strut grafts on biomechanical features of Vancouver type B1 periprosthetic femoral fractures

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ABSTRACT

Objective: This study aimed to investigate biomechanically the effects of onlay fibula grafts on Vancouver Type B1 Periprosthetic Femoral Fractures (PPFs).

Methods: Vancouver Type B1 PFF models were created in 25 fourth-generation synthetic femurs and fixed with locking plates using bicortical, unicortical screws, and cables. While no graft was used in group 1, onlay fibula grafts were placed anteriorly in group 2 and medially in group 3. In group 4, the cortical strut allograft was placed on the medial femoral cortex, and a locking compression plate (LCP) was applied to the lateral femoral cortex. In group 5, the strut allograft was placed over the anterior cortex of the femur and fixed with the same technique as in group 4. All models were then subjected to rotational and axial cyclical stiffness tests and load to failure to measure and compare the mechanical strengths of the constructs.

Results: The mean stiffness values of group 4 with medial allograft, before and after cyclical loading, were higher than all other groups, under both rotational and axial forces. The mean stiffness values of fibula autografts (groups 2 and 3) were similar to that of anterior allografts (group 5) in each test except that the mean initial axial stiffness of group 5 was higher than group 2. Failure loads were also not different between the groups.

Conclusion: Although the rigidity of Vancouver type B1 periprosthetic femur fractures is highest if allografts are placed medially, fibula autografts can also provide similar fixation strengths to allografts if locking plates with unicortical and bicortical screws and cables are used.

Introduction

In type B1 periprosthetic femur fractures (PFF) after hip arthroplasty, the treatment goal is successful union. This is often achievable with open reduction and internal fixation, although some authors recommend revision of the fixed femoral stem in such cases.¹ Although cortical strut allografts in revision total hip arthroplasty and PFF have reported success rates of 80%-100%,^{2,4} there is some concern regarding the use of allografts for PFF fixation, such as the transmission of prions, viruses and bacteria,⁵⁻⁸ their high cost, and the difficulty of obtaining suitable allografts in sufficient quantities in some regions.⁹ As an alternative, we considered that a cortical fibular autograft with appropriate mechanical and biological characteristics could solve this problem. Vascularized and non-vascularized fibula autografts have been used successfully for extremity reconstruction in orthopedics for the past 100 years.¹⁰ In addition, a fibula autograft has an important advantage over a cortical onlay allograft, in that it is fresh and has osteoinductive, osteoconductive, and osteogenic properties for graft incorporation and fracture union.¹¹

There is no published information regarding the use of onlay fibula strut autografts instead of cortical strut allografts in the treatment of Vancouver type B1 PFF, and these autografts have not been biomechanically compared. We investigated the contribution of fibula grafts to the mechanical strength of reconstructions when used in combination with locking compression plates (LCP) in Vancouver type B1 PFF models. We examined whether the initial and final torsional stiffness values, initial and final axial stiffness values, and failure loads of onlay fibula grafts are comparable to onlay allografts and those without structural grafts.

Materials and Methods

As test material, 25 fourth-generation synthetic, polyurethane, identical, left, large femur models (Sawbones 3406; Sawbones Europe, Malmö, Sweden) were used.^{12,13} They were divided into 5 groups of 5 specimens each: group 1, control group with no graft support, only lateral locking plate with screws and cables; group 2, anterior fibula and lateral plate; group 3, medial fibula and lateral plate; group 4, medial cortical strut and lateral plate; and group 5, anterior

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cortical strut and lateral plate. The tests for groups 1, 4, and 5 had been reported previously by the authors of the current study.¹²

All models underwent neck osteotomies with an angle of 45° relative to the shaft and 1.5 cm proximal to the lesser trochanter. After reaming and broaching the medullae of the models for a conical femoral stem (size 16 Synergy; Smith & Nephew, Memphis, Tenn, USA), the prostheses were inserted by press-fitting. A 4.5-mm, 12-hole, broad titanium LCP (Synthes, Paoli, Pa, USA) was placed on the lateral cortex and fixed to each model with 38- and 40-mm bicortical locking screws (Synthes) distally and 2 14-mm unicortical locking screws proximally.¹² A 1-cm circumferential segment was then removed from the femoral diaphysis with a low-profile electric saw at the level of the prosthesis tip to create a fragmented Vancouver type B1 fracture model.¹²

For the group 1 models, the plate was fixed to the bone with 2 additional 1.7-mm stainless-steel cables (TasarımMed, Istanbul, Turkey) through cerclage eyes (Synthes) from each side of the fracture level.

For group 2, 5 fourth-generation composite fibulae (Sawbones 3406; Sawbones Europe, Malmö, Sweden) were used to simulate cortical strut fibula autografting. For this reason, as in clinical practice, 20-cm diaphyseal parts of total fibula models were first bisected, and then each part was divided longitudinally to obtain 4 cortices, each 7.5 cm in length. Fibula struts were placed side to side over the anterior cortex of the model so that the defect is in the middle of the graft. The graft models were fixed to the femoral cortex and to the plate using 2 1.7-mm stainless-steel cables and cerclage eyes proximal and distal to the defect site (Figure 1A).

In group 3, the same procedures were applied as in group 2, but the fibula grafts were placed over the medial cortex of the femur, side to side (Figure 1B).

In group 4, the use of cortical strut allograft on the medial femoral cortex and LCP on the lateral cortex in the treatment of Vancouver type B1 PFF was modelled.¹² Cortical struts were prepared from fourth-generation composite femur models (Sawbones 3406; Sawbones Europe) by cutting their diaphyseal parts to 15 cm in length and 2.5 cm in width. One strut was placed over the medial femoral cortex, centering the defect level, and fixed with 2 cables proximally and two cables distally.

For group 5, the strut allograft was placed over the anterior cortex of the femur and fixed with the same technique as in group 4.

We obtained anteroposterior and lateral radiographs for all groups (Figure 2A-E).

HIGHLIGHTS

- There is a paucity of data regarding the use of onlay fibula strut autografts in the treatment of Vancouver type B1 periprosthetic femoral fractures (PFF), and these autografts have not been biomechanically compared. This study aimed to evaluate the contribution of fibular grafts to the mechanical strength of reconstructions when used in combination with locking compression plates (LCP) in Vancouver type B1 PFF models.
- The mean stiffness values of the with medial allograft (group 4), before and after cyclical loading, were higher than all other groups under both rotational and axial forces. The load to failure were not found different between the groups.
- The results suggest that the fixation rigidity of Vancouver type B1 periprosthetic fractures is highest if structural cortical allografts are placed medially and that Vancouver type B1 PFF without large segmental defects, onlay fibula strut autografts can be used as an alternative to allografts.



Figure 1. A, B. Four pieces of grafts with 7.5 cm each prepared from fibula and (A) fixed on the anterior cortex in group 2 and (B) medial cortex in group 3, using four cables.

Two 2-mm-diameter Kirschner wires were inserted into the distal femoral condyles of the models crossing at right angles. Each model was embedded in a 110-mm-diameter, 60-mm-high polyvinyl chloride (PVC) pipe, from the distal end using polyester putty so that the long axis of the stem was perpendicular to the ground. Thereafter, all models in this biomechanical experiment were tested for stiffness under cyclic rotational and axial loading and axial failure loading using a servohydraulic testing device (MTS 858 Mini Bionix II; Eden, Prairie, Minn, USA). The loading protocol was from our published biomechanical study using the same set-up.¹²

The models were fixed to the lower jaws of the device from PVC pipes vertically so that the femoral stems were rotated around their longitudinal axis through an apparatus fixed to the stems proximally to minimize bending moments in the specimens during the rotation of the stems (Figure 3A-E).^{12,13} The moments were measured and recorded using the load cell with a 25-kN axial and 200-Nm torsional capacity. After baseline stability control with a 15 N compressive force (speed, 0.2 mm/s), the system was unloaded under moment control until the moment reached 0.5 Nm. Thereafter, 10 repeated cyclical twists at a frequency of 3 Hz between 0.5 and 10 Nm were performed, and the system was released until the moment value reached 0.5 Nm. The models were loaded with the same torque and frequency for the following 1000 cycles. The test continued for 10 000 cycles. After the first 10 cycles, and every 1000 cycles thereafter, the stiffness of the models was measured and recorded.

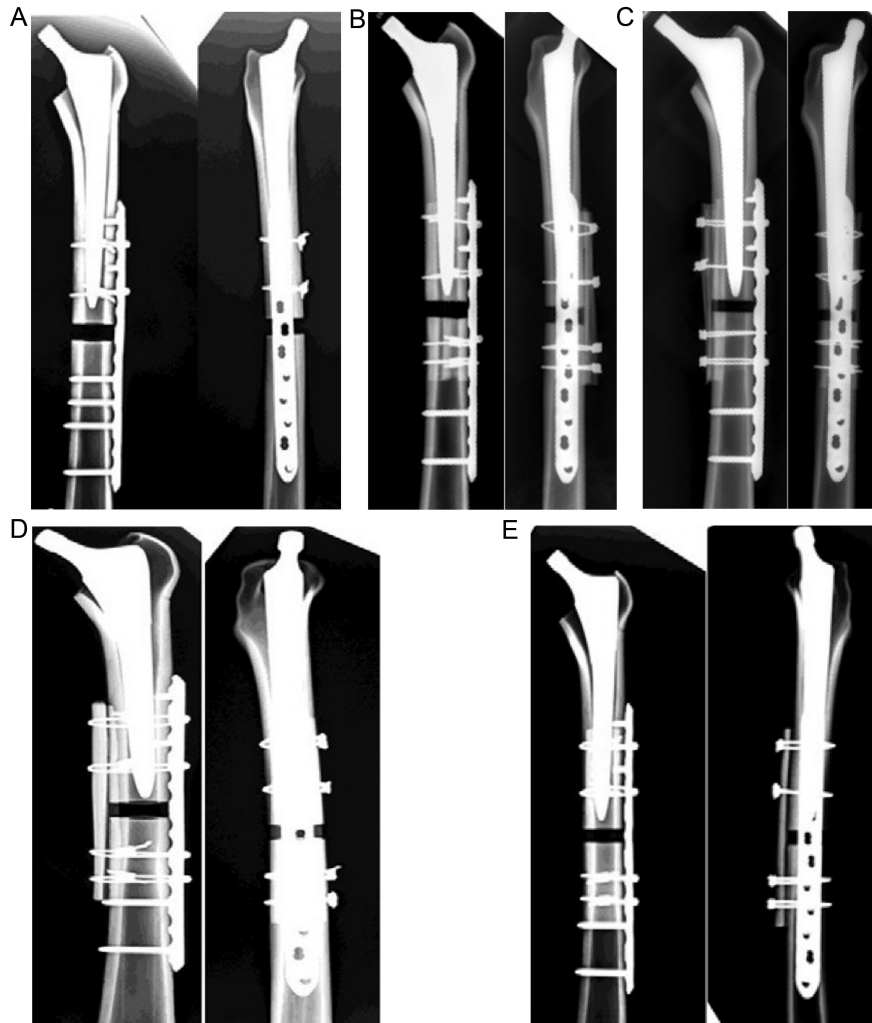


Figure 2. A-E. Antero-posterior and lateral radiographs of the specimens from (A) group 1 without graft, (B) group 2 with anterior fibula graft, (C) group 3 with medial fibula graft, (D) group 4 with medial strut allograft and (E) group 5 with anterior strut allograft.

The cyclic axial loading test was performed after the end of rotational loading. The models were fixed to the lower jaws of the test machine at 15° of valgus to simulate single-leg stance phase of the gait.^{12,13} The force was transmitted to the femoral stem through a specially prepared plate similar to an acetabular liner, which was placed in the upper hydraulic grip of the machine, and a +0 CoCr femoral

head (Smith & Nephew) (Figure 4A-E). First, a compressive force of 50 N (speed, 0.2 mm/s) was applied as a preload to stabilize the construction. Then, force control was launched and 10-cycle axial loading was performed at 3-Hz between 50 and 500 N. The system was released until it reached the lower limit of 50 N, and thereafter, the model was loaded until 500 N with a velocity of 50 N/s and then

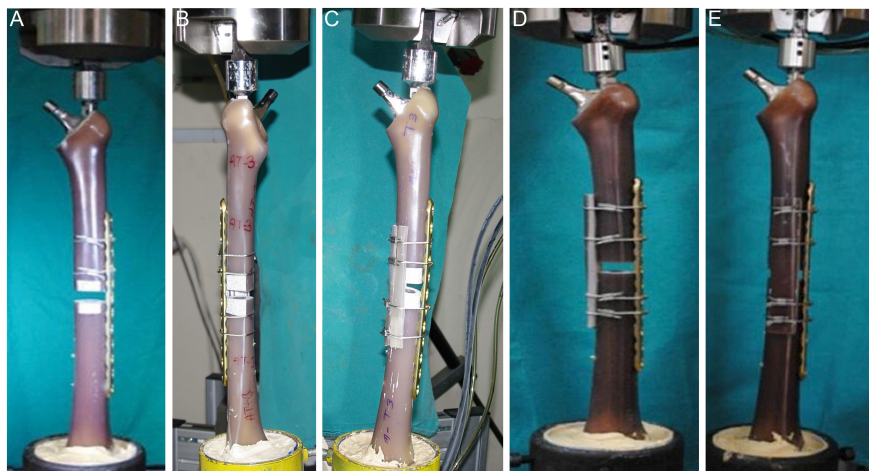


Figure 3. A-E. Rotational loading tests of (A-E) group 1 to group 5.

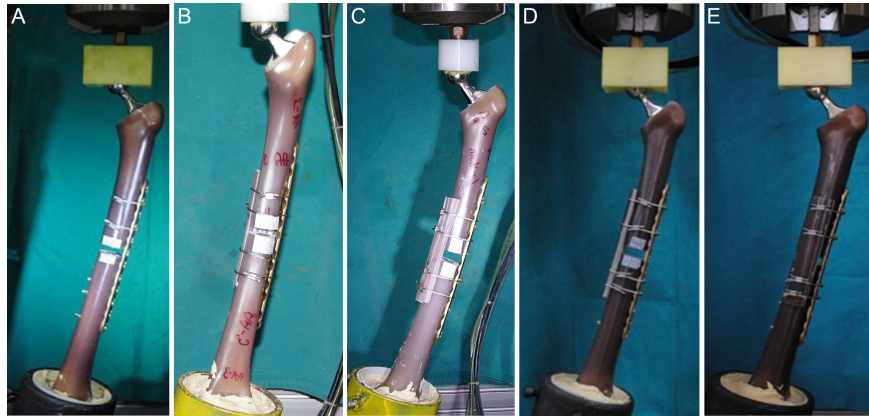


Figure 4. A-E. Axial loading tests of (A-E) the groups, respectively.

it was unloaded again. At this stage, the initial stiffness value of the model was recorded. At the second stage, the model was reloaded axially with 50-500 N at 3 Hz for 10 000 cycles and was unloaded until it reached 50 N at the end of each 1000th cycle. At the end of this stage, the final stiffness value of the specimen was recorded. At the final stage, the model underwent load to failure axially using displacement control mode at a speed of 15 mm/min, and the maximum force that caused failure in the specimen was recorded.

Failure was defined as a sudden fall in the load displacement, plastic deformation in the plate, or breakage in the bone or graft for axial loading.

Initial and final stiffness values were obtained from the torque-angle curves in torsional tests and from the load-displacement curves in axis loading tests, which were recorded by the MTS test device. Initial and final stiffness values, failure loads, and amount of displacement between the 2 ends of the models at failure were compared between the groups. Modes of failure were also described.

The statistical analyses were performed using IBM Statistical Package for the Social Sciences Statistics for Windows, ver. 22.0 (IBM Corp., Armonk, NY, USA). Comparisons were made using 1-way analysis of variance, and significance was set at $\alpha < .05$ with a 95% CI. Groups were compared using Tukey's post hoc range test.

Results

Table 1 summarizes the mean initial and final stiffness values of the groups under rotational and axial forces, failure loads, and displacement between the 2 ends of the models at failure.

The mean initial torsional stiffness value of group 4 was significantly higher than those of groups 1 ($P < 0.001$), 2, and 3 ($P = 0.002$), and 5 ($P = 0.029$). There was no significant difference in the initial stiffness among groups 1, 2, 3, and 5.

After cyclical loading, the mean final rotational stiffness value of group 4 was significantly higher than those of groups 1, 2 and 3 ($P < 0.001$). It was also significantly higher in group 5 than in group 1 ($P = 0.04$). No difference was seen between groups 4 and 5 ($P = 0.061$). There was also no significant difference among groups 1, 2, and 3 (Figure 5).

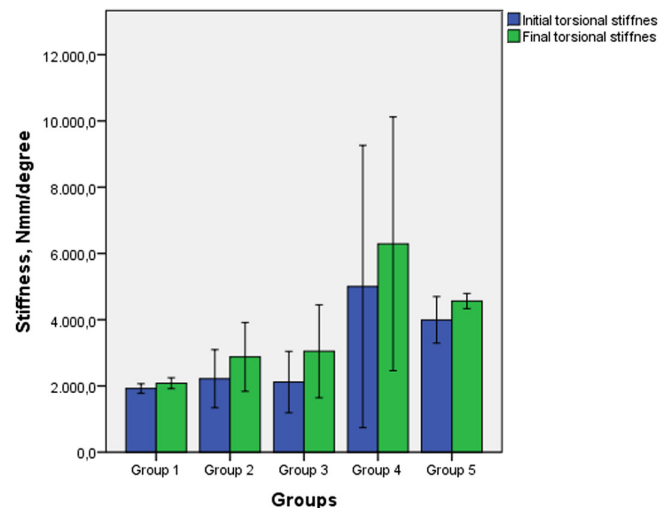


Figure 5. Initial and final stiffness values of the groups under rotational forces.

Table 1. Summary of the results

	Fibula autograft						Cortical strut allograft				
	Group 1		Group 2		Group 3		Group 4		Group 5		P
	without graft		anterior fibula		medial fibula		medial strut		anterior strut		
	Mean	SD	Mean	SD	Mean	SD	Mean	SD	Mean	SD	
Failure load	3696.9	888.9	3901.7	649.2	4026.3	539.0	4733.7	309.5	4272.7	691.0	0.148
Failure displacement	14.4	6.2	17.2	5.6	13.8	2.1	8.6	0.8	12.6	4.8	0.07
Initial axial stiffness	751.0	151.5	907.5	98.7	937.7	261.6	1820.8	146.4	1246.6	141.6	*
Final axial stiffness	802.5	170.5	1097.6	85.2	1155.9	259.9	1996.7	142.9	1391.7	133.7	**
Initial torsional stiffness	1924.1	72.1	2218.8	437.8	2117.1	463.1	5004.0	2130.0	3994.2	352.4	†
Final torsional stiffness	2082.7	81.7	2877.0	519.9	3046.5	702.2	6290.8	1914.3	4563.9	114.4	††

*.001 for group 5 versus group 1; .034 for group 5 versus group 2; and <.001 for group 4 versus all.

†.025 for group 3 versus group 1; <.001 for group 5 versus group 1; and <.001 for group 4 versus all.

†.001 for group 4 versus group 1; .002 for group 4 versus groups 2 and 3; .029 for group 5 versus group 1.

††<.000 for group 4 versus groups 1, 2, and 3; .04 for group 5 versus group 1.

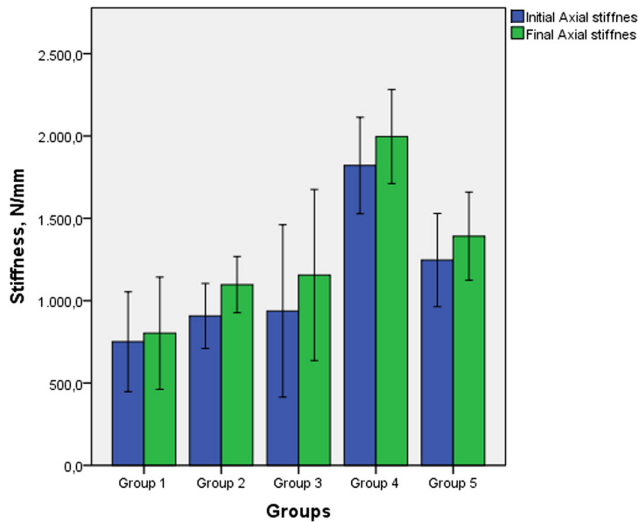


Figure 6. Initial and final stiffness values of the groups under axial forces.

The mean initial axial stiffness was significantly higher in group 4 than in all other groups ($P < 0.001$). Group 5 had higher initial axial stiffness than groups 1 ($P = 0.001$) and 2 ($P = 0.034$), but it was not significantly different among groups 1-3. Comparison between groups 3 and 5 revealed similar initial axial stiffness ($P = 0.061$).

After cyclical axial loading, the mean stiffness of group 4 was significantly higher than in all other groups ($P < 0.001$) but it was significantly lower in group 1 than in group 3 ($P = 0.025$) and group 5 ($P < 0.001$). The mean final axial stiffness value of group 5 was statistically similar to those of groups 2 ($P = 0.08$) and 3 ($P = 0.216$) (Figure 6).

The mean failure load was highest in group 4, followed in order by groups 5, 3, and 2, and it was lowest in group 1. However, the differences among the groups were not significant ($P = 0.148$) (Figure 7). The mean displacements between the 2 ends of the constructs at failure were also similar among groups ($P = .07$) (Figure 8).

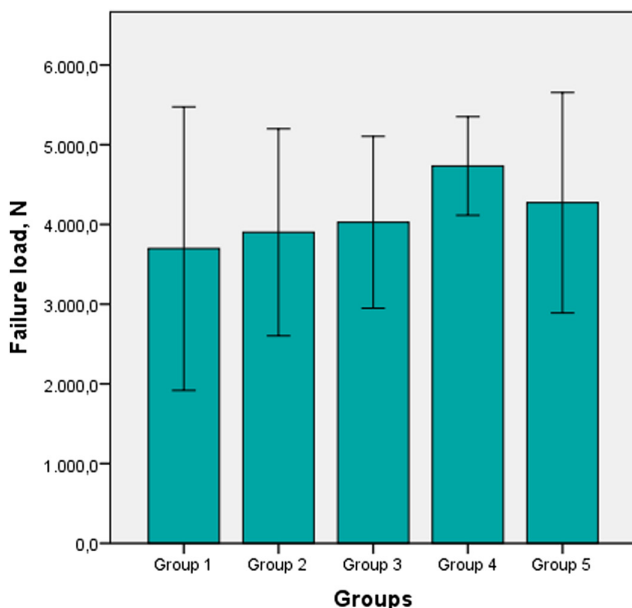


Figure 7. Load to failure results.

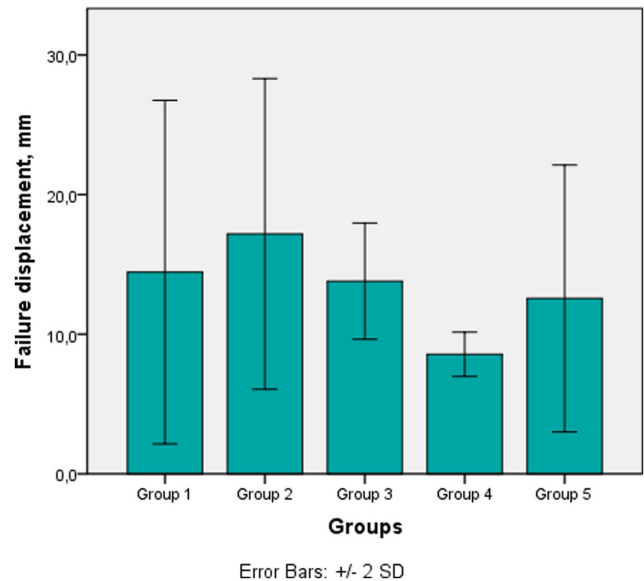


Figure 8. The amount of axial displacements between two ends of the models at failure.

We investigated all models in terms of their failure modes. In group 1, failure involved an oblique bone fracture in all models at the entry point of the first proximal screw. In groups 2 and 3, the failure mechanism was a crack around the first proximal screw and fracture of the grafts. In group 2, only 1 piece of graft in 3 models and 2 pieces of graft in 2 models had fractures around the second proximal cable. In group 3, 1 piece of graft at the level of the proximal cable in a model, 1 piece of graft at the level of the gap in 2 models, 2 pieces of graft at the level of the gap in 1 model, and 2 pieces of fibula at the level of the third proximal cable in a model underwent fracture. In group 4, a transverse fracture occurred through strut allografts of all models at the level of cable 2, and in group 5, failure involved an oblique bone fracture in 4 of the models near the distal end of the bone. In the last model of group 5, failure occurred at the entry point of the proximal screw. In all 25 specimens, prostheses were stable macroscopically.

Discussion

Structural bone allografts are useful for providing bone stock and stability of fixation to reconstruct the femur.^{4,14,15} They have been used successfully for treating PFFs for many years.^{15,16} Contrary to most algorithms,¹⁷ some authors reported successful results without using structural allografts in patients who underwent revision surgeries due to PFF.¹⁸ Although the amount of available cortical autograft is limited, and harvesting procedures may lead to additional risks for patients, fibular autografts have been used successfully to reconstruct bone defects due to tumors, trauma, and osteomyelitis in adults and children.¹⁹

This is the first biomechanical study comparing the stiffness and failure load values of onlay fibula autografts and strut allografts, under cyclic rotational and axial loading. Although we did not find statistical group differences in any parameter, the order of the groups (highest to lowest) in terms of axial and torsional stiffness (before and after cyclical loading) and failure load was as follows: medial strut, anterior strut, medial fibula, anterior fibula, and without graft groups. This order shows the strength of the constructs. This can be explained by the mechanical axis of the extremity, which passes medial to the femoral shaft, and the dimensions and thickness of the

cortical allografts. The stiffness values of the constructs in the fibula groups were statistically similar to that of an allograft placed anteriorly. This was an unexpected finding. The stiffness values of the constructs in fibula groups were similar to those of allografts placed anteriorly. We think that this can be attributed to the strength of plate fixation. Previous techniques described by Ogden et al²⁰ include cable or cerclage wire fixation because of the stem inside the medulla of the proximal part causes difficulty in placing screws. However, with the advent of locking plate technologies, higher fixation strength can be achieved, as described by Fulkerson et al.²¹

Although graft size may be important for stabilization, the bone defect around the fracture is often modest in Vancouver type B1 fractures, such that these defects can be reconstructed, and bone stock provided, using cortical autografts. However, the thickness and length of the fibular graft may have affected our results. Another important point is that the addition (and position) of the grafts did not change the failure loads of the constructs in a previous study.¹² Technically, it is very difficult to place a graft medially and it poses certain risks. Strut allografts used to fix the fractures were placed on the lateral or anterior femur to ensure less surgical dissection.^{22,23} A previous study showed that failure occurred in the strut graft in the group with a medial strut; in clinical practice, this suggests that patients who receive a medial strut allograft may experience graft breakage when physiological loading is exceeded.¹² Therefore, the surgeon should evaluate the patient for periprosthetic fractures during follow-up. Another biomechanical study of a PFF model on artificial bones compared different reconstruction methods with or without allografts and locking screws and found no differences in failure load between reconstruction types.²⁴ Similarly, fibula grafts and their position did not make a difference in terms of failure loads. Although there was no statistical difference, the amount of displacement for failure was greater in the fibula graft groups than in the allograft groups, suggesting that they are more flexible. The minimum measured failure loads of the models in groups 2 and 3 were 3219 and 3157 N, respectively, indicating that they can withstand loads of at least 3-4 times body weight.²⁵ Zdero et al²⁴ reported that immediate postoperative stability of constructs in PFFs has moderate clinical importance because most failures occur at an average of 22 months after initial fixation. Accordingly, we suggest the use of fibula autografts in Vancouver type B1 PFFs because the biomechanical characteristics are not an issue when compared to anteriorly placed allografts.

Tuncay et al²⁶ used autogenous fibulas as structural onlay grafts in 20 patients with bone defects due to periprosthetic fracture or revision and compared the results with 20 patients in whom cortical strut allografts were used. After a minimum 2-year follow-up, the patients with fibula autografts had results comparable to the allograft group and a high success rate.²⁶ More importantly, the graft incorporation time was shorter in the fibula group. None of these patients had donor site complaints.

In experimental biomechanical studies, cadaveric bone is the best in vivo environment; however, obtaining cadaveric bones with similar mechanical and geometrical properties is difficult.²⁷ Synthetic bones have been used as biomechanical test materials in experiments.^{12,13} Synthetic bone, especially fourth-generation synthetic bone, is of good quality and has geometrical, density, and mechanical characteristics, similar (but not identical) to healthy human bone.²⁴

This biomechanical study had some limitations. First, since it was conducted in vitro, the effects of muscles, ligaments, and joints, and actual walking biomechanics, could not be measured. Also, possible graft osseointegration, which might have influenced the findings, could not be investigated in this biomechanical study. Second, all tests were made using artificial bone models. Although they have similar biomechanical and geometric characteristics to human femora and fibulae, they may not behave in exactly the same way as real bones, especially osteoporotic bone in which PFFs are common. Third, we could test only the initial biomechanical properties and those after 10 000 cyclical loads; however, we could not evaluate fatigue after more cycles or the effects of healing on the grafts and constructs. Finally, PFFs occur more in patients undergoing revision surgeries or with a loose stem while we studied intact femora.

Our biomechanical test results reveal that the fixation rigidity of Vancouver type B1 periprosthetic fractures is highest if structural cortical allografts are placed medially. Surprisingly, the use of cortical fibula autografts on the medial or anterior side of the defect can provide similar fixation strength to allografts placed anterior to the cortex, under both axial and rotational forces. Neither cortical allografts nor fibula autografts significantly altered the required amount of force for the failure of the reconstruction if a long locking plate was applied laterally. With regard to the clinical relevance of this study, we conclude that in the treatment of Vancouver type B1 periprosthetic fracture without large segmental defects, onlay fibula strut autografts can be used as an alternative to allografts because they can provide mechanical support until fixation and are inexpensive and readily available. Furthermore, their potential biological effects on union and graft incorporation can be an important advantage and merit clinical and experimental investigation.

Ethics Committee Approval: Ethics committee approval is not required for this biomechanical study.

Informed Consent: Written informed consent was obtained from all participants who participated in this study.

Author Contributions: Concept - F.Y., K.S., İ.T.; Design - F.Y., O.A., K.S., İ.T.; Supervision - F.Y., K.S.; Materials - F.Y., T.E., O.A.; Data Collection and/or Processing - T.E., O.A., E.B., E.S.; Analysis and/or Interpretation - T.E., O.A., K.S., E.B., E.S.; Literature Review - F.Y., O.A., İ.T.; Writing - F.Y., O.A., İ.T.; Critical Review - F.Y., T.E., O.A., K.S., İ.T.

Declaration of Interests: The authors have no conflicts of interest to declare.

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