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Original research

Femur Strength is Similar Before and After Iatrogenic Fracture During Total Hip Arthroplasty: A Biomechanical Analysis

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Background: The purpose of this study was to compare the biomechanical strength of femurs before an iatrogenic periprosthetic fracture vs after an initial fracture with supporting cerclage fixation during cementless total hip arthroplasty.
Material and methods: Nineteen composite femurs and 5 matched pairs of cadaveric femurs were implanted with a single-wedge or dual-wedge tapered femoral stem and tested for ultimate load to failure producing a periprosthetic fracture. Following initial fracture, each femur was cerclaged with Vitallium cables and retested for ultimate load to failure. The mean force eliciting iatrogenic fracture before cabling and that after cabling were compared with a two-sided paired Student's t-test.
Results: All composite femurs developed periprosthetic fractures with an average length extension from the calcar of 75.17 mm. For the 19 composite femurs, the mean ultimate load to failure before cabling and that after cabling were not significantly different (2422.95 N vs 2505.14 N, $P = .678$). For the 10 cadaveric femurs, the mean ultimate load to failure for the initial fracture vs that after cabling was statistically comparable (5828.62 N vs 7002.63 N, $P = .126$). Subanalysis of the 5 cadaveric femurs with a double-wedge stem revealed a significantly higher mean load to failure following cabling (5007.38 N vs 7811.17 N, $P = .011$).
Conclusion: Biomechanical strength was similar for femurs that sustained an initial iatrogenic periprosthetic fracture and the same femurs cabled with cerclage wires after being fractured. These data may assist in operative decision-making for treating iatrogenic fractures during total hip arthroplasty.

Introduction

Total hip arthroplasty (THA) is a highly successful operation with greater than 85% survivorship at 25-year follow-up [1,2]. It has been revolutionary in improving quality of life for patients with end-stage hip osteoarthritis [3,4]. The annual volume of THA performed in the inpatient setting in the United States is projected to increase by 71% by the year 2030, [5] which notably does not account for increasing trends of outpatient THA [6,7]. Early THA most commonly involved various types of cemented femoral stems which demonstrated high long-term survivorship [8,9]. Although previously utilized in only half of all THA procedures, modern cementless THA implants have gained popularity in the United States with contemporary utilization higher than 93% [10].

Although THA is a remarkably successful operation, common etiologies of failure include instability [11], aseptic loosening [12], infection [13], and periprosthetic fracture [14]. During cementless THA, it is imperative that surgeons obtain initial mechanical stability of the femoral stem to reduce micromotion and increase bony fixation [15,16]. A known complication during femoral preparation is iatrogenic periprosthetic fracture, which is estimated to occur in 1.5%-27.8% of cases and is more common in uncemented THA [16-21]. Unrecognized iatrogenic fractures can propagate and

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decrease initial mechanical stability, [22] leading to early loosening and revision [17,22,23].

Iatrogenic periprosthetic fractures during THA may display various fracture morphologies, ranging from nondisplaced calcar fractures to displaced fractures extending to the proximal shaft [24,25]. Management options include extramedullary fixation with cerclage cables and reinsertion of the initial femoral stem, the use of longer tapered or modular distally fixed stems, open reduction and internal fixation with plates and screws, and strut grafts [25-30]. Numerous studies have reported no significant impact on survivorship as compared to controls when these fractures are identified and treated promptly [17,19,31-33].

Prior studies have analyzed the relative efficacy of different fixation constructs in stabilizing iatrogenic periprosthetic fractures [34-37], as well as the potential risk reduction of sustaining such fractures with prophylactic cabling [20,38]. However, the ultimate strength of a femur with a tapered stem and cerclage wires after an iatrogenic periprosthetic fracture relative to the initial biomechanical strength of the same femur has not been described. The purpose of this study was to compare the biomechanical strength of femurs before an iatrogenic periprosthetic fracture vs that after an initial fracture with supporting cerclage fixation.

Material and methods

After obtaining approval from our institutional review board (IRB #2021-060), 2 cementless femoral stem types with differing geometry from Stryker (Stryker Ltd., Kalamazoo, MI) were investigated: the Stryker Accolade II and the Stryker Secur-Fit Advanced (SFA) as seen in Figure 1. These 2 stem types were chosen to represent the most common tapered stem geometries typically used in THA: the single-wedge (Accolade II) and the dual-wedge (SFA) geometry [39].

Cadaveric specimen preparation

Cadavers in this study were screened to include those with no history of musculoskeletal disease, defects, or previous surgeries. After obtaining specimens from Science Care (Phoenix, AZ), 5 pairs of matched cadaveric femurs were procured from 3 male and 2 female donors (age range 66-90 years, mean 81.6 $\pm$ 9.8 years). Prior to harvesting all femurs, computed tomography (CT) scans were performed to template the appropriate-size implant to mitigate any undersized or oversized stems (Fig. 2). Based on the Dorr classification, the femurs from cadavers 1, 3, and 4 were classified as type B bone, and the femurs from cadavers 2 and 5 were classified as type B/C bone. All left femurs were designated the appropriately sized Stryker Accolade II femoral stem, and all right femurs were designated the appropriately sized Stryker SFA femoral stem as there was no difference in bone quality or size between right and left.

After templating based on CT scans, the femoral neck osteotomy was created proximal to the lesser trochanter on each femur in the correct angle and position according to the preoperative template. The canal was prepared by reaming and broaching the SFA stems to the appropriate-size implant per system guide, and for the Accolade II, a broach-only system was used per implant guide recommendations. All right-sided femurs were implanted with the SFA, and all Accolade II stems were placed in the left side of the cadavers. A second validation was performed using X-ray to ensure the broach was in the appropriate position, and the stem was appropriately sized to ensure there were no undersized stems (Fig. 3). After broaching was performed and the final stem was inserted, the femurs were visually inspected to ensure there were no fractures created. To standardize the length of the femur mechanically, the distal end of each femur was cut so that 80% of the total length of the femur remained. The intramedullary canal of the distal end of the femur was cemented with a 1/2 in. $\times$ 4 in. tapered steel rod using Stryker Simplex HV cement (Stryker Ltd., Kalamazoo, MI) in a neutral position. The tapered steel rod was inserted into a custom holder for the MTS so that the femoral canal representing a neutral mechanical axis is at 0° angle. The femurs were immediately tested after implanting the final femoral stem to mitigate dehydration and represent a surgical environment.

Composite bone model preparation

Twenty commercially available composite osteoporotic femurs with 10 pounds per cubic foot solid foam and 16-mm canal models

Figure 1. (a) Stryker SFA implant representative of dual-wedge femoral stem geometry. (b) Stryker Accolade II implant representative of single-wedge femoral stem geometry.

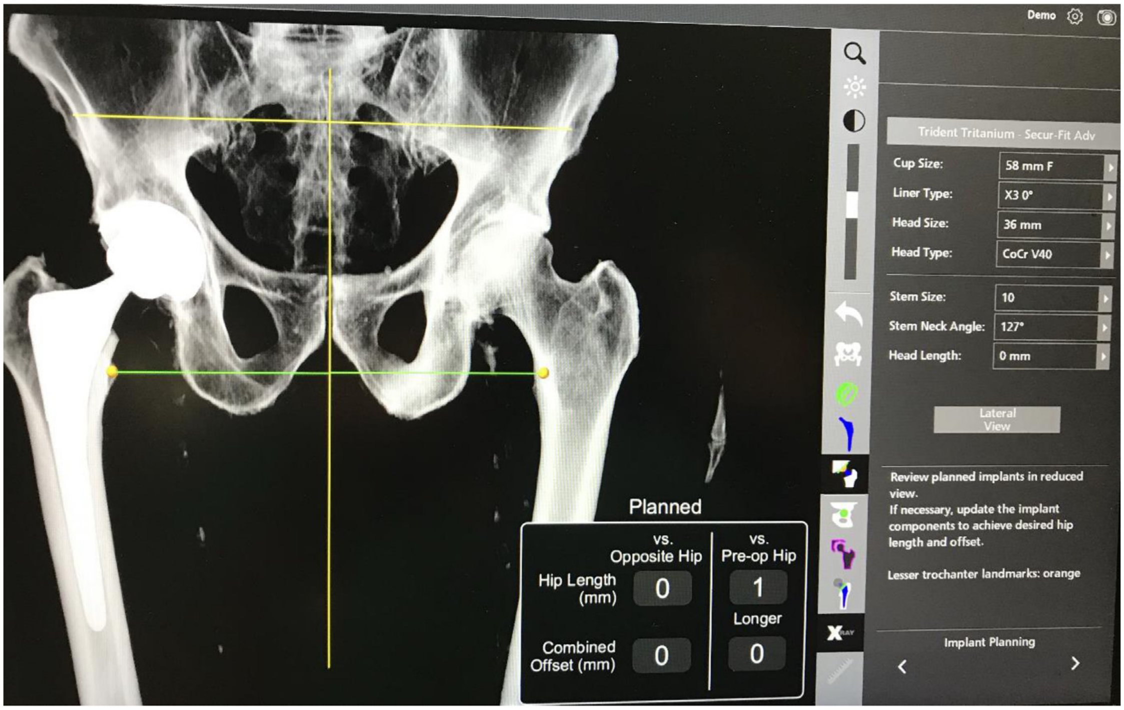

Figure 2. CT scanning and templating for SFA stem.

from Sawbones USA (Pacific Research Laboratories, Vashon Island, WA) were used for this investigation. The uniform size, shape, and material properties were used to reduce the variability often associated with the use of human cadaveric bones and have been shown to be accurate in mechanical testing [40].

Each composite femur was prepared in the same method as the cadavers resulting in 10 specimens receiving an SFA stem and 10 receiving an Accolade II stem. One Accolade II specimen was excluded due to a mechanical malfunction of the MTS machine, which prevented data collection. Therefore, 10 composite Sawbone femurs with an SFA stem and 9 composite Sawbone femurs with an Accolade II stem underwent biomechanical testing.

Figure 3. Second validation of stem sizing using X-ray; the image shows Accolade II size 8 stem broach in place.

Mechanical testing

Each femur was loaded in a biaxial servohydraulic test system (MTS Bionix 370; MTS Systems Corp., Eden Prairie, MN). A Stryker McReynolds Distal Stem Adaptor (Stryker Ltd., Kalamazoo, MI) was threaded into the femoral stem and secured to the compression plate via a cylindrical attachment [41,42]. Room temperature was controlled at 22°C. Each femur was first axially preloaded to 100 N for 30 seconds prior to loading to failure at a displacement rate of 5 mm/s in axial compression [41]. The maximum load before failure was recorded for each specimen by the servohydraulic test system to ensure accuracy.

After each cadaveric femur was fractured on the MTS machine with these forces and ultimate load to failure was established (Fig. 4), the femoral implant was removed. Two, 2.0-mm Vitallium Dall-Miles cables (Stryker Ltd., Kalamazoo, MI) were placed on the proximal femur with 1 placed above and 1 immediately below the lesser trochanter and tightened to 150 PSi (pounds per square inch). After cabling of the femur, the same femoral stem was inserted to the same level prior to the fracture and retested on the MTS machine with the same parameters to determine the ultimate load to failure of the fractured femur with 2 cables in place (Fig. 5). This method was repeated for the composite Sawbone models (Figs. 6 and 7). The initial fracture pattern observed in all cadaveric specimen and composite models was a displaced fracture extending from the cut surface of the calcar to the lesser trochanter, nearly all with a spiral component. This fracture propagated from the calcar to just below the lesser trochanter in all femurs after retesting with cerclage cables.

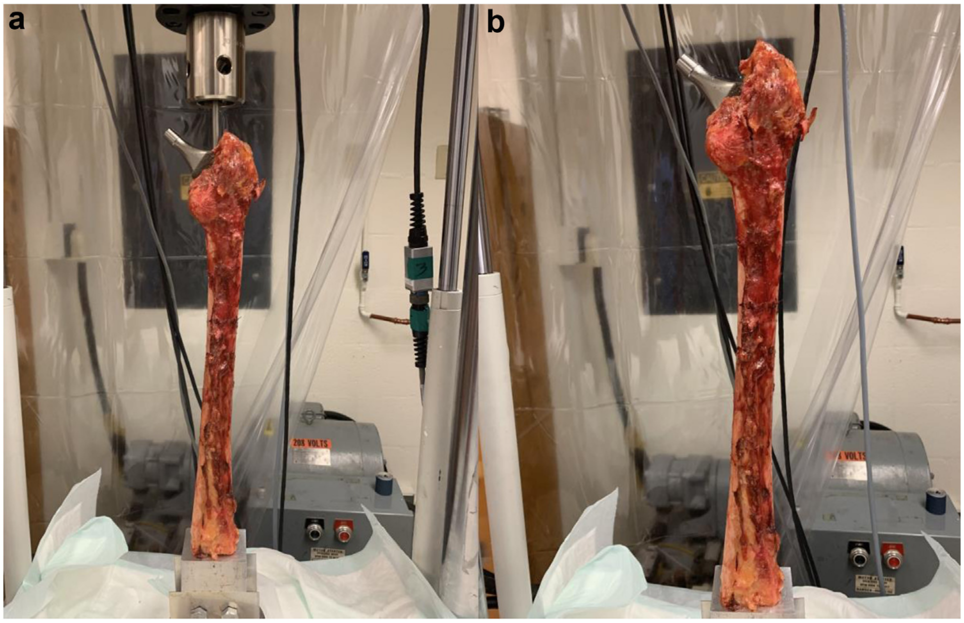

Figure 4. (a) Cadaveric femur loaded on an MTS machine. (b) Periprosthetic fracture caused by an MTS machine.

Statistical analysis

Statistical analysis was performed using Microsoft Excel (Microsoft Corp., Redmond, WA) with the XLSTAT add-on (Addinsoft Inc., New York, NY). All data are reported as mean ± standard deviation (SD). Mean force causing ultimate load to failure for the native cadaveric and Sawbone femurs during initial instrumentation vs the same femurs refractured after cabling were compared using a two-sided, paired Student's t-test. The analysis was performed both with data for both implants pooled together and at an implant-specific level (ie, mean ultimate load to failure before cerclage and that after were compared individually for the Stryker Accolade II and Stryker SFA implants) for both the Sawbone and cadaver models. A $P < .05$ was considered statistically significant. *Post hoc* power analyses were performed to assess adequacy of the included sample sizes based on the observed differences.

Results

All Sawbones developed fractures with an average length extension of 75.17 mm from the calcar. Measurements of ultimate load to failure for the initial fracture and refracture following cabling are outlined in Table 1. Pooling data for all 19 Sawbones, the mean ultimate load to failure before cabling and that after were not significantly different (2422.95 ± 1030.47 N vs 2505.14 ± 582.03 N,

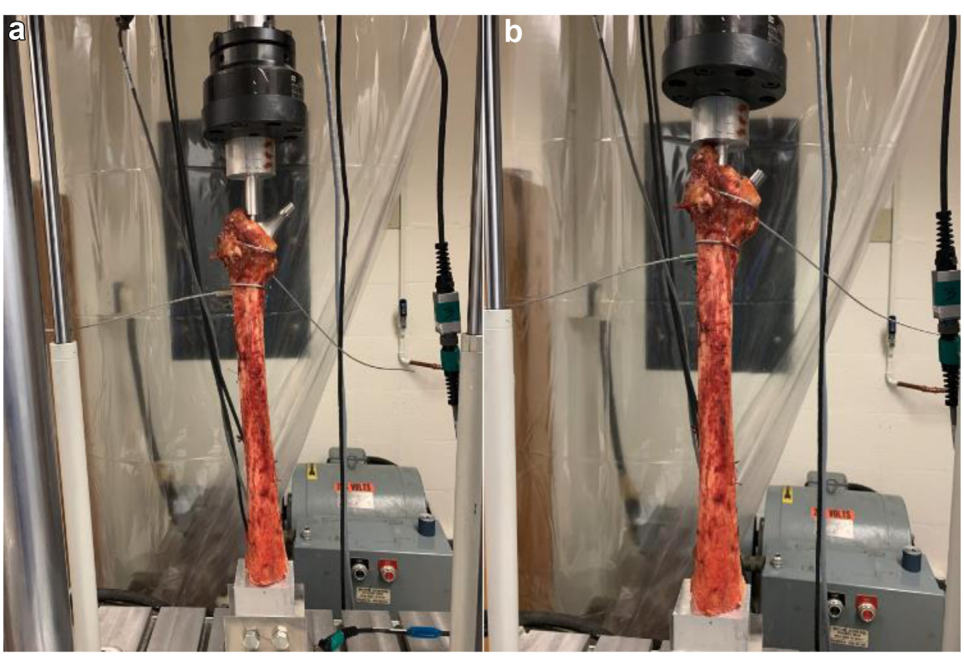

Figure 5. (a) Cadaveric femur with cables after periprosthetic fracture. (b) Cabled cadaveric femur retested on an MTS machine after fracture.

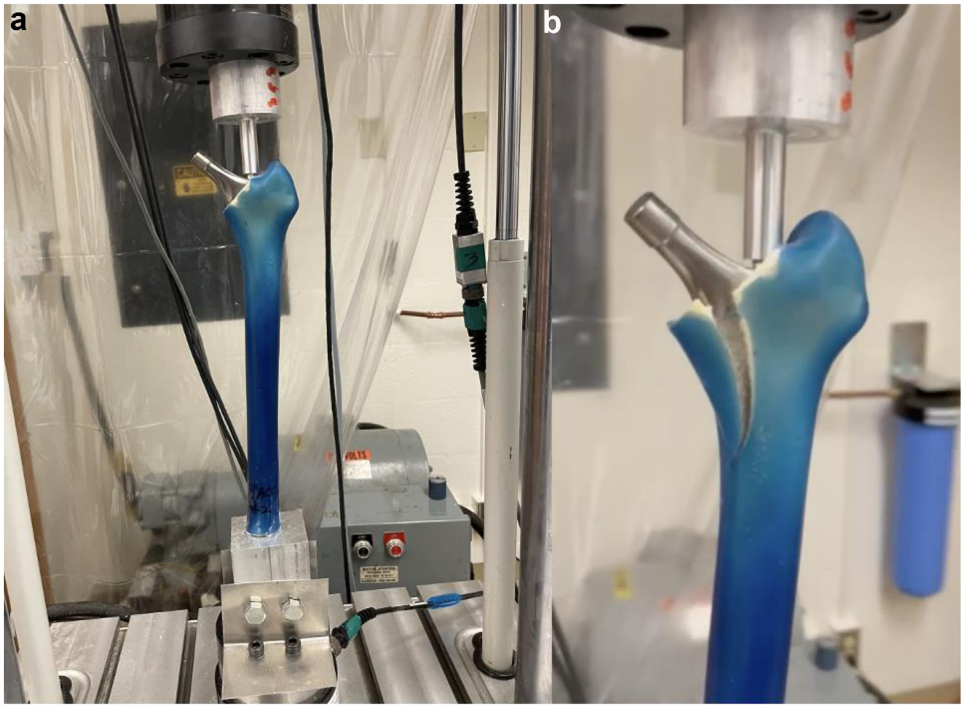

Figure 6. (a) Composite Sawbone femur loaded on an MTS machine. (b) Periprosthetic fracture created during biomechanical testing.

$P = .678$). Analyzing each stem type individually, the mean ultimate load to failure for Sawbones with an SFA stem (1983.91 ± 969.20 N vs 2294.07 ± 593.56 N, $P = .270$) and an Accolade stem (2910.56 ± 907.75 N vs 2739.67 ± 498.28 N, $P = .560$) were also statistically comparable (Fig. 8).

A *post hoc* power analysis revealed low power for the pooled Sawbone analysis (~3%), SFA stem subanalysis (~18%), and Accolade stem subanalysis (~8%). The analysis suggests that 842, 59, and 194 composite femurs would be required for the pooled data, SFA, and Accolade analyses, respectively, to achieve 80% power for these parameters.

Measurements of ultimate load to failure for the initial fracture of the 10 cadaveric femurs and refracture following cabling are outlined in Table 2. Pooling data together for all 10 matched femur pairs, the mean ultimate load to failure was not significantly different between the initial fracture and after cabling (5828.62 ±

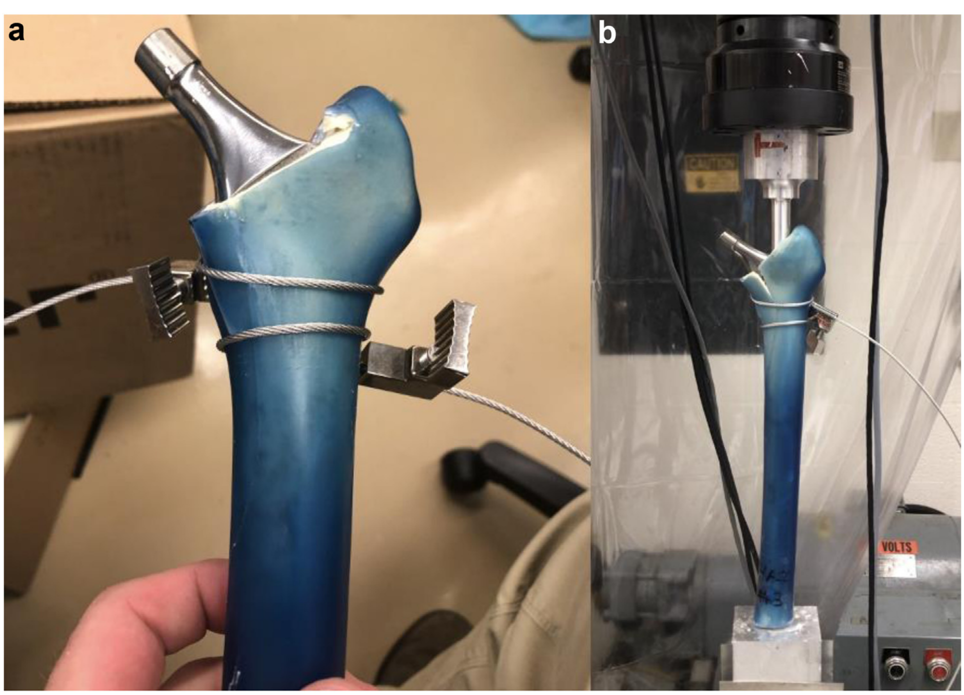

Figure 7. (a) Composite Sawbone femur with cables after an initial fracture. (b) Periprosthetic fracture created during biomechanical testing after cabling.

Table 1
Results of biomechanical testing on Sawbone femurs.

Sawbone #	Implant	Stem size	Ultimate load to failure (N)		
			Precerclage	Postcerclage	Difference (post − pre)
1	SFA	7	2060.84	2433.54	372.70
2	SFA	7	1801.26	2553.74	752.48
3	SFA	7	1630.74	1569.31	−61.43
4	SFA	7	2610.40	1866.35	−744.06
5	SFA	7	2146.12	2182.35	36.23
6	SFA	7	786.65	1276.44	489.79
7	SFA	7	605.36	2192.72	1587.36
8	SFA	7	2781.69	2992.84	211.15
9	SFA	7	1513.74	2976.17	1462.43
10	SFA	7	3902.24	2897.24	−1005.00
11	Accolade	4	3049.36	2915.07	−134.29
12	Accolade	4	1030.12	2245.72	1215.60
13	Accolade	4	3333.77	3655.57	321.81
14	Accolade	4	2348.00	2079.80	−268.20
15	Accolade	4	3436.65	2260.26	−1176.38
16	Accolade	4	2971.43	2778.09	−193.35
17	Accolade	4	2282.41	3079.66	797.25
18	Accolade	4	3774.92	2625.34	−1149.58
19	Accolade	4	3968.39	3017.54	−950.85
Mean ± SD			2422.95 ± 1030.47	2505.14 ± 582.03	82.30 ± 851.20

Table 2
Results of biomechanical testing on cadaveric femurs.

Cadaver #	Femur #	Stem	Stem size	Ultimate load to failure (N)		
				Precerclage	Postcerclage	Mean difference (post − pre)
1	1	Accolade	6	8935.40	6156.07	−2779.33
	2	SFA	9	6291.14	7257.96	966.82
2	3	Accolade	5	6569.03	6073.34	−495.69
	4	SFA	7	4351.74	6347.55	1995.81
3	5	Accolade	10	3658.60	5108.02	1449.42
	6	SFA	12	3239.67	7011.50	3771.83
4	7	Accolade	4	6521.11	6379.18	−141.93
	8	SFA	8	4623.05	7359.79	2736.74
5	9	Accolade	8	7565.13	7253.81	−311.32
	10	SFA	10	6531.32	11,079.05	4547.73
Mean ± SD				5828.62 ± 1808.91	7002.63 ± 1593.76	1174.01 ± 2204.08

1808.91 N vs 7002.63 ± 1593.76 N, P = .126). Analysis of the 5 cadaveric femurs with an SFA stem (Fig. 9a) revealed significantly higher mean load to failure following cabling (5007.38 ± 1394.99 N vs 7811.17 ± 1868.86 N, P = .011). For the 5 cadaveric femurs with an Accolade stem (Fig. 9b), the mean load to failure before cabling and that after were comparable (6649.85 ± 1938.57 N vs 6194.09 ± 766.95 N, P = .538).

A *post hoc* power analysis showed that the analysis of the 5 matched pairs with SFA implants was adequately powered (~90%), while the Accolade (~8%) and pooled data (~15%) analyses had low power. To achieve 80% power with these parameters, the pooled data and Accolade-only analyses would need 30 and 89 matched pairs, respectively.

Discussion

The present study demonstrated comparable biomechanical strength of femurs that sustained an initial iatrogenic periprosthetic fracture and the same femurs cabled with cerclage wires after being fractured. This result was replicated in both a composite bone and cadaveric model. Additionally, in the analysis of 5 matched cadaver femurs with SFA implants, the mean ultimate load to failure was significantly higher following cabling. The SFA stem is a double-wedge design such that the cables may confer more stability to this geometry as opposed to the Accolade II, which is a single wedge, as the SFA may have more contact with the metaphyseal-diaphyseal bone at the area of the circumferential cables [39]. Although performed in vitro, these data suggest biomechanical strength of the femur with cables after iatrogenic fracture during THA is not significantly lower, which has important implications for operative management strategy of these injuries.

Despite abundant innovations in implant design and surgical technique, iatrogenic periprosthetic fractures remain a key issue during primary THA affecting between 1.5% and 27.8% of cases [16-20]. It must also be acknowledged that, because many surgeons do not bill for repair of an iatrogenic periprosthetic fracture during primary THA, the true incidence may be higher [43]. Several risk factors have been identified including minimally invasive surgical techniques, female sex, metabolic bone diseases, technical issues during surgery, and the use of press-fit cementless stems [18,21,44]. Abdel et al. reported intraoperative periprosthetic fractures were 14 times more common with the use of uncemented stems vs a cemented prosthesis [14]. This elevated risk is likely attributed to the enhanced importance of maintaining initial mechanical stability during preparation of the femur in the absence of cement.

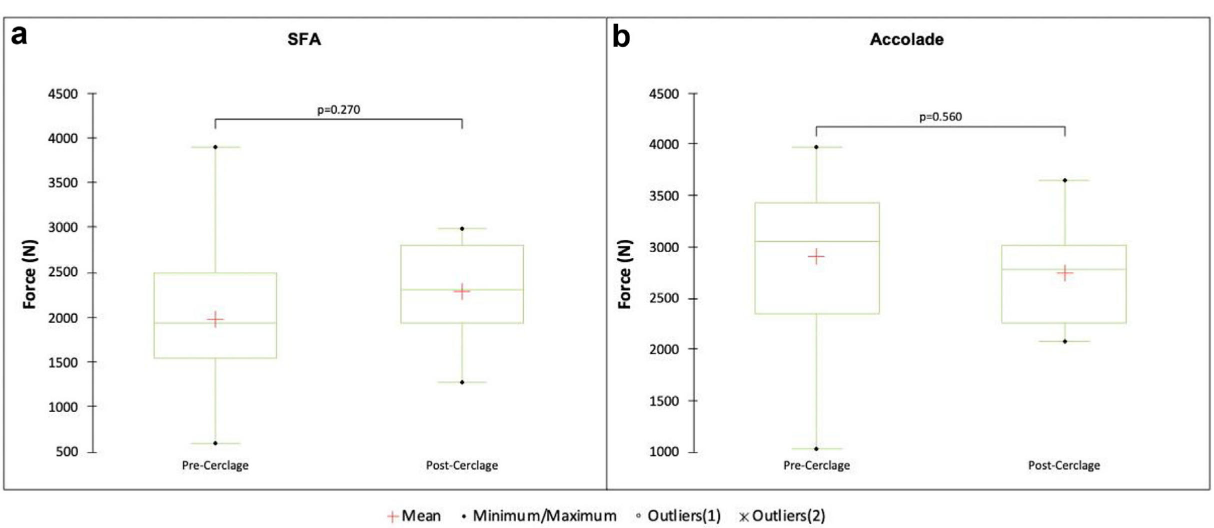

Figure 8. Implant-specific ultimate load to failure for Sawbones with (a) an SFA stem and (b) Accolade stem.

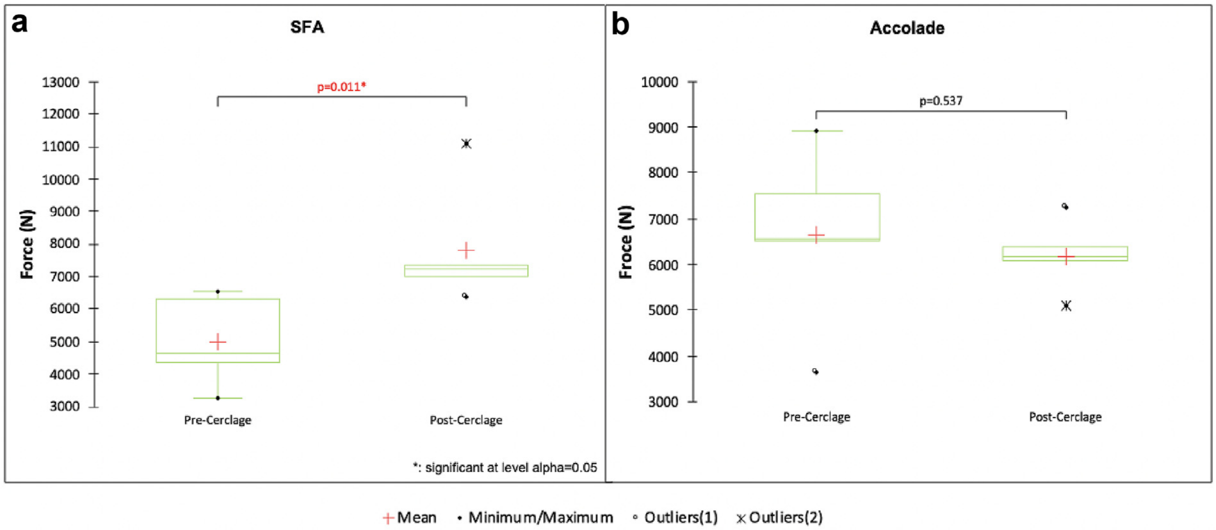

Figure 9. Implant-specific ultimate load to failure for cadaveric femurs with (a) an SFA stem and (b) Accolade stem.

During insertion of the press-fit femoral implant, calcar hoop stresses increase by 25%-400%, which can cause iatrogenic fracture [45].

With small, nondisplaced intraoperative calcar fractures, joint surgeons often remove the implant, cable the proximal femur, and replace with the same stem [27,33]. However, this study presents the first mechanical data suggesting there is no decrease in initial mechanical strength of the femur after iatrogenic periprosthetic fractures during femoral preparation. When fractures propagate further into the proximal femur and below the calcar, surgeons often use distally fixed implants to bypass the stress on the fractured area [29,46]. There are, however, disadvantages to using distally fixed stems in primary cases including cost of the implant and the potential for thigh pain that can result from canal-filling stems [39,47]. There is also stress shielding that can be seen for a long term in patients receiving long distally fixed stems, which occurs less commonly with tapered proximal fit designs [39,48,49]. The fractures in this study were created with a force large enough to represent a displaced fracture averaging over 75 mm in length, and cabling was still able to provide the same time zero strength, which may aid surgeons in making the decision to use distally fixed vs proximally tapered stems when treating intraoperative insertional fractures.

It has been shown that it is difficult for patients older than 75 years to maintain a non-weight-bearing status, and most patients are unable to do so after a hip fracture [50,51]. There is also an increased morbidity and adverse events associated with non-weight-bearing in patients after hip fractures [52,53]. These results suggest that complete non-weight-bearing may not be necessary with insertional fractures after cabling with this method. There is also fatigue of the implants that has to be considered such that early weight-bearing with repetitive loading could result in cable loosening; however, these data may be useful when weighing the risks and benefits of strict non-weight-bearing protocols in certain patients.

There are several limitations to this study. Implant survivorship and osteointegration could not be determined as all specimens and composite models were tested in vitro. However, implant testing with respect to ultimate load to failure was done intentionally on in vitro specimens to standardize the procedure and not produce iatrogenic fractures in living individuals. Additionally, a standardized fracture morphology was not artificially created with drill bits and/or osteotomes, but rather the fracture was created using a real implant to best simulate an iatrogenic fracture due to loading. This variation could have contributed to an inconsistency in fracture patterns; however, the intention was to produce a clinically relevant fracture. A third limitation is that although the Sawbone mechanical models are validated, they may not behave in an exact manner as human bone. Furthermore, while cadaveric femurs were classified using the Dorr classification, standardized measures such as dual-energy X-ray absorptiometry or quantitative CT were not performed, and variation in bone quality may have influenced the results. The data collected were also an ultimate force to failure and, as such, did not take into account repetitive loading that would be seen with ambulation over a period of months until bone healing is complete. Another limitation is that, aside from the sub-analysis of cadaveric femurs with an SFA stem, *post hoc* analyses suggested all analyses were underpowered to detect the observed mean differences in ultimate load to failure before and after cerclage. However, as illustrated by the remarkable sample sizes required to achieve statistical significance with the mean differences found in the present study, it seems more likely that our results reflect comparable strength of the femur before and after cerclage rather than inadequate sample sizes. Future analyses with larger samples should seek to test this hypothesis. Additionally, the stems of all femurs in both composite models and cadaveric specimen were loaded with the application of force at a constant rate. This simulation may not be representative of in vivo conditions, in which intermittent impaction is performed with a mallet; however, this methodology was chosen to best represent weight-bearing in the immediate postoperative time period. Lastly, only 2 models of press-fit stems were tested during this study to represent the 2 most common geometries currently in use. There is a chance that different proximally coated press-fit stems could have different strength profiles.

Conclusions

The present study demonstrated comparable biomechanical strength between femurs that sustained an iatrogenic periprosthetic fracture and the same femurs cabled with cerclage wires after being fractured. These data may assist in operative decision-making for treating iatrogenic fractures during THA.

Acknowledgments

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Conflicts of interest

The authors declare that there are no conflicts of interest.

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