Arthroplasty Today 15 (2022) 68–74

Contents lists available at ScienceDirect

Arthroplasty Today

journal homepage: http://www.arthroplastytoday.org/

Original research

Load Sharing in the Femur Using Strut Allografts: A Biomechanical Study

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ARTICLE INFO

Article history:
Received 11 September 2021
Received in revised form
1 February 2022
Accepted 12 February 2022

Keywords:
Strut allograft
Thigh pain
Femoral stem
Total hip replacement
Implant stiffness
Bone strains

ABSTRACT

Background: Femoral strut allografts are used in revision hip arthroplasty for management of bone loss associated with implant failure or periprosthetic fractures. They have also been used to treat unremitting thigh pain in well-fixed cementless femoral stems, to address the differential in structural stiffness between the stem and femoral shaft. Our study used an in vitro biomechanical model to measure the effect of placement of allografts on femoral strains, to determine their load-sharing capacity.
Material and methods: Three rosette strain gauges were applied to the femoral surface of each of 6 cadaveric femurs, at the stem tip level on anterior, medial, and lateral cortices. After stem implantation, cortical strut allografts were applied to the lateral femoral shaft and secured with 4 Dall-Miles cables. A fourth gauge was placed on the midpoint of the allograft. Strains were recorded in the intact femur, then the implanted femur with and without the allograft under simulated physiologic loading in a load frame.
Results: Reduction in distal femoral principal strains, between 12% and 59%, was seen in all cortices following placement of the allograft. Under axial loading, 30% of the strain in the lateral cortex was borne by the allograft. Greater reductions in strain, by as much as 59%, occurred under axial load and torque.
Conclusion: The results of this biomechanical model indicate that by placement of an allograft, cortical strains can be reduced to levels approaching those in an intact femur, supporting this technique for treatment of unremitting thigh pain in well-fixed prostheses.

Introduction

Strut allografts are in common use in revision hip arthroplasty in the setting of bone loss or in the treatment of periprosthetic fractures [1-3]. In addition to these uses, some authors have advocated application of a strut allograft in the treatment of enigmatic thigh pain, that is, thigh pain present in a well-fixed, noninfected femoral stem [4,5].

Some of the clinical success of cementless femoral stems has been marred by the incidence of thigh pain, which has ranged in recent literature between 4% and 18% [6-16]. Often the thigh pain resolves or is mild to moderate in nature and can be treated with analgesics. However, there are a subset of patients who have unremitting thigh pain requiring more aggressive management [4,5,7].

Although the mechanism of thigh pain in a well-fixed stem is not completely understood, some theories have been proposed, suggesting a mechanical etiology. Some authors suggest that thigh pain arises from focal stress transfer to the diaphysis, consistent with cortical hypertrophy seen at the stem tip [17-19]. Micromotion at the stem tip of proximally fixed systems is another proposed mechanism [20,21]. Others propose that a mismatch in structural stiffness between the implant and the femoral shaft causes

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https://doi.org/10.1016/j.artd.2022.02.010

increased strain at the implant bone interface and within the cortical bone, resulting in thigh pain [8,15,16,22-28].

Numerous femoral stem designs have been introduced to optimize periprosthetic strain distribution. The majority of these designs use variations of stem shapes to maximize load transfer to the proximal femur and minimize stress shielding in this region, with the intention of mitigating bone resorption due to stress shielding [29]. At the same time, the distal stem is tapered or sometimes even split to avoid higher bone stresses near the stem tip. However, a number of designs successfully achieve press-fit fixation using precise diaphyseal fit, which may be necessary depending on patient anatomy and bone quality. In addition to using titanium alloy, which has a lower modulus of elasticity, 3D-printed porous titanium femoral stems have been introduced with the intention of minimizing stress shielding [30]. Collectively, these design changes also help mitigate higher femoral stresses near the stem tip. While these improvements in implant design may reduce the overall incidence of thigh pain in the future, the question still remains of how to treat patients with unremitting thigh pain in the presence of a well-fixed femoral component.

To address the patient with unremitting thigh pain while retaining the femoral stem, a technique of using a laterally placed femoral strut allograft cabled to the femoral shaft was used by Dr. Anthony Hedley. In principle, it is designed to increase the sectional modulus of the bone in order to decrease the discrepancy in structural stiffness between the stem and the femoral shaft. Over the last few years, there have been a few small case series that have reported on this technique [4,5].

The purpose of this study was (1) to determine the effect of femoral strain at the stem tip by placement of a cabled allograft and (2) to determine the load-sharing capacity of the allograft cabled to the femur.

Material and methods

Six cadaveric femurs were selected based on radiographs in order to obtain disease-free bone with adequate bone stock. Radiographs with 15% magnification were used to template for the porous-coated anatomic femoral stems (Howmedica, Rutherford, NJ; now Stryker, Kalamazoo, MI). This femoral stem design was selected based on prior clinical series showing a relatively higher rate of enigmatic thigh pain [23,31]. The bones were stripped of soft tissues, and the femoral templates were used to determine the strain gauge position at the anticipated stem tip location. Three rosette strain gauges (Micro-Measurements, Inc., Raleigh, NC) were placed on the prepared cortical surfaces using M-Bond 200 adhesive. The distal femur was cut at the proximal end of the distal metaphysis, and the femurs were potted to a 10-cm depth in epoxy potting material. Specimens were brought to room temperature. Using an MTS-812 servohydraulic load frame, the specimens were loaded 5 times to peak axial load prior to recording measurement. The femurs were then ramp-loaded to 500 N with the load directed through the femoral head and parallel to the shaft of the femur. A linear bearing restricted motion of the femoral head to the frontal plane. While maintaining axial load, internal and external torsional loads of 10 Nm were applied about the femoral shaft. Strains were recorded continuously during axial and torsional loading. This was performed twice for each gauge in order to obtain individual load strain curves.

The intact potted femurs were then implanted with appropriately templated porous-coated anatomic stems by Dr. Anthony Hedley using standard surgical technique. Polymethyl methacrylate (PMMA) cement was used around the proximal porous coating to simulate bone ingrowth. Since fixation with PMMA is established in the arthroplasty literature to provide secure initial fixation of femoral stems, it was deemed to adequately secure the proximal femoral stem in the cadaveric experimentation in the present study to simulate an ingrown implant. Adequate fixation was expected around the porous-coated surface where cement would interdigitate in the pores, particularly for the few cycles of testing and measurement. Radiographs of the implanted femurs were then obtained to confirm strain gauge location relative to the stem tip. In some cases, the strain gauges had to be replaced to a more proximal position in order to keep the gauge within 0.5 cm of the stem tip ($n = 2$). Load was then applied, and strain was measured using the same protocol as for the intact femur.

Next, lateral strut allografts measuring 12-14 cm in length were contoured to fit the lateral cortex of each femur. A fourth rosette strain gauge was applied to the midpoint of the allograft. The strut was then centered over the lateral cortex such that the strain gauge on the allograft was centered over the stem tip. It was then carefully affixed to the femur using four 2.0-mm stainless steel Dall-Miles cables. Load was applied, and strains were recorded using the same protocol as with the intact femur. Additional recordings were made for the rosette on the allograft.

Data analysis and statistical methods

The primary outcomes in this study were principal femoral cortical strains near the stem tip and principal strains measured on the allograft. From the readings of each rosette strain gauge, the maximum and minimum principal strains, maximum shear strains, and the directions of these strains were calculated. This facilitated assessing the greatest strains at each gauge location, regardless of the orientation of the gauge. In the present study, all maximum principal strains were tensile strains, and all minimum principal strains were compressive strains.

The most linear portion of each load-strain curve was used to calculate the slope of the curve. For each gauge, the mean of 2 runs was calculated. Paired *t*-tests were used to compare the principal strains between the intact and implanted specimens and between the implanted and implanted-with-allograft specimens. The use of paired *t*-tests ensured that variations in bone shape and quality were minimized by using the differences in strain among the loading conditions for each femur.

Bar graphs were used to present the mean and standard error of principal strain and shear strain among the cadaver femurs tested with each condition: intact, implanted, and implanted with allograft. Separate *P* values were calculated for comparing tensile strains (maximum principal strains) and compressive strains (minimum principal strains).

Results

Intact bones

As anticipated, under axial load, principal strains in the lateral cortex were predominantly tensile, and those on the medial side were predominantly compressive. Anteriorly, principal strains were predominantly compressive. Under axial load, the peak strain was 1080 $\mu\varepsilon$ tensile on the lateral cortex, 620 $\mu\varepsilon$ compressive on the anterior cortex, and 1700 $\mu\varepsilon$ compressive on the medial cortex (Fig. 1a and b).

With the condition of axial load combined with external rotation, maximal and minimum principal strains were approximately equal, around 600 $\mu\varepsilon$ on the lateral cortex, suggesting strain was affected predominately by torsional loads rather than bending. The relative equivalence of maximum and minimum principal strains in the intact bone under conditions of internal and external rotation

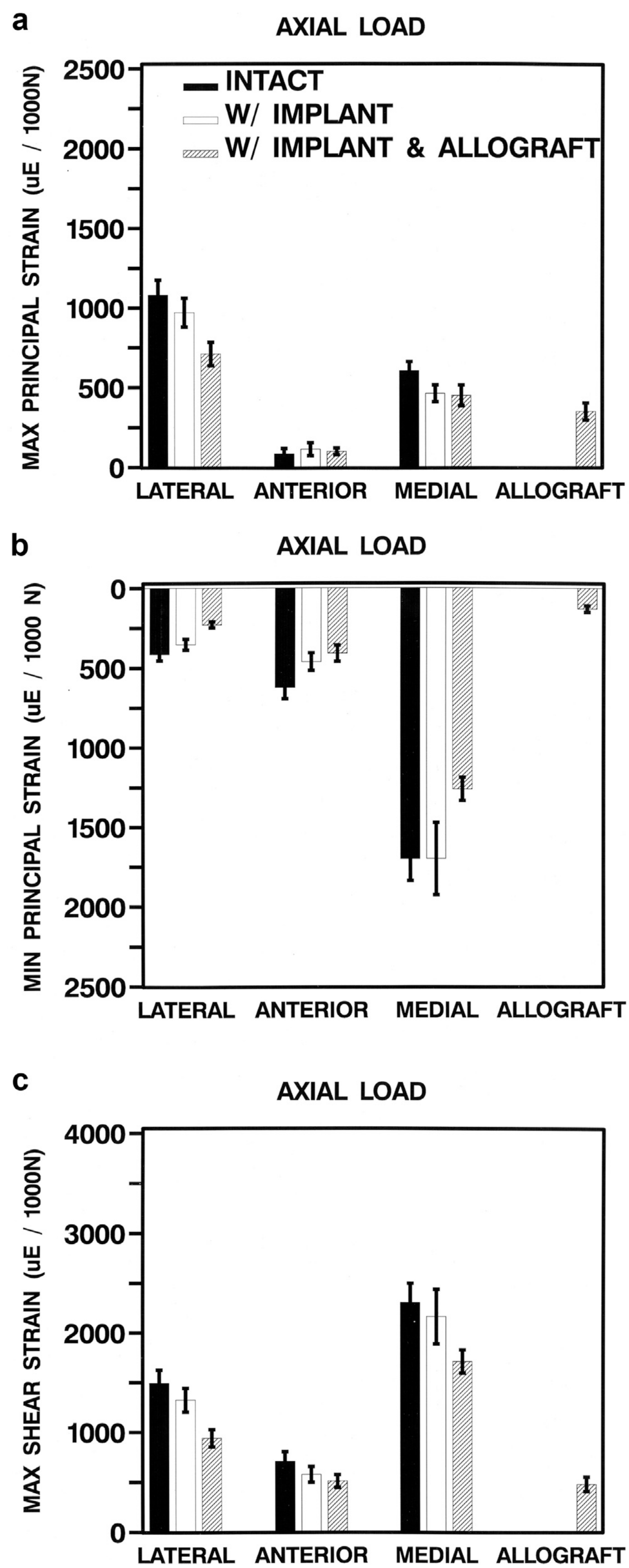


Figure 1. Principal and shear strains in intact specimens, implanted specimens, and implanted specimens with allografts, under axial load alone. Maximum principal strains were all tensile, and minimum principal strains were all compressive. Specimens were loaded to 500 N of axial load, but strains presented here were extrapolated to 1000 N to facilitate comparison with reports using higher loading. (a) Maximum principal strain with axial load; (b) minimum principal strain with axial load; (c) maximum shear with axial load.

was seen in all cortices, again suggesting primarily torsional effects (Figs. 2a, b and 3a, b).

Shear strain was highest in the medial cortex under axial load, with strains around 2300 $\mu\varepsilon$. Under torsional loads, shear strain was highest in the anterior cortex, around 1390 $\mu\varepsilon$ (Figs. 1c, 2c, and 3c).

Implanted bones

Following implantation, under axial load, all gauge measurements remained essentially the same, with a slight decrease in principal strain of 50-100 $\mu\varepsilon$. Shear strains also remained similar, decreasing by about 50-200 $\mu\varepsilon$ (Fig. 1a-c). However, under axial load combined with torsional loads, there was a significant increase in strain, most notably in the anterior cortex, where a nearly threefold increase in tensile strain was observed with external rotation (800 $\mu\varepsilon$ to 2320 $\mu\varepsilon$) ($P = .02$) (Fig. 2a). Similarly, under axial load and internal rotation, compressive strains on the anterior cortex increased more than 1000 $\mu\varepsilon$ from 775 $\mu\varepsilon$ to 1850 $\mu\varepsilon$ ($P = .01$) (Fig. 3b). Shear strain increased on all cortices by 1.5- to 2-fold compared with those measured with the intact femur (Figs. 2c and 3c).

Addition of allograft

Following placement of the laterally cabled allograft, a reduction in strain under axial and torsional loads was observed on all cortices. Under axial load, the allograft reduced strain mainly in the lateral and medial gauges by 15%-20%. The strain gauge on the allograft experienced approximately 350 $\mu\varepsilon$ tensile.

A greater reduction in strain was observed under combined axial and torsional loads, especially on the anterior cortex. With external rotation, strain on the anterior cortex was reduced by approximately 25%, from 2320 $\mu\varepsilon$ tensile to 1600 $\mu\varepsilon$ ($P = .05$). With internal rotation, strain on the anterior cortex was reduced by approximately 25%, from 1850 $\mu\varepsilon$ compressive to 1340 $\mu\varepsilon$ ($P = .04$) (Figs. 2a, b and 3a, b). Again, similar reductions were observed in shear strain, with the allograft experiencing approximately 40% of the shear strain observed on the anterior cortex and 70%-75% of that observed on the lateral cortex (Figs. 1c, 2c, and 3c).

Discussion

In the present study, a biomechanical model was constructed using cadaveric femurs to measure the effect of a strut allograft to reduce femoral strains near the distal femoral stem tip under load. The application of a strut allograft using stainless steel cables reduced strains on the femoral shaft, by as much as 59%. The findings of this laboratory study support the load-carrying capacity of the strut allograft. This finding has potential implications for reducing enigmatic thigh pain, as well as applications in revision hip arthroplasty and in the treatment of bone loss. Strut allografts have been used clinically to avoid revision of well-fixed stems, for the treatment of unremitting thigh pain [4,5], and also in the treatment of periprosthetic fractures [1-3].

In the present biomechanical study, cadaveric femurs were loaded under 500 N axial load combined with $\pm$10 Nm internal and external rotational torsion. The axial load produced bending on the shaft and tensile strains on the lateral cortex. This axial load was applied parallel to the femoral shaft to maximize the load's bending moment on the femur. The magnitude of torsion, $\pm$10 Nm, more closely approximated the physiological range applied during normal gait, 12 to 18 Nm [32]. Torsional loads produce shear strains longitudinally and transversely, as observed in our study. Both axial loads and torsional loads were intentionally kept sub-physiological in order to avoid cadaveric femur fractures in this laboratory model.

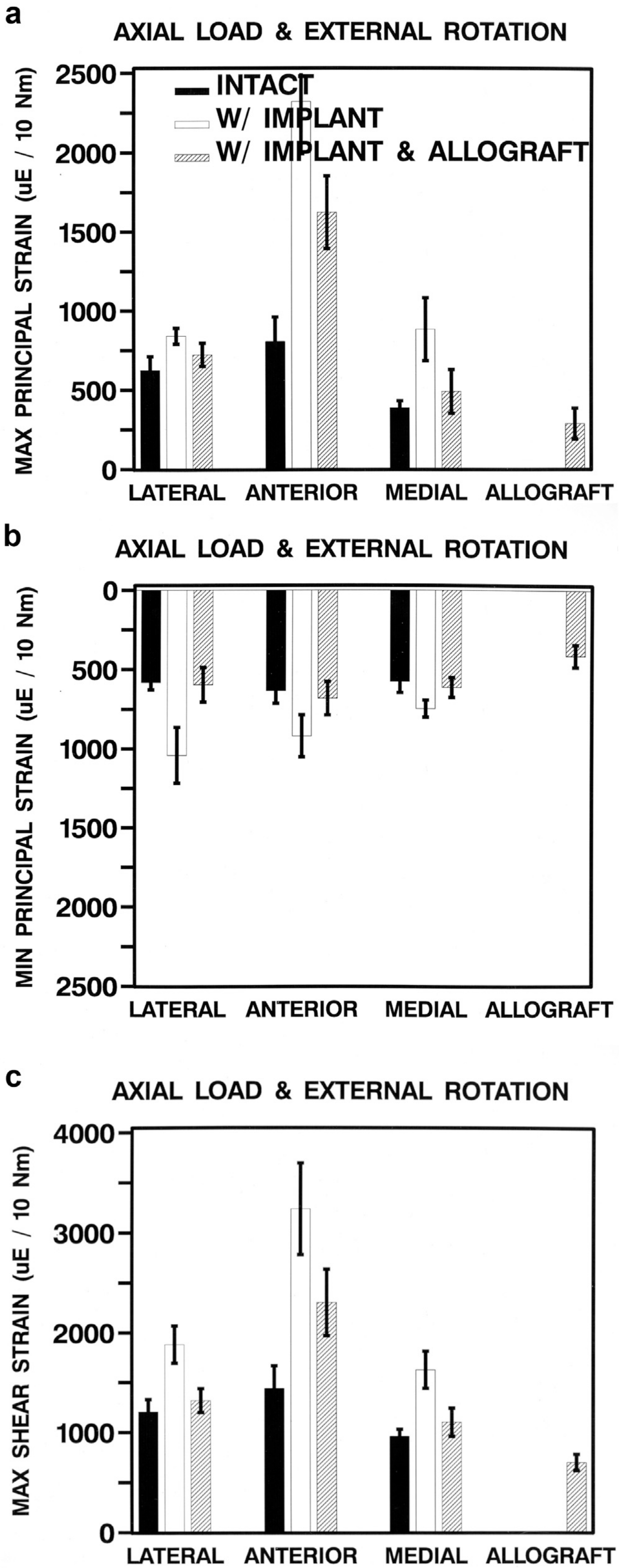


Figure 2. Principal and shear strains in intact specimens, implanted specimens, and implanted specimens with allografts, under combined axial load of 500 N and external rotation. Maximum principal strains were all tensile, and minimum principal strains were all compressive. (a) Maximum principal strain with axial load and external rotation; (b) minimum principal strain with axial load and external rotation; (c) maximum shear with axial load and external rotation.

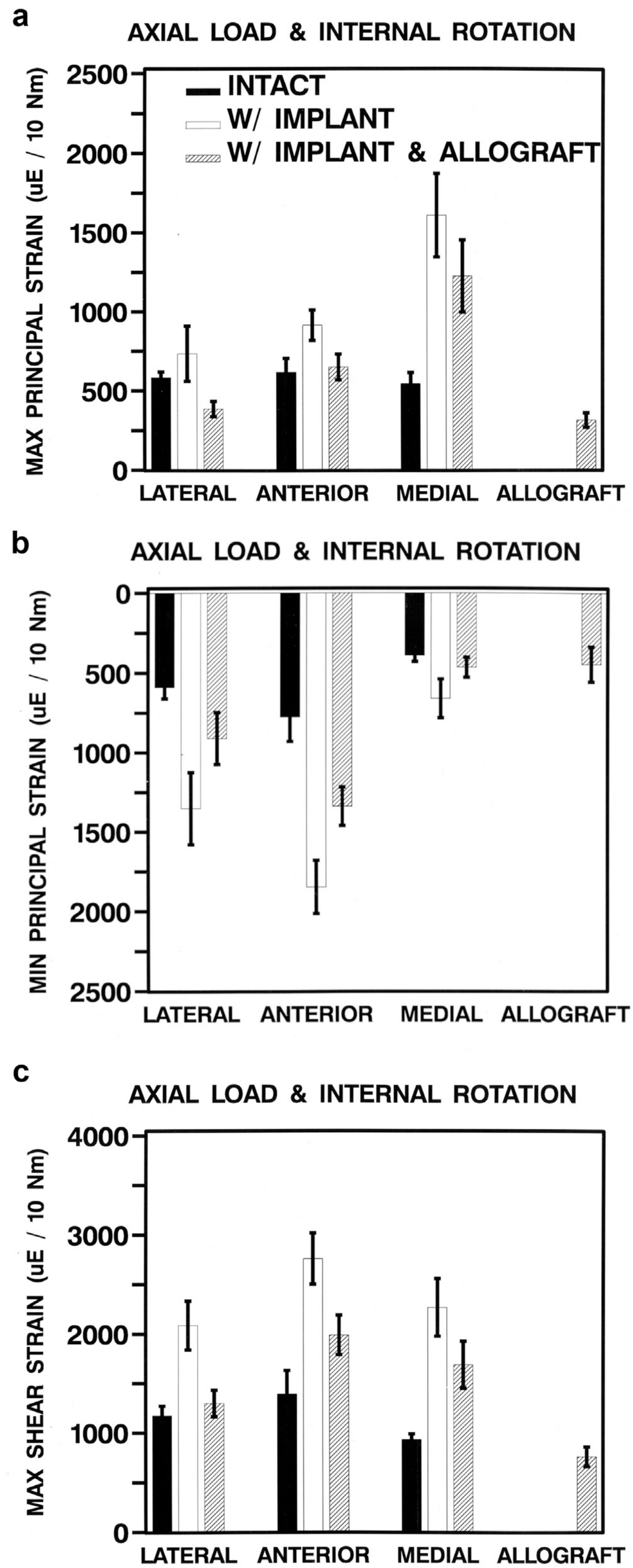


Figure 3. Principal and shear strains in intact specimens, implanted specimens, and implanted specimens with allografts under combined axial load of 500 N and internal rotation. Maximum principal strains were all tensile, and minimum principal strains were all compressive. (a) Maximum principal strain with axial load and internal rotation; (b) minimum principal strain with axial load and internal rotation; (c) maximum shear with axial load and internal rotation.

The risk of fracture in this type of biomechanical model is high due to the absence of the supporting soft tissue and muscle forces. Since strain is approximately a linear function of load until failure, the measured values were extrapolated to estimate physiological strains.

Accordingly, tensile and compressive strains measured on the femoral shaft in the present biomechanical study were well below strains required for failure of cortical bone in the femoral shaft. Specifically, mean tensile strains (estimated for 1000 N axial load) for intact femurs were well below 1500 $\mu\varepsilon$, increasing to a mean strain below 2500 $\mu\varepsilon$ with an implant under axial load combined with torsion. Even considering a more physiological axial load, the strains would still not approach the failure strength of femoral cortical bone, reported as 10,700 $\mu\varepsilon$ for older patients and 13,200 $\mu\varepsilon$ for younger patients [33]. Presuming a linear progression of strain with axial load, physiological axial load with an implant would not produce sufficient strain in this simulated model to lead to failure even with repetitive loading.

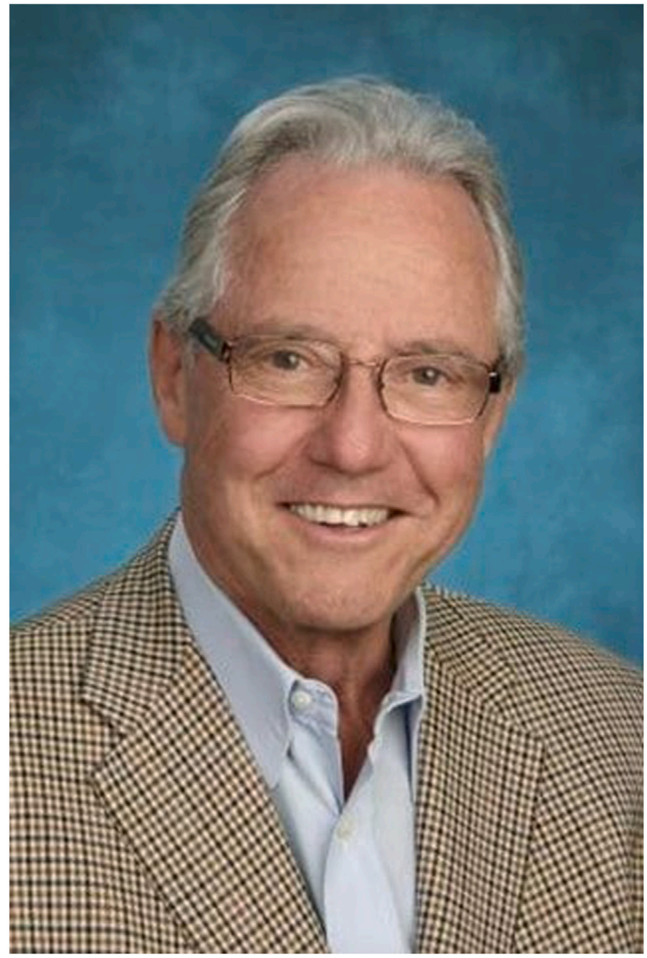
On the other hand, shear strains in the present study increased with an implant under axial load and external rotation to a mean of 3239 $\mu\varepsilon$, corresponding to 94% of longitudinal shear strength reported by Turner et al. for patients aged 63 to 83 years [34]. Shear strains measured in the present simulated model are consistent with an increased risk of bone failure, potentially leading to pathologic periprosthetic fractures, particularly with repetitive loading. Although this was not the focus of the present study, it may represent an important finding with implications for predicting implant periprosthetic fractures in older patients and should be potentially considered in future implant designs.

Numerous femoral stem designs have been introduced with the goal of reducing both proximal strain shielding and strain at the bone-implant interface, particularly in the diaphysis. In general, tapered femoral stems are intended to reduce femoral strains near the distal stem tip. However, cylindrical femoral stems, such as the one used in the present study, have been shown to produce large cortical strains near the stem tip, [16,35,36] likely contributing to thigh pain reported more commonly with this type of implant [7,9,23,24,26,31,37-43]. The results of the present simulated cadaveric model suggest that the use of a strut allograft with stainless steel cables may substantially reduce the strains on all aspects of the femur around the stem tip. In our study, strains were reduced by as much as 59% by the application of a lateral strut allograft, such that they approached strains in the intact femur.

In all simulated loading conditions of the present cadaveric femurs, the strains in the anterior, medial, and lateral cortices were reduced by the application of the femoral allograft, especially under torsional loads, suggesting that this technique can mechanically increase the apparent structural stiffness of bone and that the use of the tensioned stainless steel cables allows load transfer to bone. A possible reason for greater reduction under torsional loads may be an improved interlock of the bone to the allograft under this condition, with improved load transmission to the allograft. Greater reductions in torsional strains may be possible when using the allograft in more osteoporotic bone [44]. However, the number of specimens in this biomechanical study is too small to draw definite conclusions.

Although reports of enigmatic thigh pain have decreased in recent literature, [5,6,9,14,15,20,42,45-48] more than 500,000 total hip replacements are being performed each year in a younger population with increased life expectancy [49-51]. In the United States population, these are predominantly noncemented femoral components. In this setting, even a small percentage of patients with thigh pain represent a significant clinical problem.

The technique used by Dr. Anthony Hedley, with a lateral strut allograft cabled to the femur, addressed the theory that thigh pain



results from a difference in structural stiffness of the femur relative to the stem. With this technique, the sectional modulus (or flexural rigidity) of the bone is increased by the allograft, thereby better matching structural stiffness of the bone to the stem.

This study was not without limitations. The laboratory cadaver biomechanical model in this study used a simplified loading protocol and did not incorporate abductor or extensor muscles or other physiological forces acting on the femur. Other limitations include a small sample size, simulated ingrowth of the implant using cement, and an implant design that is no longer in use. Also, although this study measured reductions in strain with a cabled allograft in the femoral cortex and established that loads can be transferred to the allograft in vitro, it did not address the intermediate and long-term fate of the allograft in vivo. The model also did not address micromotion of the stem tip relative to the cortical shaft, as this may also be related to thigh pain. The model tested the initial condition of the cables after tightening, but the tension may change as the patient begins to mobilize and bear weight. Previous studies have reported gradual incorporation of cortical strut allografts, ranging from 80% to 94.7% [3,52-54].

If the allograft bone follows Wolff's law, it should remodel over time as it is experiencing strain in the loaded femur, at least in its initial phases. Further clinical studies may be required to quantify this potential mechanism. Should the theory of discrepancy in the flexural rigidity of the stem as compared with the bone indeed be a source of thigh pain, then the technique used in the present study has the potential to increase the flexural rigidity of bone in vivo. Within the limitations of the biomechanical simulation in this study, the technique described may provide a useful method to treat thigh pain without the disadvantages of removing a well-fixed femoral component. The data generated in this study raise concerns about risks of periprosthetic fractures, which should be further investigated with current stem designs and materials.

Acknowledgments

The authors would like to recognize the contributions of Dr. Anthony (Tony) Hedley (1943-2021) to the inception and development of this paper and surgical technique. His untimely passing prior to the final iteration of the manuscript prevented his inclusion as an author. Dr. Hedley was a master surgeon and devoted educator. His contributions to orthopedics, joint reconstruction, and implant design have been recognized in numerous awards and in the adulation of his patients and fellows. His motto to "always do my best" inspired all those with whom he worked.

Funding

This work was supported by the Doctors Education Research Fund (DERF) of the Los Angeles Orthopaedic Institute for Children Foundation (Formerly Los Angeles Orthopaedic Hospital Foundation). The implants used in this study were provided by Stryker (Kalamazoo, MI).

Conflicts of interest

The authors declare the following financial interests/personal relationships which may be considered as potential competing interests: E. Ebramzadeh receives research support as a principal investigator from DePuy Orthopaedics, Monogram Orthopaedics, Spinal Kinetics, and MicroPort Orthopaedics; is in the editorial board of the Journal of Bone and Joint Surgery, the Journal of Orthopaedic Trauma, and the Journal of Applied Biomaterials and Functional Materials; and is a board member in Hip Society. K. P. Sharpe is in the speakers' bureau of or gave paid presentations for and is a paid consultant for Conformis and Baudax Bio and receives research support as a principal investigator from Stryker, Cingal, and Rom Tech.

For full disclosure statements refer to https://doi.org/10.1016/j.artd.2022.02.010.

References

[1] Li D, Hu Q, Kang P, et al. Reconstructed the bone stock after femoral bone loss in Vancouver B3 periprosthetic femoral fractures using cortical strut allograft and impacted cancellous allograft. Int Orthop 2018;42(12):2787.
[2] Moore RE, Baldwin K, Austin MS, Mehta S. A systematic review of open reduction and internal fixation of periprosthetic femur fractures with or without allograft strut, cerclage, and locked plates. J Arthroplasty 2014;29(5):872.
[3] Pak JH, Paprosky WG, Jablonsky WS, Lawrence JM. Femoral strut allografts in cementless revision total hip arthroplasty. Clin Orthop Relat Res 1993;295:172.
[4] Domb B, Hostin E, Mont MA, Hungerford DS. Cortical strut grafting for enigmatic thigh pain following total hip arthroplasty. Orthopedics 2000;23(1):21.
[5] Granger L, Bankes M, Sandiford NA. Cortical strut graft for enigmatic thigh pain in uncemented total hip replacement. Cureus 2020;12(5):e8233.
[6] Amendola RL, Goetz DD, Liu SS, Callaghan JJ. Two- to 4-year followup of a short stem THA construct: excellent fixation, thigh pain a concern. Clin Orthop Relat Res 2017;475(2):375.
[7] Barrack R, M J, Bragdon C, Haire T, Harris W. Thigh pain despite bone ingrowth into uncemented femoral stems. J Bone Joint Surg Br 1992;74-B(4):507.
[8] Brown TE, Larson B, Shen F, Moskal JT. Thigh pain after cementless total hip arthroplasty: evaluation and management. J Am Acad Orthop Surg 2002;10(6):385.
[9] Bourne R, Rorabeck C, Ghazal M, Lee M. Pain in the thigh following total hip replacement with a porous-coated anatomic prosthesis for osteoarthrosis. A five-year follow-up study. J Bone Joint Surg Am 1994;76-A(10):1464.
[10] Fumero S, Dettoni A, Gallinardo M, Crova M. Thigh pin in cementless hip replacement. Ital J Orthop Traumatol 1992;18(2):167.
[11] Hozack W, Rothman R, Eng K, Mesa J. Primary cementless hip arthroplasty with a titanium plasma sprayed prosthesis. Clin Orthop Relat Res 1996;333:217.
[12] Lins R, Barnes B, Callaghan J, Mair S, McCollum D. Evaluation of uncemented total hip arthroplasty in patients with avascular necrosis of the femoral head. Clin Orthop Relat Res 1993;297:168.
[13] Moskal J, Shaffrey C, Ripley L. Prospective analysis of uncemented and hybrid primary porous coated anatomic total hip arthroplasties in a community setting. Clin Orthop Relat Res 1994;304:139.
[14] Gielis WP, van Oldenrijk J, Ten Cate N, Scholtes VAB, Geerdink CH, Poolman RW. Increased persistent mid-thigh pain after short-stem compared with wedge-shaped straight-stem uncemented total hip arthroplasty at medium-term follow-up: a randomized double-blinded cross-sectional study. J Arthroplasty 2019;34(5):912.
[15] Nam D, Nunley RM, Sauber TJ, Johnson SR, Brooks PJ, Barrack RL. Incidence and location of pain in young, active patients following hip arthroplasty. J Arthroplasty 2015;30(11):1971.
[16] Herzwurm PJ, Simpson SL, Duffin S, Oswald SG, Ebert FR. Thigh pain and total hip arthroplasty: scintigraphy with 2.5-year followup. Clin Orthop Relat Res 1997;336:156.
[17] Amstutz H, Nasser S, More R, Kabo J. The anthropometric total hip femoral prosthesis. Clin Orthop Relat Res 1989;242:104.
[18] Kim Y-H, Kim V. Uncemented porous-coated anatomic total hip replacement. J Bone Joint Surg Br 1993;75-B(1):6.
[19] Sumner D, Galante J. Determinants of stress shielding. Design versus materials versus interface. Clin Orthop Relat Res 1992;274:202.
[20] Bourne R, Rorabeck C, Burkart B, Kirk P. Ingrowth surfaces: plasma spray coating to titanium alloy hip replacements. Clin Orthop Relat Res 1994;298:37.
[21] Berzins A, Sumner D, Andriacchi T, Galante J. Stem curvature and load angle influence the initial relative bone implant motion of cemented femoral stems. J Orthop Res 1993;11(5):758.
[22] Cameron H. The 3-6 year results of a modular noncemented low-bending stiffness hip implant. J Arthroplasty 1993;8(3):239.
[23] Hedley A, Gruen T, Borden L, Hungerford D, E H, Kenna R. Two-year follow-up of the PCA noncemented total hip replacement. In: The hip: proceedings of the 14th meeting of the hip society, 225. St. Louis: CV Mosby; 1987.
[24] Maistrelli G, Fornasier V, Binnington A, McKenzie K, Sessa V, Harrington I. Effect of stem modulus in a total hip arthroplasty model. J Bone Joint Surg Br 1991;73-B(1):43.
[25] Schmidt J, Hackenbroch M. The cenos hollow stem in total hip arthroplasty: first experiences in a prospective study. Arch Orthop Trauma Surg 1994;113:117.
[26] Vresilovic E, Hozack W, Rothman R. Incidence of thigh pain after uncemented total hip arthroplasty as a function of femoral stem size. J Arthroplasty 1996;11(3):304.
[27] Sarmiento A. Austin more prosthesis in the arthritic hip. Experiences with 224 patients. Clin Orthop Relat Res 1972;82:14.
[28] Morscher E. Cementless total hip arthroplasty. Clin Orthop Relat Res 1983;181:76.
[29] Cilla M, Checa S, Duda GN. Strain shielding inspired re-design of proximal femoral stems for total hip arthroplasty. J Orthop Res 2017;35(11):2534.
[30] Arabnejad S, Johnston B, Tanzer M, Pasini D. Fully porous 3D printed titanium femoral stem to reduce stress-shielding following total hip arthroplasty. J Orthop Res 2017;35(8):1774.
[31] Callaghan JJ, Dysart SH, Savory CG. The uncemented porous-coated anatomic total hip prosthesis. Two-year results of a prospective consecutive series. J Bone Joint Surg Am 1988;70(3):337.
[32] Bergmann G, Deuretzbacher G, Heller M, et al. Hip contact forces and gait patterns from routine activities. J Biomech 2001;34(7):859.
[33] Evans FG. Mechanical properties and histology of cortical bone from younger and older men. Anat Rec 1976;185(1):1.
[34] Turner CH, Wang T, Burr DB. Shear strength and fatigue properties of human cortical bone determined from pure shear tests. Calcif Tissue Int 2001;69(6):373.
[35] Callaghan JJ, Van Nostrand D, Dysart SH, Savory CG, Hopkins WJ. Prospective serial technetium diphosphonate and indium-111 white blood cell labeled imaging in primary uncemented total hip arthroplasty. Iowa Orthop J 1996;16:104.
[36] Kandel L, Kligman M, Sekel R. Distal femoral stem tip resection for thigh pain complicating uncemented total hip arthroplasty. Five patients followed up for 6-10 years. Hip Int 2006;16(3):210.
[37] Engh C, Bobyn J, Glassman A. Porous-coated hip replacement. The factors governing bone ingrowth, stress shielding, and clinical results. J Bone Joint Surg Br 1987;69-B(1):45.
[38] Stulberg S, Stulberg B, Wixson R. The rationale, design characteristics, and preliminary results of primary custom total hip prosthesis. Clin Orthop Relat Res 1989;249:79.
[39] Cheal E, Spector M, Hayes W. Role of loads and prosthesis material properties on the mechanics of the proximal femur after total hip arthroplasty. J Orthop Res 1992;10(3):405.
[40] Burkart B, Bourne R, Rorabeck C, Kirk P. Thigh pain in cementless total hip arthroplasty. A comparison of two systems at 2-years follow-up. Orthop Clin North Am 1993;24(4):645.

[41] Kinov P, Radl R, Zacherl M, Leithner A, Windhager R. Correlation between thigh pain and radiological findings with a proximally porous-coated stem. Acta Orthop Belg 2007;73(5):618.
[42] Forster-Horvath C, Egloff C, Valderrabano V, Nowakowski AM. The painful primary hip replacement - review of the literature. Swiss Med Wkly 2014;8(144):w13974.
[43] Yu H, Liu H, Jia M, Hu Y, Zhang Y. A comparison of a short versus a conventional femoral cementless stem in total hip arthroplasty in patients 70 years and older. J Orthop Surg Res 2016;11(33):016.
[44] Dorr L, Faugere M, Mackel A, Gruen T, Bognar B, Malluche H. Structural and cellular assessment of bone quality of proximal femur. Bone 1993;14(3):231.
[45] Goetz DD, Reddy A, Callaghan JJ, Hennessy DW, Bedard NA, Liu SS. Four- to six-year follow-up of primary THA using contemporary titanium tapered stems. Orthopedics 2013;36(12):01477447.
[46] Jo WL, Lee YK, Ha YC, Park MS, Lyu SH, Koo KH. Frequency, developing time, intensity, duration, and functional score of thigh pain after cementless total hip arthroplasty. J Arthroplasty 2016;31(6):1279.
[47] Cinotti G, Della Rocca A, Sessa P, Ripani FR, Giannicola G. Thigh pain, subsidence and survival using a short cementless femoral stem with pure metaphyseal fixation at minimum 9-year follow-up. Orthop Traumatol Surg Res 2013;99(1):30.
[48] Baert IAC, Lluch E, Van Glabbeek F, et al. Short stem total hip arthroplasty: potential explanations for persistent post-surgical thigh pain. Med Hypotheses 2017;107:45.
[49] Inacio MCS, Paxton EW, Graves SE, Namba RS, Nemes S. Projected increase in total knee arthroplasty in the United States - an alternative projection model. Osteoarthritis Cartilage 2017;25(11):1797.
[50] Kurtz SM, Ong KL, Lau E, Bozic KJ. Impact of the economic downturn on total joint replacement demand in the United States: updated projections to 2021. J Bone Joint Surg Am 2014;96(8):624.
[51] Sloan M, Premkumar A, Sheth NP. Projected volume of primary total joint arthroplasty in the U.S., 2014 to 2030. J Bone Joint Surg Am 2018;100(17):1455.
[52] Head WC, Emerson RH, Cuellar AD. Cortical strut allografts for femoral reconstruction in revision hip arthroplasty. Semin Arthroplasty 1993;4(1):9.
[53] Park JS, Moon KH. Medium- to long-term results of strut allografts treating periprosthetic bone defects. Hip Pelvis 2018;30(1):23.
[54] Hamer AJ, Suvarna SK, Stockley I. Histologic evidence of cortical allograft bone incorportaion in revision hip surgery. J Arthroplasty 1997;12(7):785.