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A fatigue loading model for investigation of iatrogenic subtrochanteric fractures of the femur

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ABSTRACT

Background: Biomechanics of iatrogenic subtrochanteric femur fractures have been examined. Previously-described loading models employed monotonic loading on the femoral head, which is limited in emulating physiological features. We hypothesize that cyclic loading combined with the engagement of abductor forces will reliably cause iatrogenic subtrochanteric fractures.
Methods: Finite element analysis determined the effects of adding the abductor muscle forces to the hip contact force around holes located in the lateral femoral cortex. Finite element analysis predictions were validated by strain gage measurements using Sawbones™ femurs (Pacific Research Laboratories, Inc., Vashon, Washington, USA) with or without abductor muscle forces. The newly developed physiologically-relevant loading model was tested on cadaveric femurs ($N = 8$) under cyclic loading until failure.
Findings: Finite element analysis showed the addition of the abductor muscle forces increased the maximum surface cortical strain by 107% and the strain energy density by 332% at the lateral femoral cortex. Strain gages detected a 72.9% increase in lateral cortical strain using the combined loading model. The cyclic, combined loading led to subtrochanteric fractures through the drill hole in all cadaveric femurs.
Interpretation: Finite element analysis simulations, strain gage measurements, and cyclic loading of fresh-frozen femurs indicate the inclusion of abductor forces increases the stress and strain at the proximal-lateral femoral cortex. Furthermore, a cyclic loading model that incorporates a hip contact force and abductor muscles force creates the clinically encountered subtrochanteric fractures *in vitro*. This physiologically-relevant loading model may be used to further study iatrogenic subtrochanteric femur fractures.

1. Introduction

An iatrogenic subtrochanteric fracture (ISF) may complicate operations that involve drilling or reaming the lateral femoral cortex, particularly internal fixation of femoral neck fractures (Canale et al., 1994; Fung et al., 2008; Jansen et al., 2010; Karr and Schwab, 1985; Pelet et al., 2003; Stronach et al., 2010). Approximately 3% of femoral neck fractures fixed with compression screws may be complicated by an ISF and require reoperation (Jansen et al., 2010). However, very little biomechanical data is available to support the validity of the recommendations for surgery in the context of reducing the incidence of ISFs. Sporadic reports of postoperative subtrochanteric fractures continue to emerge in recent literature, highlighting the need for more rigorous research to understand this complication (Yang et al., 2013). A reliable, physiologically- and clinically-relevant loading model is needed to better understand the biomechanics of ISFs.

Many ISFs may occur without trauma or with indirect trauma (Canale et al., 1994; Howard and Davies, 1982; Karr and Schwab, 1985; Kloen et al., 2003; Neumann, 1990; Pelet et al., 2003; Stronach et al., 2010). The few biomechanical experiments that evaluate the association between surgical constructs and ISFs do so in the context of monotonic loading, which would be representative of an isolated, high-energy event (Litchblau et al., 2008; Oakey et al., 2006). Many ISFs occur less than six weeks after surgery, but they have been noted to occur up to two years post-operatively (Andrew and Thorogood, 1984; Blyme et al., 1992; Howard and Davies, 1982; Jansen et al., 2010; Karr and Schwab, 1985; Kloen et al., 2003; MacEachern et al., 1984; Neumann, 1990; Pelet et al., 2003). The non-traumatic nature of approximately half of the documented fractures and the delayed time to fracture after the initial operation suggests that fatigue failure is a relevant and important factor to the etiology of ISFs that has been neglected by prior studies.

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Therefore, cyclic loading is probably more representative of conditions surrounding ISFs than monotonic loading.

In addition to the absence of fatigue models to experimentally evaluate ISFs, previous experiments also test femur construct stability and strength with only a single load on the femoral head (Booth et al., 1998; Litchblau et al., 2008; Oakey et al., 2006). Previous *in vitro* studies have simulated the force of the abductor muscles (Cristofolini, 1997); however, they did so to evaluate the stresses in the proximal femur rather than to study subtrochanteric fractures. Based on finite element analysis (FEA), it has been demonstrated that the addition of the abductor muscles on the greater trochanter significantly alters the stresses and strains in the proximal femur during all phases of gait (Stolk et al., 2001). The combined influence of the adductor muscles and iliotibial band on a reconstructed hip was calculated by FEA and found to be small during the mid-stance phase of gait (Stolk et al., 2001). Abductor muscle forces have been concluded to be the main muscle forces affecting the cortical stresses and strains of the intact proximal femur and should be included when studying the biomechanics of the proximal femur (Cristofolini, 1997; Rohlmann et al., 1982).

There is a paucity of biomechanical loading models which faithfully emulate the failure in the subtrochanteric region in association with screw holes. To the best of our knowledge, no biomechanical test model simulated the combined loads of the joint contact force and abductor forces in the context of fatigue to reproduce subtrochanteric fractures. This study aimed to determine whether subtrochanteric fractures can be reliably induced in a fatigue loading model where the abductor muscle forces are emulated in addition to the hip joint contact force.

2. Methods

2.1. Finite element analysis

FEA (Solidworks, Dassault Systèmes, Waltham, MA USA) was performed to contrast strains around the screw hole for these loading scenarios:

- Case A: loading on the head only (Oakey et al., 2006),
- Case B: combined loading at the femoral head (simulating the body weight) and the greater trochanter (emulating the hip-stabilizing force of the abductor muscles) during *in vitro* biomechanical loading performed in the lab (Cristofolini, 1997; Rohlmann et al., 1982; Stolk et al., 2001), and
- Case C: physiological loading previously described for single-legged stance, which includes a load at the joint surface representing body weight and the force at the greater trochanter representing the gluteus medius and minimus (Lotz et al., 1995).

Comparison of Case A and Case B evaluated the contribution of the abductor muscle to the stress state around the hole. The abductor muscles impart compression and shear at the femoral head through its action on the pelvis but cause tension at its insertion and at the proximal, lateral cortex, which could contribute to subtrochanteric fractures through drill holes. Case C served as a physiological reference point to which the convergence of Case B is investigated.

The finite element model based on CT-scan slices of a Sawbones™ femur created by Marcello Papini was obtained online from the BEL Repository, managed by the Instituti Ortepedici Rizzoli, Bologna, Italy. The model contained 95,506 nodes and 65,405 elements. Mesh convergence was performed to obtain a convergence within 3% for maximum strain and von Mises stress in the region of interest. On the femoral model, the shaft extended 15 cm below the lesser trochanter, and the distal 3 cm was fixed (Oakey et al., 2006). The three cases were simulated assuming a 75 kg adult (Fig. 1A–C), and the force on the femoral head was consistent between all cases. In Case A, the head-only loading, a 2500 N compressive force (approximately 350% of body weight for a 75 kg person) on the femoral head was used to mimic the joint contact force seen during fast walking *in vivo* (files jb5103b and rhr0192b from database OrthoLoad, Bergmann, 2008; Lengsfeld et al., 2005). The force was applied on the joint surface and directed through the center of the head along the mechanical axis of the femur, which was assumed to be offset 7° from the anatomic axis of the femur. Case B, representing *in vitro* biomechanical testing conditions, was similar to Case A, but with the addition of a 1785 N force at the inferior margin of the greater trochanter, which was applied to simulate an abductor strap used by the combined loading model. The force was directed upward and parallel to the joint contact force, resulting in a moment and tensile stress at the lateral cortex. Loads for Case C were adopted from previously-described physiologic single-legged stance loading (LeVeau and Williams, 1992; Lotz et al., 1995). Loads were applied as point forces except in Case B, where the abductor force was distributed over a line to simulate the hose clamp contact that was employed in the ensuing cadaveric femur fatigue model. A 7.1 mm circular defect directed toward the femoral head, consistent with the cannulated screw size used for hip fracture fixation, was created through the lateral femoral cortex centered 8.25 mm superior to the middle of the lesser trochanter. Properties for cortical bone (E = 16.4 GPa, $\nu = 0.26$) and trabecular bone (E = 1.37 GPa, $\nu = 0.3$) were assumed to be linear elastic and isotropic (Lotz et al., 1990; Oonishi et al., 1983; Reilly and Burstein, 1975). The model predicted the surface strain at the cortex (ε_C) in the inferior–superior direction 2 mm posterior to the drill hole. This direction was chosen because it closely matched the location and orientation of the unidirectional strain gage measurements in the ensuing experiments. The FEA model was also used to find the strain energy density (SED) in the region posterior to the hole defect.

2.2. Mechanical testing apparatus

All mechanical tests were conducted using an Instron 8501M (Instron, Norwood, MA USA). Custom aluminum fixtures were used for applying loads and fixing the femur specimens in position (Fig. 2). The actuator fitted with a 1000 lb cell (model 1010AF-1K-B, Interface Mfg., Scottsdale, Arizona, USA) applied a vertical load on a crosshead, transmitting the force to the near end of an adjustable aluminum bar measuring 1 cm × 5 cm × 29 cm. The bar's end distal to the actuator was used to secure a polyester strap, the 'abductor strap', which itself was secured to the proximal femur by a padded hose clamp placed around the intertrochanteric region. The padding of the edge of the hose clamp was achieved by ten layers of aluminum foil to distribute the load more evenly and to counter notching of the clamp into test specimens. The hose clamp was lightly fastened around the femur just enough to prevent the assembly from slipping superiorly over the inferior border of the greater trochanter during loading. Otherwise, except the engagement at the inferior border of the greater trochanter the clamp was loosely applied around the periphery of the proximal femur. A thick-walled cylindrical aluminum clamp was bolted to a base-plate with the cylinder angled 7° from vertical. The cylinder secured the potted, distal end of the femoral specimen, fixing the femoral shaft along its axis. A custom bearing molded from polymethylmethacrylate (PMMA, Henry Schein, Melville, NY USA) was positioned on the undersurface of the aluminum bar and articulated with the femoral head. Since all fixtures were adjustable, the base-plate and bar were adjusted so that the ratio of the length of the bar from the crosshead to the center of the femoral head was 2.5 times the length of the bar from the center of the femoral head to the abductor strap. By static analysis, the force of the abductor strap was estimated to be 71.4% of the joint contact force, which is similar to loads used previously in literature to mimic normal forces around the hip (Lengsfeld et al., 2005; Rohlmann et al., 1982).

2.3. Strain measurements from the synthetic femur

A fourth-generation large Sawbones™ femur (Pacific Research Laboratories, Inc., Vashon, Washington, USA) was cut 15 cm distal to the