Patient specific quantitative analysis of fracture fixation in the proximal femur implementing principal strain ratios. Method and experimental validation

Eran Peleg a,1,*, Maarten Beek b,2, Leo Joskowicz c,3, Meir Liebergall d,4, Rami Mosheiff d,1,4, Cari Whyne b,1,5

a Department of Biomedical Engineering, Hadassah University Medical Centre, Jerusalem, P.O.B. 12000, Israel
b Orthopedic Biomechanical Laboratory, Sunnybrook Science Centre, Toronto, Canada M4N 3M5
c School of Engineering and Computer Science, The Hebrew University of Jerusalem, Jerusalem 91904, Israel
d Department of Orthopedic Surgery, Hadassah University Medical Centre, Jerusalem, P.O.B. 12000, Israel

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Abstract

Computational patient-specific modeling has the potential to yield powerful information for selection and planning of fracture treatments if it can be developed to yield results that are rapid, focused and coherent from a clinical perspective. In this study we introduce the utilization of a principal strain fixation ratio measure (SR) defined as the ratio of principal strains that develop in a fixated bone relative to the principal strains that develop in the same bone in an intact state. The SR field output variable is theoretically independent of load amplitude and also has a direct clinical interpretation with SR \( \frac{\sigma_0}{\sigma_a} \) representing stress shielding and SR \( \frac{\sigma_4}{\sigma_1} + b \) representing overstressed bone. A combined experimental and numerical study was performed with cadaveric proximal femora (\( n = 6 \)) intact and following fracture fixation to quantify the performance of the SR variable in terms of accuracy and sensitivity to uncertainties in density–elasticity relationships and load amplitude as model input variables. For a given axial compressive force the SR field output variable was found to be less sensitive to changes in density–elasticity relationships and the response function to be more accurate than strain values themselves; errors were reduced by 44\% on comparing SR with strain in the fixated model. In addition, the experimental data confirmed the assumption that the SR values behave independent of load amplitude. The load independent behavior of SR and its direct clinical interpretation may ultimately provide an appropriate and easily understood comparative computational measure to choose between patient specific fracture fixation alternatives.

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1. Introduction

Despite its potential to assist in quantifying fracture fixation, computational patient-specific modeling for selection and planning of fracture treatments is limited at present. Patient specific modeling is regarded as a time consuming process with little effect on patient treatment and outcome.

Extensive work has been reported regarding patient specific finite element (FE) based modeling. Clinical applications have been predominantly aimed at failure prediction of intact bone models under static loading (Bessho et al., 2007; Cody et al., 1999; Keyak, 2001; Schileo et al., 2008) and biomechanical responses to total hip implant insertion (Huiskes et al., 1987; Pettersen et al., 2009; van Rietbergen and Huiskes, 2001). Numerous studies have reported on the development, validation and automation of patient specific FE modeling techniques from quantitative CT data sets (Helgason et al., 2008; Keyak et al., 1993; Schileo et al., 2007; Taddei et al., 2004, 2006; Zannoni et al., 1998), however, these investigations primarily focused on strategies for element based application of bony material properties at different densities based on Young's modulus relationships for intact bone models.

A patient specific quantitative process that can be applied in a true clinical environment must cope with profound uncertainties. Previous studies have identified uncertainties related to material property assignments and surface geometry. (Keyak and Falkinstein, 2003; Taddei et al., 2006; Weins et al., 2000). Beyond these issues, the most apparent uncertainty is related to the fact...
that in-vivo load amplitudes and directions are not well defined, as such load characteristics including gait patterns are usually roughly estimated. Additional error is expected from uncertainties in interface interactions between implant surfaces. Patient specific FE based studies that involve fracture fixation resembling that of a true clinical workflow are not reported despite the clinical importance of this matter (Audige et al., 2005; Baumgaertner et al., 1998; Davis et al., 1990), perhaps due in part to the multiple uncertainties inherent in the analyses.

To address these uncertainties, we introduce the utilization of a principal strain fixation ratio measure (SR). SR is defined as the ratio of principal strains that develop in a fixated bone relative to the principal strains that develop in the same bone in an intact state. By definition, the SR measure is independent of load amplitude as long as a linear response is assumed and modeled. The SR field output variable may provide the ability to view strain shielding (SR < 1 – a) and overstrained patterns (SR > 1 + b) at the time of fixation and as such may be an indicator for bone adaptation (Cowin et al., 1985; Currey, 2003; Lanyon, 1987; van der Meulen and Huiskes, 2002) and failure prediction (Schileo et al., 2003) providing a more appropriate comparative measure to assist in the selection of fixation alternatives.

The present study aims to examine the performance of the SR field output variable generated through patient specific FE analysis compared with experimentally measured SR values following proximal femur fracture fixation. It is hypothesized that for a given force, SR is more accurate and less sensitive to material property uncertainties than principal strain measures alone in quantifying the structural impact of proximal femur fracture fixation.

2. Methods and materials

Six fresh frozen cadaveric proximal femurs were obtained from the department of anatomy at the University of Toronto (Table 1). All the specimens were subjected to volumetric quantitative computed tomography (Lang et al., 1997) of the proximal region and fell in the range from normal to severe osteoporosis (Table 1). Prior to experimentation each specimen was thawed and the soft tissue was removed. The base of each femur was potted in PMMA such that the average length of the femoral neck (M1—rosette), medial shaft opposite to the plate (M2—rosette, M3—rosette), medial-distal region (M4—axial gauge) and lateral-distal region approximately 2 cm beneath the end of the plate (L—axial gauge). Under the loading utilized in this study, the directions of the principal strains were assumed empirically in certain locations (i.e. along the shaft) and confirmed in the FE modeling. As such, and due to a limited number of available channels of our data acquisition unit, axial gauges were used in these locations. The alignment of each axial gauge was assumed to coincide with the highest magnitude principal strain. The axial direction of the gauge was registered in the experimental phase and was accounted for during the simulation phase.

Prior to determining the locations of measurement, preliminary modeling was carried out with a CT data set of a cadaveric femur with an induced pertrochanteric fracture. SR patterns were examined and the locations that were found to be most affected by the fixation and eligible in terms of practicality for strain gauge attachment were chosen.

### Table 1

<table>
<thead>
<tr>
<th>Parameter</th>
<th>Value</th>
</tr>
</thead>
<tbody>
<tr>
<td>Sex</td>
<td>Females</td>
</tr>
<tr>
<td>Age (year)</td>
<td>56–83</td>
</tr>
<tr>
<td>Length (mm)</td>
<td>160–206</td>
</tr>
<tr>
<td>BMD (mg/cm³)</td>
<td>122–516</td>
</tr>
</tbody>
</table>

*a BMD was measured by volumetric quantitative CT applied to the trochanteric region.

Four fiducials were attached to each specimen and their locations registered to the strain gauges (Microscribe3D dx HCl 2.0). Spatial coordinates of each fiducial were later extracted from the CT data set and modeled against the respective spatial coordinates as measured during the experimental phase. A transformation matrix was created, enabling a rigid transformation of the real world coordinate system to that of the FE model created from the CT data set.

Load protocols were a prerequisite for the FE and model specimens and were performed in sequence without moving the tested specimen. A uniaxial force was applied (Bionix 858, MTS Systems Corporation) with the load amplitude gradually increased from 100 to 800 [N] in increments of 100 [N]. The load was held for each increment for 15 s while strain data were collected (Daqview DaqBook/2000 V7.13.2) at the frequency of 2 [Hz]. Between the load increments the crosshead speed was 0.1 mm/s. Gradually increasing the load amplitude to the range of a patient's body weight was utilized to represent a one legged stance and prevented mechanical damage to the specimens. (Pettersen et al., 2009; Schileo et al., 2007). The load protocol that was used to validate the predicted maximum and minimum principal strain values with the measured values was 400 [N]. This load amplitude prevented large changes in geometry as a result of bone implant crushing, which could have introduced non-linearity. All the load amplitudes were used to examine the independency of the SR with load amplitude.

Following experimental testing of the intact specimens, fractures were introduced and then fixated with a short extra-medullary plate and a static lag screw. The fracture fixation was performed by the authors following training by an experienced orthopaedic surgeon. All fixations were verified by the surgeon.

The distance between the tip of the lag screw and the femoral head apex was measured for modeling purposes. Only one side screw was used to reduce uncertainties related to the load sharing of a multiple screw system. Based on the OTA classification, the fractures ranged from type pertrochanteric simple (31-A1) to subcapital non-impacted (31-B3).

Following mechanical testing, hardware and strain gauges were removed and each fractured femur was CT scanned (Light-speed VCT—GE Medical Systems) at a slice thickness of 0.625 mm and pixel spacing 0.36 × 0.36 mm², together with a calibration phantom (Skyscan NV). Each CT data set was calibrated assuming a linear relationship between Hounsfield unit and bone ash density (Les et al., 1994). The apparent density was then calculated assuming a constant ratio \( \rho_{\text{app}}/\rho_{\text{ash}} = 0.6 \) (Schileo et al., 2008), Acquiring the CT of the fractured specimens resembled a clinical scenario in which the diagnostic stage is performed prior to a surgical procedure.

The CT data files and the files containing the geometric data of the side-plate (Richards et al., 2006) and side screws (prepared in-house) were imported to AmiraDevs5.2.2 (Visage Imaging, Inc.) for preparation of the models prior to the FE analysis. The following steps were performed: (1) the bone fragments were segmented, (2) the fracture pattern was CT scanned (Light-speed VCT—GE Medical Systems) and (3) material properties were assigned to each element as described by Taddei et al. (2007), (4) the fractures were reduced and implants positioned (Fig. 2), (5) boolean cutting was conducted between the implants and bone grids and finally (6) all the model information was exported to Abaqus6.9-1 (Dassault Systems, USA) for FE analysis.

Three different density-based elasticity relationships were used to describe the bone material properties: Eq. (1) \( E = 3750\rho_{\text{ash}} \) (Carter and Hayes, 1977); Eq (2) \( E = 6500\rho_{\text{ash}}^{0.6} \) (Morgan et al., 1994) and Eq (3) \( E = 10500\rho_{\text{ash}}^{1.25} \) (Keller, 1994). These relationships were chosen because they represent the best approximations for CT based determination of bone material properties as recently compared (Schileo et al., 2007).

Boundary and loading conditions simulated the experimental setup—the distal femur had a zero displacement condition and a spread force was applied to the femoral head in the direction of the distal end (Fig. 2). No contact was applied...
at the fracture surface between the femoral head and shaft (a constant gap prevailed during the experimental phase). The screw-thread region and the bottom portion of the DHS barrel were bonded to the bone. It was assumed that during the loading phase, force was not exerted between the top portion of the implant and the bone interface.

The contact condition between the screw head and plate seat was modeled as an exponential relationship of pressure to clearance between the contact surfaces ("soft contact"). This type of contact condition has the advantage of fast convergence and is appropriate to use if local tangential forces are not of interest.

A compatible intact model was created for each fixated model. Intact models were created by duplicating the fixated model, removing all implants and bonding the fracture surface between the femoral head and shaft. As such, the intact models were identical in bone grid topology and material property assignments to the fixated models. Based on convergence testing results, all the models were created with an average element size length less than 2 mm (Vicenzi et al., 2004). All simulations were carried out assuming linear elastic material properties and small geometrical displacements.

The calculated maximum and minimum principal strains at the surface nodes, corresponding to the sensing area of each strain gauge, were averaged and compared with the experimental measurements. SR values were calculated by dividing the principle strain values in each fixated model by the corresponding values in each intact model. The calculated FE-derived SR values were compared to the corresponding experimentally derived SR values. A linear regression between experimental and predicted strains was performed to quantify the prediction accuracy. Root mean square (RMSE) errors and the peak errors were calculated.

### 3. Results

The regression lines between the experimental data and the numerical calculations for the intact specimens exhibited high sensitivity to the different elasticity–density relationships (Fig. 3). The best fit was that related to models created with Eq. (2) (Table 2). The fit for the models created with Eqs. (1) and (3) created regression lines with higher slopes (1.58 and 2.2, respectively) and RMSE values (176µε and 373µε, respectively). These data are in close agreement with those reported by Schileo et al. (2007).

The regression lines for the fixated specimens exhibited less sensitivity to density–elasticity relationships than those of the intact models (Fig. 4). The overall range of the slopes was between 1.138 and 1.445. The maximum and average errors were also reduced relative to the intact models (Table 2). As a result of the intense stress shielding (SR < 0.0005) at the femoral neck region the strains generated at the M1 location were very small (below the signal to noise ratio) and therefore cannot be accounted for as part of the experimental data of the fixated models.

The correlation between the predicted SR values and the experimental values exhibited low sensitivity to the different density–elasticity relationships (Fig. 5). The slope values ranged between 0.63 and 0.76. In terms of $R^2$ the results were not significantly different from those of the fixated models (Table 2), however RMSE and the average and maximum errors of the SR exhibited an improved performance over the intact and fixated strain parameters.

For each specimen, SR values were measured for all the load increments and regressed for linear correlation (confidence of $p < 0.05$), with the respective load amplitudes. The range of slopes of the linear fits was between −0.022 to +0.029 (average 0.00099). In all of the specimens the SR values were independent of load amplitude.

### 4. Discussion

The aim of this study was to present and assess the performance of the SR field output variable for evaluating automated patient specific FE modeling of a fixated fractured

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### Table 2

Goodness of fit parameters for pooled data reported for all model types (intact, fixated strain measurements and strain ratio measurements) under the three different density–based elasticity relationships. Eq. (1) $E = 3790\mu_{\text{fy}}^2$ (Carter and Hayes, 1977), Eq. (2) $E = 6050\mu_{\text{fy}}^2$ (Morgan et al., 2003) and Eq. (3) $E = 10,500\mu_{\text{fy}}^2$ (Keller, 1994).

<table>
<thead>
<tr>
<th>Fit parameters</th>
<th>Intact models</th>
<th>Fixated models principal strains</th>
<th>Fixated models strain ratios (SR)</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Eq. (1)</td>
<td>Eq. (2)</td>
<td>Eq. (3)</td>
</tr>
<tr>
<td>$R^2$</td>
<td>0.85</td>
<td>0.81</td>
<td>0.86</td>
</tr>
<tr>
<td>Slope</td>
<td>1.59</td>
<td>1.1</td>
<td>2.2</td>
</tr>
<tr>
<td>RMSE</td>
<td>176 µε</td>
<td>25 µε</td>
<td>373 µε</td>
</tr>
<tr>
<td>Max. err. (%)</td>
<td>364</td>
<td>218</td>
<td>536</td>
</tr>
<tr>
<td>Average err. (%)</td>
<td>103</td>
<td>60</td>
<td>177</td>
</tr>
</tbody>
</table>
The SR values are independent of load amplitude. In addition, the experimental data confirmed the assumption that accurate strain measurements are more consistent with the FE calculated strain values of the intact model than the fixated strain data. This is because the density–elasticity relationships and the response function is more sensitive to changes in the FE SR field output variable than the results obtained in the intact specimen. The predicted strain values for the intact femur were obtained by dividing the computed FE strains in the fixated model by the results obtained in the intact specimen. The predicted strain values were obtained by dividing the computed FE strains in the fixated model by the results obtained in the intact specimen. The predicted strain values were obtained by dividing the computed FE strains in the fixated model by the results obtained in the intact specimen.

Further work can be initiated to look at additional loading geometrical movement were introduced to the fixated models, the SR value, which combined results from the intact and fixated models, maintained linearity as expressed in the goodness of fit parameters (Table 2).

Relatively high sensitivity of SR to different density–elasticity relationships was noticed at the distal lateral region of the femoral shaft a few centimeters below the end of the plate (Fig. 6) expressed in high gradients of SR values. Failure modes of fracture fixation at the distal region of the femur below the implant tip have been reported in clinical literature and are referred to as failures due to a ‘stress riser’ effect (DiMaio et al., 1992; Haynes et al., 1997); however, the limited literature on this issue has reported clinical evidence of this failure mode but has not provided quantitative explanation. Assessing the degree of a stress riser effect as a result of abnormal loading conditions is not possible utilizing the oft-used strain and stress measurements due to the fact that the patient specific in-vivo strain and stress values that develop in an intact or fixated bone structure are not known. The use of SR values ‘magnifies’ such effects since it is an innate attribute of SR to indicate abnormal changes in strain values independent of load amplitude.

The material mapping strategy used in this analysis (Taddie et al., 2007) has been implemented in various models and has been shown to provide accurate results in validation studies (Schileo et al., 2007). Recent studies have presented modified material mapping strategies allowing for spatial variation of material properties within individual elements and compensation for partial volume effects at surface elements (Chen et al., 2009; Helgason et al., 2008). These studies have shown that results are indeed affected by change in the material mapping strategy but have not proved that results from this new strategy are more accurate.

The well-assessed anisotropic behavior of bone tissue was not considered in the present work. However, recent work (Peng et al., 2006) has reported non-significant differences in the evaluation of femoral stress states under a single leg stance condition for isotropic vs orthotropic material property assignment.

Computational studies are reported on the effect of muscle loading on the simulation of bone remodeling in the proximal femur (Bitsakos et al., 2005; Weinans et al., 2000). In this study the main goal was to validate the predicted SR magnitude values with the measured SR values without relating to physiological functionality. For this, a simplified loading condition was utilized in order to reduce uncertainties related to physiological functionality. For this, a simplified loading condition was utilized in order to reduce uncertainties related to boundary conditions involved in a more complicated loading condition (Keyak et al., 1998, 2001; Schileo et al., 2007, 2008). Further work can be initiated to look at additional loading conditions utilizing the oft-used strain and stress measurements due to the fact that the patient specific in-vivo strain and stress values that develop in an intact or fixated bone structure are not known. The use of SR values ‘magnifies’ such effects since it is an innate attribute of SR to indicate abnormal changes in strain values independent of load amplitude.

The main finding of the present work is that for a given force the FE SR field output variable is less sensitive to changes in density–elasticity relationships and the response function is more sensitive to changes in the FE SR field output variable than the results reported by Schileo et al. (2007) thus verifying that the ‘fused’ bone structure responded similar to a model of an intact bone.

The main finding of the present work is that for a given force the FE SR field output variable is less sensitive to changes in density–elasticity relationships and the response function is more sensitive to changes in the FE SR field output variable. The predicted strain values for the intact models utilizing the different density–elasticity relationships were in close agreement with the results reported by Schileo et al. (2007) thus verifying that the ‘fused’ bone structure responded similar to a model of an intact bone.

The relatively low slope values of the predicted SR (Fig. 5) response function imply that the predicted SR values are lower than the measured values. This is to be accounted for if SR is to be used as a failure criterion. Although non-linear effects such as contact at the screw head and possibility for increased density–elasticity relationships and the response function is more sensitive to changes in the FE SR field output variable. The predicted strain values for the intact models utilizing the different density–elasticity relationships were in close agreement with the results reported by Schileo et al. (2007) thus verifying that the ‘fused’ bone structure responded similar to a model of an intact bone.
scenarios including more physiologic conditions (Heller et al., 2005).

A further limitation of the SR method presented herein is that it results in loss of information with respect to directionality (tension compression) of strain values. This information, however, can be accessed from the raw strain data and directionality added to the SR output.

The use of the SR criterion introduces some major advantages over the oft-used strain based criteria for interpretation of FE models of bone structures. The main advantage is that by definition the SR is independent of load amplitude. This is extremely important in a clinical environment, where physiologic loads are not well characterized. The relatively low sensitivity of the SR criterion to uncertainties in density-based elasticity relationships is also important as uncertainty remains in CT-based material property assignment strategies. These advantages of the SR parameter and its direct clinical interpretation (SR < 1 – a, stress shielding; SR > 1+b, overstressed) may ultimately provide a simple way to understand the clinically relevant, comparative measure to be chosen between fracture fixation alternatives based on patient specific FE modeling.

Conflict of interest

I am not aware of any conflict of interest situation that would inappropriately influence the work and research conducted to produce this manuscript titled ‘Patient specific quantitative analysis of fracture fixation in the proximal femur implementing principal strain ratios: method and experimental validation’.

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