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An Investigation of the Inelastic Behaviour of Trabecular Bone during the Press-fit Implantation of a Tibial Component in Total Knee Arthroplasty

Kelly, N.^{1,2}, Cawley, D.T.³, Shannon, F.J.³, McGarry, J.P.^{1,2}

1 Department of Mechanical and Biomedical Engineering, National University of Ireland, Galway, Ireland

2 National Centre for Biomedical Engineering Science, National University of Ireland, Galway, Ireland

3 Department of Orthopaedic Surgery, National University of Ireland, Galway, Ireland

Keywords: Total knee arthroplasty; press-fit implantation; finite element analysis; trabecular bone plasticity; polyurethane foam; pressure dependent yielding.

Abbreviations: total knee arthroplasty (TKA); polyurethane foam (PU); von Mises (VM); crushable foam with isotropic hardening (CFI); apparent Young's modulus (E); elastic Poisson's ratio (ν_e); apparent yield stress (σ_y); friction coefficient (μ); plastic Poisson's ratio (ν_p); compression yield stress ratio (K); crushable foam with volumetric hardening (CFV).

Abstract

The stress distribution and plastic deformation of peri-prosthetic trabecular bone during press-fit tibial component implantation in total knee arthroplasty is investigated using experimental and finite element techniques. It is revealed that the computed stress distribution, implantation force and plastic deformation in the trabecular bone is highly dependent on the plasticity formulation implemented. By incorporating pressure dependent yielding using a crushable foam plasticity formulation to simulate the trabecular bone during implantation, highly localised stress concentrations and plastic deformation are computed at the bone-implant interface. If the pressure dependent yield is neglected using a traditional von Mises plasticity formulation, a significantly different stress distribution and implantation force is computed in the peri-prosthetic trabecular bone. The results of the study highlight the importance of: (i) simulating the insertion process of press-fit stem implantation; (ii) implementing a pressure dependent plasticity formulation, such as the crushable foam plasticity formulation, for the trabecular bone; (iii) incorporating friction at the implant-bone interface during stem insertion. Simulation of the press-fit implantation process with an appropriate pressure dependent plasticity formulation should be implemented in the design and assessment of arthroplasty prostheses.

1. Introduction

Aseptic loosening is the most common cause of total hip and knee arthroplasty revision surgery, accounting for 16.1-41.5% of total knee arthroplasty (TKA) revision procedures [1-4]. Typically, surface cemented and uncemented TKA tibial components are press-fit into the tibial metaphysis to achieve initial primary fixation. The effect of implantation on the surrounding trabecular bone is of great importance in understanding the mechanics of press-fit total arthroplasty components. During press-fit implantation the trabecular bone undergoes multiaxial loading, high confinement pressures and inelastic deformation at large strains. Stresses sufficiently high to cause inelastic deformation of trabecular bone can also be generated during the press-fit implantation of acetabular cups [5] and femoral stems [6] in total hip arthroplasty. Despite this, biomechanical studies of total joint arthroplasty largely focus on the comparison of the pre-implanted intact bone with the post-implanted bone [7-12], neglecting the effect of the inelastic deformation of the trabecular bone during surgical implantation. Previous finite element studies investigating press-fit implantation of total arthroplasty stems have used linear elastic material models [13]. To the authors' knowledge, a plasticity formulation that includes pressure dependent yielding of the trabecular bone or PU, a synthetic trabecular bone analogue material, has not been considered for the implantation of a press-fit stem in total knee or hip arthroplasty.

A recent study by the authors (Kelly and McGarry (2012)) investigated the inelastic deformation of samples of bovine tibial trabecular bone and polyurethane foam (PU) under uniaxial and confined compression loading to large inelastic strains [14]. Pressure dependent yield behaviour of trabecular bone under confined compression was uncovered and it was demonstrated that a crushable foam with isotropic hardening (CFI) plasticity formulation reasonably predicts the inelastic behaviour of both bovine trabecular bone and PU trabecular bone analogue [14]. The objective of the current study is to demonstrate the importance of using a crushable foam plasticity formulation to capture the inelastic pressure dependent behaviour of the trabecular material during the macroscale press-fit implantation of a TKA tibial component. The study also investigates the potential inaccuracies that result from the commonly adopted von Mises (VM) plasticity formulation to model the inelastic behaviour of the trabecular bone. Experimental implantation of a tibial component into a synthetic tibia is also performed and the results are compared to the computed results for both plasticity formulations.

2. Materials and Methods

2.1 Macroscale Press-fit Tibial Component Implantation

Seven synthetic composite tibiae (#3402, 10 pcF PU, $\rho = 0.16 \text{ g/cm}^3$) by Sawbones® (Pacific Research Labs, Malmö, Sweden) were used in the present study (Figure 1). All stages of tibial preparation were performed in accordance with the manufacturer's instructions by an orthopaedic resident (DTC) under the supervision of a consultant orthopaedic surgeon (FJS). Experimentation was performed at room temperature in dry conditions. The distal end of the tibia was potted. A tibial component of the Genesis II Total Knee System (Smith & Nephew, Memphis, TN, USA) was implanted into seven synthetic tibiae using an Instron testing machine (Model 4467, Instron Corp., Canton, MA, USA) under displacement control at a constant rate of 20 mm/min. The proximal ~30 mm of the tibial component had a diameter of 12 mm and the distal ~20 mm had a diameter of 10 mm (Figure 1). The punched hole had a constant diameter of 11 mm. Hence, the proximal ~30 mm of the

tibial component was press-fit to achieve initial implant stability, with an interference of 1 mm. Force-displacement data were recorded for each tibial implantation.

Computed tomography (CT) images of a composite tibial Sawbone® (#3402, 10 pcf PU, $\rho = 0.16 \text{ g/cm}^3$) and replica models of the Genesis II tibial punch and tibial component were utilized to create 3D models of the experimental set-up (Figure 1). The CT images (SOMATOM Sensation Cardiac 64™, Siemens, Forchheim, Germany) had a slice thickness of 1 mm. The cortical bone and trabecular bone were segmented according to the CT data. The proximal tibia was prepared as per the experimental testing using the software Mimics (v14.11, Materialise, Leuven, Belgium). A 3D continuum finite element model of the prepared tibia was created (v6.0 3-matic, Materialise, Leuven, Belgium) consisting of ~445,000 four noded linear tetrahedral elements and exported into the finite element solver Abaqus (v6.11 Abaqus Explicit, Dassault Systèmes, Providence, RI, USA). Four noded linear tetrahedral elements (no mid-side nodes) were chosen over 10 noded quadratic tetrahedral elements (with mid-side nodes) in order to accurately simulate contact between the implant and bone. Loads and boundary conditions were applied to replicate the experimental set-up. The distal end of the tibia was constrained in all directions. To achieve the press-fit implantation a displacement boundary condition was applied to the tibial tray in the axial direction (along the stem axis). The titanium alloy tibial component was treated as a rigid body as it is several orders of magnitude stiffer than the trabecular bone. Hence, interaction between the implant and the bone will result in significant deformation of the bone and negligible deformation of the implant. The cortical bone (short fiber glass epoxy resin) has an apparent Young's modulus (E) of 16.7 GPa and an elastic Poisson's ratio (ν_e) of 0.26 (according to the manufacturer [15]) and was assumed linear elastic, homogeneous and isotropic. The synthetic trabecular material, PU, was assumed homogeneous and isotropic and was modelled in separate analyses using a VM plasticity formulation and a CFI plasticity formulation. The elastic, yield and strain hardening material parameters for the PU were based on the uniaxial and confined compression testing of individual samples of PU as described in section 2.2. A surface-to-surface penalty contact algorithm was implemented between the implant and the bone. The effect of friction at the bone-implant interface was also investigated parametrically, varying the coulomb friction coefficient (μ) between 0 and 0.3 [6, 13]. A mesh sensitivity study was also performed.

2.2 Characterisation of PU Material Behaviour

In order to accurately simulate the implantation process of the tibial component, as described above, it was firstly important to characterise the multiaxial elasto-plastic behaviour of the PU. The method of testing the individual PU samples has been detailed in a previous study by the authors for a different density of PU (20 pcf, $\rho = 0.32 \text{ g/cm}^3$) [14] and is briefly summarised in this section. Cubic samples of 15 mm ($n = 10$) of grade 10 cellular rigid closed cell PU (10 pcf, $\rho = 0.16 \text{ g/cm}^3$) by Sawbones® (Pacific Research Labs, Malmö, Sweden) were cut using a bandsaw (Jubilee VBS 360, Addison Saws Ltd., West Midlands, UK). Uniaxial compression testing ($n = 5$) was performed using the standard platens test. Confined compression testing ($n = 5$) was performed on a custom made confinement rig, preventing lateral displacement of the specimen and hence generating a multiaxial stress state. A mechanical test machine was used to load the specimens under displacement control at a constant rate of 5 mm/min up to 60% nominal strain in the axis of loading (Instron Corp., Canton, MA, USA). The VM and CFI plasticity formulations were implemented in order to investigate

their ability to replicate the inelastic behaviour of the PU under uniaxial and confined compression loading configurations.

3. Results

The results of this study are presented into two sections: In section 3.1, the material properties of the PU are characterised and the ability of the VM and CFI plasticity formulations to replicate the experimental behaviour is investigated. In section 3.2, the experimental and computational macroscale implantation of a TKA tibial component into synthetic tibiae is performed. The trabecular material (PU) is modelled using the VM and CFI plasticity formulations.

3.1 Results of Characterisation of PU Material Behaviour

The results of the characterisation of the individual samples of PU under uniaxial and confined compression loading are shown in Figure 3. A mean E of 39.13 ± 5.90 MPa and a mean yield stress (σ_y) of 1.35 ± 0.16 MPa are measured under uniaxial compression where the PU undergoes distinctive yielding followed by a stress plateau and densification at approximately 50% strain (Figure 3A). Under uniaxial compression loading both the VM and CFI plasticity formulations are calibrated to the experimental results (Figure 3A). However, under confined compression loading the VM plasticity formulation greatly over predicts the stresses post yield, as an increase in pressure does not result in material yielding (Figure 3B). In contrast, the CFI plasticity formulation accurately predicts the PU behaviour under confined compression for a uniaxial compression to hydrostatic compression yield stress ratio of $K = 0.7$ and a plastic Poisson's ratio of $\nu_p = 0.23$ (Figure 3B).

3.2 Macroscale Press-fit Tibial Component Implantation Results

Distributions of equivalent plastic strain, von Mises equivalent stress and pressure stress are shown in Figures 4, 5 and 6 for the three section views illustrated in Figure 2. Simulations are performed implementing a VM plasticity formulation and a CFI plasticity formulation for the trabecular material with a friction coefficient of 0.2 at the bone-implant interface (Figures 4-6). As previously stated, the proximal tibial stem is press-fit into the trabecular bone with an interference of 1 mm (Figure 1). As shown in Figure 4A, at 50% implantation higher localisations of equivalent plastic strain are evident at the bone-implant interface for the CFI plasticity formulation than for the VM plasticity formulation. It should also be noted that the plastic zone extends further into the peri-prosthetic trabecular bone for the VM plasticity formulation than for the CFI plasticity formulation (Figure 4A). Similar plastic strain results are also evident at full implantation for both plasticity formulations, as shown in Figure 4B.

Computational results are shown for a transverse section taken at 5 mm from the tibial cut surface in Figure 5 for both plasticity formulations. Again, as shown in Figure 5A, a larger zone of plastic strain is evident for the VM plasticity formulation (extending ~ 5.7 mm from the bone-implant interface) than for the CFI plasticity formulation (extending ~ 4.1 mm from the bone-implant interface). The distribution of high von Mises equivalent stress is more localised at the bone-implant interface for the CFI plasticity formulation in comparison to the VM plasticity formulation, where regions of high stresses extend further into the peri-prosthetic bone (Figure 5B). In Figure 5C, higher pressures are computed in the peri-prosthetic bone for the VM plasticity formulation than the CFI plasticity formulation, due to pressure dependent yielding in the latter case. As expected, very similar trends of stress and plastic strain are also evident for a transverse section at 15 mm from

the tibial cut surface (Figure 6). Overall, the predicted stress states are quite different at the implant-bone interface between the two plasticity formulations.

The predicted force-displacement results for the macroscale press-fit implantation of the tibial component for the VM and CFI plasticity formulations are shown in Figure 7, again with a friction coefficient of 0.2 at the bone-implant interface. Experimentally measured force-displacement insertion results are also shown in Figure 7. As expected given the differences in the predicted stress states, significant differences in insertion force are predicted for both plasticity formulations. For the CFI plasticity formulation, a friction coefficient of 0.2 provides a good match to the experimental results. However, when a VM plasticity formulation with a friction coefficient of 0.2 is used to model the synthetic trabecular material the computed results generally over predict the experimental implantation results.

A parametric investigation of the effect of friction coefficient on the predicted results for the macroscale tibial component implantation is shown in Figure 8. For higher friction coefficients, bigger differences are noted between the two plasticity formulations, with the VM plasticity formulation becoming more inaccurate. For example, for a friction coefficient of 0.3 at 20 mm insertion, the predicted force is ~31% higher for the VM plasticity formulation than for the CFI plasticity formulation. The present study demonstrates that friction plays an important role during insertion at the bone-implant interface. To further emphasise this, a simulation with frictionless contact ($\mu = 0$) at the bone-implant interface is also shown in Figure 8, providing highly inaccurate results for both the VM and CFI plasticity formulations. The mesh sensitivity study reveals that for the VM plasticity formulation ($\mu = 0$), increasing the number of elements by a factor of 2 results in changes in implantation force of less than 3% at 15 mm implantation and a change in plastic zone size of 0.9 % at 5 mm from the tibial cut surface.

4. Discussion

The present study examines plasticity in the peri-prosthetic trabecular bone during press-fit tibial component implantation in TKA. The study reveals that the computed stress distribution, implantation force and plastic deformation in the trabecular bone is highly dependent on the plasticity formulation implemented. By incorporating pressure dependent yielding using a CFI plasticity formulation to simulate the trabecular bone during implantation, highly localised stress concentrations and plastic deformation are computed at the bone-implant interface. If the pressure dependent yield is neglected using a traditional VM plasticity formulation, a significantly different peri-prosthetic stress distribution and implantation force is computed in the trabecular bone. The results demonstrate the importance of modelling the implantation of press-fit stem prostheses due to the high levels of stress and deformation computed in the peri-prosthetic trabecular bone during this process. Moreover, the study demonstrates the significance of choosing the correct plasticity formulation to model the inelastic deformation of the trabecular bone during implantation. The importance of incorporating friction at the bone-implant interface during simulation of the stem insertion process is also emphasised.

The significance of implementing the correct plasticity formulation when simulating the inelastic deformation of trabecular bone during the implantation process of press-fit stems in total arthroplasty prostheses is demonstrated in the present study. Overall, the predicted stress states in the trabecular material are quite

different at the implant-bone interface for both the VM plasticity formulation and the CFI plasticity formulation. Unlike the VM plasticity formulation, the multiaxial stress state developed in the trabecular material during press-fit stem implantation leads to pressure dependent yielding for the CFI plasticity formulation. Highly localised pressures are computed near the bone-implant interface for the CFI plasticity formulation which leads to high levels of localised plastic yielding directly at the interface; hence, lower stresses and pressures are computed further into the trabecular material. Although the present study focuses on PU, a synthetic trabecular analogue material, the ability of a CFI plasticity formulation to capture the multiaxial pressure dependent plasticity of natural trabecular bone has previously been observed [14, 16]. Highly localised peri-prosthetic plasticity has also been observed experimentally during vertebral fusion device subsidence [16]. Correct prediction of this stress distribution at the bone implant-interface will have implications for device design and in particular for peri-prosthetic bone remodelling and initial implant stability of the tibial component, which have been investigated in previous biomechanical studies [7, 17-21]. To our knowledge, the current study is the first to implement the CFI plasticity formulation for the trabecular material in the investigation of press-fit stem implantation in total arthroplasty.

In the present study, for a friction coefficient of 0.2 at the bone-implant interface during implantation, the CFI plasticity formulation provides a reasonable match to the experimental data, whereas, the VM plasticity formulation over predicts the experimental results. An increase in friction results in an increase in the difference in force predicted for both plasticity formulations. For a frictionless implantation, highly inaccurate insertion forces are computed for both plasticity formulations, indicating the importance of the role of friction at the interface during press-fit implantation. A study by Grant et al. (2007), measured friction coefficients of 0.22-0.71 for dry Sawbone® PU on grit-blasted or porous-beaded titanium [22]. However, it should be noted that the applied contact pressures (≤ 1 MPa) of Grant et al. (2007) are not sufficiently high to result in plastic deformation of the PU [22]. In comparison to an elastic case, localised plasticity at the implant-bone interface may lead to significant changes in the PU surface roughness and hence to the friction coefficient. Indeed, it could be suggested that the experimental-computational approach used in the present study, using an accurate pressure dependent plasticity formulation for the PU, provides a method of calibrating the appropriate friction coefficient during press-fit stem insertion where localised plasticity occurs at the contact surface. In the present study, a friction coefficient of ~ 0.2 provides the closest correlation to the experimental implantation data when a CFI plasticity formulation is implemented. Experimental errors may have occurred due to malalignment or slippage of the tibial stem during implantation, although due care was taken during bone preparation and implantation to account for the 3° posterior slope of the Genesis II tibial component. An increase in experimental standard deviation is evident in the insertion results which may be a result of debris generated due to the large deformation of the PU, a feature that is not incorporated into the computational models. The present study firmly establishes that it is extremely important to consider friction at the bone-implant interface (in addition to using an appropriate plasticity formulation) when modelling press-fit stem implantation. A previous study of Monea et al. (2012) included friction ($\mu = 0-0.2$) at the interface for the insertion of a femoral hip arthroplasty stem with press-fit rings [13]. Earlier studies investigating press-fit stem implantation using interface elements at the bone-implant interface [6, 23, 24], without directly simulating the insertion process, fail to capture the significant effect of friction at the interface.

The importance of implementing a pressure dependent plasticity formulation, such as the CFI plasticity formulation, in the simulation of the press-fit stem implantation process is evident in the present work. Implantation stresses computed in the present study are significant to cause yielding of the trabecular material and are comparable or higher to trabecular stresses that occur during normal physiological loading [6, 7, 13]. Despite this, numerous previous arthroplasty studies that aim to accurately predict stresses in the cortical and trabecular bone, neglect to consider these significant stresses generated during implantation and solely investigated the posted-implanted case [7-12, 25]. Apart from the study of Monea et al. (2012) that employed a linear elastic model during stem insertion modelling [13], other studies neglected to simulate the actual press-fit stem insertion process and employed interface elements to simulate the post-implantation press-fit conditions, assuming linear elasticity [24] or a VM plasticity formulation [6, 23] for the trabecular material.

Linear elasticity has also been employed for the investigation of press-fit acetabular cups in hip arthroplasty, either by simulating the insertion process [26, 27] or by neglecting the insertion process and using a volume increase via thermal expansion of the elements [28, 29] to generate prestresses due to implantation in the trabecular material. A recent study by Souffrant et al. (2012) investigated pressure dependent yielding by implementing a crushable foam with volumetric hardening (CFV) plasticity formulation for the synthetic trabecular material when investigating the stability of press-fit acetabular cups in hip arthroplasty [5]. The implantation of a press-fit acetabular cup is, however, quite different to a press-fit stem implantation as it involves very little sliding during insertion. In a recently published experimental and computational study by the authors, Kelly et al. (2012) investigated the subsidence of a vertebral interbody fusion device into ovine vertebrae and a good correlation to the experimental results was achieved when a CFI and a CFV plasticity formulation were used to represent the trabecular bone [16].

Trabecular bone in the proximal tibia is naturally confined by a stiff cortical shell and undergoes multiaxial compressive loading during normal physiological loading in addition to during the press-fit implantation of a TKA tibial component. Therefore, the material parameters used for the plasticity formulations in macroscale testing should be validated against multiaxial loading conditions such as confined compression testing as demonstrated in the present study. Under confined compression a CFI plasticity formulation captures the inelastic pressure dependent behaviour of the cellular rigid PU ($\rho = 0.16 \text{ g/cm}^3$), with the predicted results being qualitatively similar to the experimental stress-strain curves. The results of the present study are in agreement with the confined compression results of Kelly and McGarry (2012) that demonstrate the ability of a CFI plasticity formulation to predict the inelastic behaviour of bovine tibial trabecular bone and an alternate grade of cellular rigid PU (20 pcf, $\rho = 0.32 \text{ g/cm}^3$) [14]. For a VM plasticity formulation the PU stresses are largely over predicted under confined compression due to the absence of pressure dependent yielding. It should also be noted that for an alternative grade and type of PU, solid rigid PU (50 pcf, $\rho = 0.80 \text{ g/cm}^3$), the study of Kelly and McGarry (2012) demonstrated that no pressure dependent yielding is observed under confined compression [14]. Unlike the cellular PU used in the present study, the near poreless 50 pcf solid rigid PU ($\rho = 0.80 \text{ g/cm}^3$) obeys a VM plasticity formulation, hence it is not representative of natural porous trabecular bone under multiaxial loading, due to the fact that it lacks a cellular structure [14]. Therefore, when choosing a synthetic substitute for trabecular bone for experimentation that involves inelastic multiaxial deformation, it is important to select a type and grade of PU that also undergoes similar pressure dependent yielding to trabecular bone.

A limitation of the current work is the use of PU as a substitute for natural trabecular bone. Synthetic bones were used due to their reproducibility in terms of material properties, consistent specimen size, consistent porosity and availability. The use of synthetic bones, tested in dry conditions at room temperature, may also affect the experimental results. Previous studies have performed mechanical testing of implanted devices in water at body temperature [30]. The measured values of Young's modulus (39.13 ± 5.90 MPa) and yield stress (1.35 ± 0.16 MPa) for the PU under uniaxial compression are in the lower range of reported values for human proximal tibial trabecular bone [31-33] and may be representative of osteoporotic trabecular bone. However, as previously stated, under confined compression similar pressure dependent yield behaviour is observed for the PU to that observed for trabecular bone [14]. The friction coefficient between the implant and PU is directly measured in the present study. However as mentioned previously, the localised crushing and yielding of the PU that occurs during the implantation process, makes standard friction measuring difficult. A relationship between contact stress, pressure dependent plasticity and friction coefficient is beyond the scope of the present study but warrants future investigation. It may be considered a limitation that a single implant geometry was considered in the present work. However, it is likely that similar differences in predicted results for the VM and CFI plasticity formulations would also occur for other implant geometries and press-fit stems in other arthroplasty locations. It is a common assumption in computational studies that trabecular bone is homogeneous and isotropic, although, it is widely accepted that trabecular bone is both anisotropic and heterogeneous. In addition to incorporating a pressure dependent plasticity of the trabecular bone, incorporation of anisotropic and heterogeneous trabecular properties in the computational models may yield improved results. In addition to the simulation of plasticity during implantation, future studies should also incorporate damage during implantation using the element deletion techniques [34] or the extended finite element methods (XFEM) for crack growth prediction, as developed by Feerick et al. (2013) for cortical bone [35].

The current work highlights the importance of implementing a pressure dependent plasticity formulation for the trabecular material when using continuum based finite element models. Microstructural based finite element models of the trabecular bone microarchitecture based on high-resolution CT imaging have also been employed for trabecular bone. However, such microstructural models are generally limited to small specimens with limited material behaviour, contact conditions and deformation due to the high computational expense. Using both microstructural based models and continuum based models, a recent study by the authors demonstrated that the apparent level pressure-dependent yielding of trabecular bone occurs as a result of localised stress concentrations and yielding in the trabeculae at a microstructural level without the requirement that the localised yield is pressure dependent [16]. With ever increasing computational power, microstructural based models of macroscale bone-implant applications that include complex non-linear material behaviour, contact conditions and finite deformation may be feasible in the near future. As demonstrated by Kelly et al. (2012), the explicit modelling of the trabecular bone microstructure will still require plastic yielding or damage at a material level in order to provide realistic macroscopic behaviour [16].

A further implication of the findings of the current study, and of the study of Kelly et al. (2012) [16] is that the macroscale multiaxial yield behaviour of porous orthopaedic implants should also be considered, where pressure dependent yield may also occur at a macroscale even though the constituent scaffold may only exhibit pressure-independent yielding. Simulation of a rapid prototyped porous titanium scaffold under uniaxial compression revealed anisotropic yielding at the macroscale [36]. However, confined and hydrostatic compression should

also be simulated to determine if the scaffold exhibits similar pressure dependent yielding to trabecular bone at the macroscale. Such analyses should be highly important for the design of porous titanium implant devices for applications such as spinal fusion, where device subsidence and fracture occurs due to the complex multiaxial loading environment.

5. Conclusions

The present study demonstrates the importance of implementing the correct plasticity formulation for trabecular bone when simulating the inelastic trabecular deformation during the implantation of a press-fit tibial component in TKA. The computed stress distribution, implantation force and plastic deformation in the trabecular bone is highly dependent on the plasticity formulation used. These differences in stress distributions can be explained by the fact that the multiaxial stress state developed in the trabecular material during press-fit stem implantation leads to pressure dependent yielding for the CFI plasticity formulation; hence, reducing the stresses in the peri-prosthetic bone. For a VM plasticity formulation, the pressure dependent yielding is neglected and a very different stress state, implantation force and plastic deformation are predicted. In order to accurately capture the inelastic deformation of trabecular bone during press-fit stem implantation in total arthroplasty, the results of the present study highlight the importance of: (i) simulating the insertion process of press-fit stem implantation; (ii) implementing a pressure dependent plasticity formulation, such as the CFI plasticity formulation, for the trabecular material; (iii) incorporating friction at the implant-bone interface during stem insertion. The implications of simulating the implantation process and correctly modelling the trabecular bone plasticity should be considered in the design and assessment of press-fit total arthroplasty prostheses.

6. Appendix

Von Mises Constitutive Plasticity Formulation

Yield criterion: $F = q$

where the von Mises equivalent stress is given as $q = \sqrt{\frac{3}{2} S:S}$ and the deviatoric stress tensor S is obtained from the stress tensor σ such that $S = \sigma + pI$ noting that the pressure stress is given as $p = -\frac{1}{3} \text{trace}(\sigma)$.

Crushable Foam with Isotropic Hardening Constitutive Plasticity Formulation

Yield criterion: $F = \sqrt{q^2 + \alpha^2 p^2} - B = 0$

$$B = \alpha p_c \qquad \alpha = \frac{3K}{\sqrt{9-K^2}} \qquad K = \frac{\sigma_c^0}{p_c^0}$$

Flow potential: $G = \sqrt{q^2 + \beta^2 p^2} \qquad \beta = \frac{3}{\sqrt{2}} \sqrt{\frac{1-2\nu_p}{1+\nu_p}}$

Where α is the shape of the yield ellipse in the q - p plane, B is the size of the yield ellipse, p_c is the yield stress in hydrostatic compression, K is the compression yield stress ratio, σ_c^0 is the initial yield stress in uniaxial compression, p_c^0 is the initial yield stress in hydrostatic compression, β is the ellipse for the potential flow and ν_p is the plastic Poisson's ratio [37].

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Figure Captions

Figure 1 Experimental and computational models of the Sawbone® tibia (Pacific Research Labs, Malmö, Sweden) and the tibial component of the Genesis II Total Knee System.

Figure 2 (A) Transverse view of the tibial cut surface detailing the coronal section view. (B) Coronal view of the proximal tibia detailing the transverse section view taken at 5 mm and 15 mm from the tibial cut surface. (C) Transverse section of the proximal tibia taken at 5 mm from the tibial cut surface. (D) Transverse section of the proximal tibia taken at 15 mm from the tibial cut surface.

Figure 3 Experimental results (mean±SD) of the samples of polyurethane foam (PU), a synthetic trabecular bone analogue material, under uniaxial compression and confined compression loading configurations. Predicted behaviour for the von Mises (VM) plasticity formulation and crushable foam with isotropic hardening (CFI) plasticity formulation are also shown.

Figure 4 Coronal section views of equivalent plastic strain computational results of the press-fit implantation of a tibial component using a von Mises (VM) plasticity formulation (*left*) and a crushable foam with isotropic hardening (CFI) plasticity formulation (*right*) for the synthetic trabecular bone ($\mu = 0.2$) at (A) 50% implantation and (B) 100% implantation. For clarity the implant and cortical bone are not shown.

Figure 5 Transverse section views taken at 5 mm from the tibial cut surface of the press-fit implantation of the tibial component following full implantation. Results are presented for a von Mises (VM) plasticity formulation (*left*) and a crushable foam with isotropic hardening (CFI) plasticity formulation (*right*) for the synthetic trabecular bone ($\mu = 0.2$). Contour plots are presented for the computed distribution of: (A) Equivalent plastic strain; (B) Von Mises equivalent stress; (C) Pressure stress. For clarity the implant and cortical bone are not shown.

Figure 6 Transverse section views taken at 15 mm from the tibial cut surface of the press-fit implantation of the tibial component following full implantation. Results are presented for a von Mises (VM) plasticity formulation (*left*) and a crushable foam with isotropic hardening (CFI) plasticity formulation (*right*) for the synthetic trabecular bone ($\mu = 0.2$). Contour plots are presented for the computed distribution of: (A) Equivalent plastic strain; (B) Von Mises equivalent stress; (C) Pressure stress. For clarity the implant and cortical bone are not shown.

Figure 7 Experimentally measured implantation force during displacement of the press-fit tibial component into a synthetic tibia. Predicted results are also presented for a von Mises (VM) plasticity formulation and a crushable foam with isotropic hardening (CFI) plasticity formulation with a friction coefficient (μ) of 0.2 at the implant and bone.

Figure 8 Predicted results of the tibial component implantation into a synthetic tibia for a von Mises (VM) plasticity formulation and a crushable foam with isotropic hardening (CFI) plasticity formulation with friction coefficients (μ) of between 0 and 0.3 at the implant and bone. The experimental force-displacement results are also shown for comparison.

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