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Comparison of long-term numerical and experimental total knee replacement wear during simulated gait loading

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Abstract

Pre-clinical experimental wear testing of total knee replacement (TKR) components is an invaluable tool for evaluating new implant designs and materials. However, wear testing can be a lengthy and expensive process, and hence parametric studies evaluating the effects of geometric, loading, or alignment perturbations may at times be cost-prohibitive. The objectives of this study were to develop an adaptive FE method capable of simulating wear of a polyethylene tibial insert and to compare predicted kinematics, weight loss due to wear, and wear depth contours to results from a force-controlled experimental knee simulator. Finite element-based computational wear predictions were performed to 5 million gait cycles using both force- and displacement-controlled inputs. The displacement-controlled inputs, by accurately matching the experimental tibiofemoral motion, provided an evaluation of the simple wear theory. The force-controlled inputs provided an evaluation of the overall numerical method by simultaneously predicting both kinematics and wear. Analysis of the predicted wear convergence behavior indicated that 10 iterations, each representing 500,000 gait cycles, were required to achieve numerical accuracy. Using a wear factor estimated from the literature, the predicted kinematics, polyethylene wear contours, and weight loss were in reasonable agreement with the experimental data, particularly for the stance phase of gait. Although further development of the simplified wear theory is important, the initial predictions are encouraging for future use in design phase implant evaluation. In contrast to the experimental testing which occurred over approximately 2 months, computational wear predictions required only 2 h.

Keywords: Total knee replacement; Wear simulation; Kinematics; Knee mechanics

1. Introduction

Experimental knee wear simulators have been developed in an effort to provide understanding and insight into polyethylene wear mechanisms and pre-clinically evaluate implant designs and bearing materials (Barnett et al., 2002; Burgess et al., 1997; DesJardins et al., 2000; Walker et al., 1997). Although this in vitro testing is invaluable for evaluating bearing materials and total knee replacement (TKR) geometry, typical wear tests are relatively slow and have substantial associated cost. An experimental simulator operating at 1 Hz will take approximately two months to complete a 5 million cycle test, including delays to conduct periodic weight loss evaluation and to change the lubricant. Comprehensive testing is therefore impractical with current knee simulators, as evaluating the influence of implant alignment or various loading conditions may require several years. Perhaps most importantly, perturbations of implant design and their subsequent effects on wear require component manufacture before experimental testing, holistically a costly and time consuming proposition, and therefore do not typically fit within the design-phase timescale of product development.

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In an effort to provide efficient implant wear evaluation to augment experimental testing procedures, researchers initially developed computational wear models of the hip (Maxian et al., 1996a–c) based on finite element (FE) analysis coupled with simple wear theory (Archard, 1953). After hip simulator wear testing provided experimental determination of a wear factor, the numerical wear simulation developed was verified to be predictive for a change in implant head size under identical tribological conditions (Maxian et al., 1997). Subsequent studies have used the methodology to evaluate potential effects of implant head size, liner thickness, and clearance (Maxian et al., 1996a–c, 1997; Teoh et al., 2002), and also demonstrated that long-term predictions of implant wear require an iterative, adaptive-remeshing approach so that changes in surface conformity that occur due to material removal are included (Maxian et al., 1996a–c).

Fregly et al. (2005) developed a model to predict TKR wear driven by in vivo kinematics. The kinematic data were measured using fluoroscopy, and input profiles for gait and stair ascent were approximated from a combination of fluoroscopy and estimated loading, which was assumed to be split into 85% gait and 15% stair climbing. The wear model calculated total penetration as the sum of the predicted wear depth and the surface deformation due to creep. Wear predictions were extrapolated from a single analysis cycle and did not account for changing surface geometry as wear occurred, but were in reasonable agreement with penetration measured on a retrieved tibial insert. However, this agreement was mainly qualitative as the precise patient loading conditions and kinematic profiles over time were unknown. Hence, there is still a need to develop a validated numerical TKR model, using the closely controlled conditions available with an experimental knee wear simulator.

Since changes in TKR geometry can affect knee kinematics, numerical studies of implant wear during the design phase should simultaneously predict both relative tibiofemoral motion and wear. Simultaneous prediction of implant kinematics and contact mechanics has been recently demonstrated using explicit FE models of the Stanmore knee wear simulator (Godest et al., 2002; Halloran et al., 2005a, b). In these models, kinematic verification was performed by comparing experimental and model-predicted motion for a single implant. Halloran et al. (2005a, b) also compared kinematic results from rigid body models using an estimated contact pressure–surface overclosure relationship to results from fully deformable models. Both models were found to produce similar kinematic results. Estimated contact pressure distributions were also closely correlated, as long as significant edge-loading conditions were not present. The rigid body predictions required only 6 min of computational time, providing potential for efficient numerical wear simulation.

The goal of the present study was to evaluate the feasibility of an adaptive remeshing, rigid body FE approach for efficient prediction of TKR wear. Based on the previous model of the Stanmore knee wear simulator (Halloran et al., 2005a, b), the specific objectives were to implement adaptive wear modeling to predict both wear depth and wear volume during gait loading conditions, and to compare results with experimental weight loss and qualitative surface profilometry measurements. A cycle convergence test was performed to determine whether an iterative approach was required, and, if so, the update interval for adaptive remeshing predictions. Wear predictions using rigid body contact estimates were subsequently made using both displacement and force-controlled boundary conditions. The displacement-control inputs, by accurately matching the experimental tibiofemoral motion, provided an evaluation of the simple wear theory, while the force control inputs provided an evaluation of the overall numerical implant assessment by simultaneously predicting kinematics and wear.

2. Methods

2.1. Experimental wear simulation

Experimental wear testing was performed on a Stanmore knee wear simulator with gait loading conditions at a frequency of 1.1 Hz to 5 million cycles. Two samples of a cruciate-retaining implant were used (*NexGen*® Complete Knee Solution Cruciate Retaining, Zimmer, Inc., Warsaw, IN). The tibial inserts were machined from the same lot of compression molded GUR 1050 ultra-high molecular weight polyethylene (UHMWPE), and were sterilized by gamma radiation in nitrogen, before being artificially aged under pressurized oxygen at 70 °C for 14 days. The tibial component was aligned at 0° tilt and the femoral component was set with a 7° shift in flexion (towards hyper-extension). The neutral position for the experiment (zero anterior–posterior (AP) position) was defined as where a fully extended femoral component would settle statically on a horizontal tibial component under an applied vertical load.

To simulate a retained posterior cruciate ligament (PCL) and a resected anterior cruciate ligament (ACL), stiff springs (33.8 N/mm) were positioned in the AP plane of motion for the posterior restraint of the tibial component, while soft springs (7.24 N/mm) were used in the anterior direction (Haider and Walker, 2002). The natural laxity of the knee in the neutral position was modeled with a 2.5 mm gap on each side, to allow 2.5 mm of free motion in each direction before the springs were engaged.

In the experiment, the force-controlled inputs used were nearly identical to ISO gait loading conditions (ISO Draft Standard 14243-2) with the exception that the axial force input included a higher peak load magnitude (3200 N compared with the standard walking cycle peak of approximately 2500 N) and increasing load after 80% of the cycle (Fig. 1). The tibial component was aligned with a 5 mm medial offset to the axial loading axis to simulate the natural in vivo varus loading in the average TKR (Haider and Walker, 2002).

During the test, the specimens were lubricated with bovine serum at 37 °C, and testing included pre-soaking the components to stabilize the liquid content and temperature. An active soak control was utilized to account for fluid absorption in the test specimens, and was set up in a similar manner to the wear specimens but the loading conditions consisted of purely axial loading. Weight loss due to wear was determined gravimetrically at regular intervals (not exceeding 500,000 cycles) throughout the test. Surface profilometry was performed using a coordinate measuring machine before and after the test to map the net change in surface geometry.

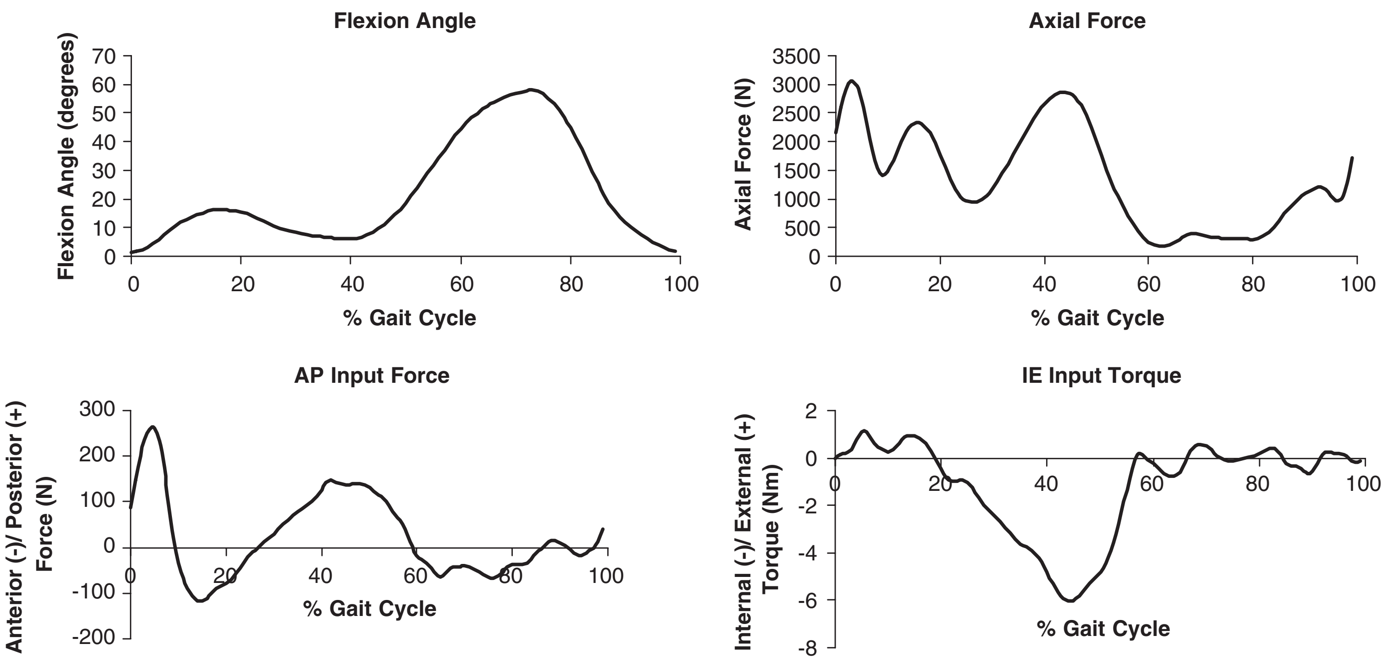

Fig. 1. Knee wear simulator inputs: flexion angle, axial force, AP force and IE torque.

2.2. Explicit FE model

An explicit FE model based on the work of Halloran et al. (2005a, b) was developed in Abaqus/Explicit (Abaqus, Inc., Providence, RI) from CAD models of the TKR. It is a fully dynamic model able to predict the motions and polyethylene stresses when loaded in a gait cycle. The tibial insert was meshed using 8-noded, three-dimensional hexahedral elements (~9000). The femoral component was represented by rigid triangular surface elements (~25,000), which accurately captured the contact area without greatly affecting the solution time (Halloran et al., 2005a, b). Insert element edge lengths were determined in prior convergence studies under similar conditions (Halloran et al., 2005a, b). The coefficient of friction between the articulating surfaces was assumed to be 0.07, in agreement with the range reported in the literature (Estupinan et al., 1998; Sathasivam and Walker, 1997). A penalty-based method was employed to define contact. For computational efficiency, both the femoral and tibial components were represented as rigid bodies. The nonlinear pressure–overclosure relationship was optimized specifically for the mesh and loading conditions utilized so that the kinematics and contact mechanics predicted were comparable with a fully deformable analysis (Halloran et al., 2005a, b).

The FE model sought to reproduce the mechanical environment present in the Stanmore knee simulator. Simulated soft-tissue constraints present in the knee simulator, consisting of a set of four springs that constrain the insert in AP displacement and internal–external (IE) rotation and a spring gap designed to represent anatomical laxity, were reproduced in the model (Fig. 2). The femoral component was constrained in IE, medial–lateral (ML), AP, and varus-valgus (VV) degrees of freedom, while flexion rotation and compressive loading were applied (Fig. 2). The distal surface of the tibial insert was supported in the inferior–superior (IS) direction, insert tilt was constrained, and VV and ML degrees of freedom were free. As in the experiment, the VV rotational axis was offset medially by 5 mm to create a physiological ML load split, i.e. with a greater proportion of the load passing through the medial condyle.

Two FE models were constructed, based upon the nature of the prescribed boundary conditions, either force- or displacement-controlled. The force-controlled model used the feedbacks from the control system of the inputs used by the simulator, i.e. the actual forces seen by the implant rather than the applied inputs, applying a flexion angle and compressive load to the femoral component and an AP force and IE torque to the tibial insert. The displacement-controlled model used the experimentally measured initial kinematics as inputs, substituting the AP force and IE torque with AP translation and IE rotation, respectively (Fig. 2). Driving the FE model with the measured displacements provided an evaluation of the ability to predict wear based on simplified wear theory, while the force-controlled simulations provided an evaluation of the whole numerical experiment, i.e. the ability to simultaneously predict the kinematics and wear.

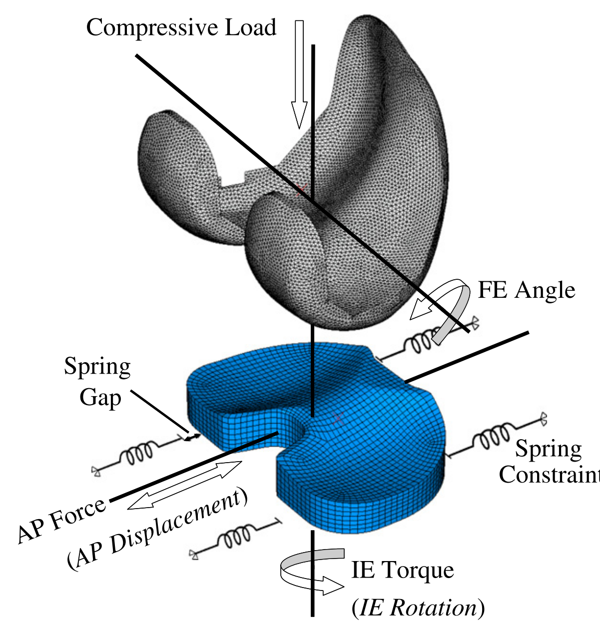

Fig. 2. Finite element model of Stanmore knee simulator illustrating force-control loading conditions (displacement-control conditions shown parenthetically).

2.3. Numerical wear simulation

The numerical wear simulation utilized Archard's Law (Archard, 1953) to calculate surface wear of the UHMWPE insert:

$$H = K_w p S, \tag{1}$$

where H is the wear depth, K_w is an experimentally determined wear factor, p is contact pressure, and S is sliding distance. The wear factor (2.64E−07 mm^3/N m) employed in this study was estimated from an average of wear factors from TKR and ball-on-flat wear tests in the literature (McGloughlin et al., 2004).

An adaptive remeshing procedure was implemented in order to simulate the progression of surface wear (Fig. 3). The adaptive wear simulation was carried out using Python scripts (Stichting Mathematish Centrum, Amsterdam, The Netherlands) to interface with the Abaqus/Explicit output database. For each iteration of the wear simulation, a list of tibial surface nodes was obtained, along with the contact pressure and relative slip (sliding distance) at each of those nodes for one gait cycle, input into Archard's wear law, and extrapolated to a pre-determined number of cycles. Simultaneously, a list of tibial surface elements and their associated nodes were extracted from the model. From this information, simple vector theory is used to calculate the surface normals of each element. The nodal normals are defined as the average of the normals of the surrounding elements. The calculated wear depth at each node was used to update the nodal position along the nodal normal by moving the node the required distance. The total wear volume of the insert was also calculated using a Python script after the wear depth results were determined. In order to compare the FE results to the experimentally determined gravimetric weight loss, the wear volume was converted to a weight using a density of 0.93 g/cm^3.

The displacement-controlled model was used to establish an appropriate update interval, i.e. the number of experimental gait cycles represented by each iteration of the wear simulation. Maximum wear depth and volumetric wear, in addition to contact pressure and area, were evaluated with a variable update interval, starting with 5 million cycles (only one FE analysis), and reducing until the wear predictions converged. Using the converged update interval, insert wear depth and weight loss estimates (calculated from the wear volume loss) were calculated for 5 million cycles using both the force- and displacement-controlled models.

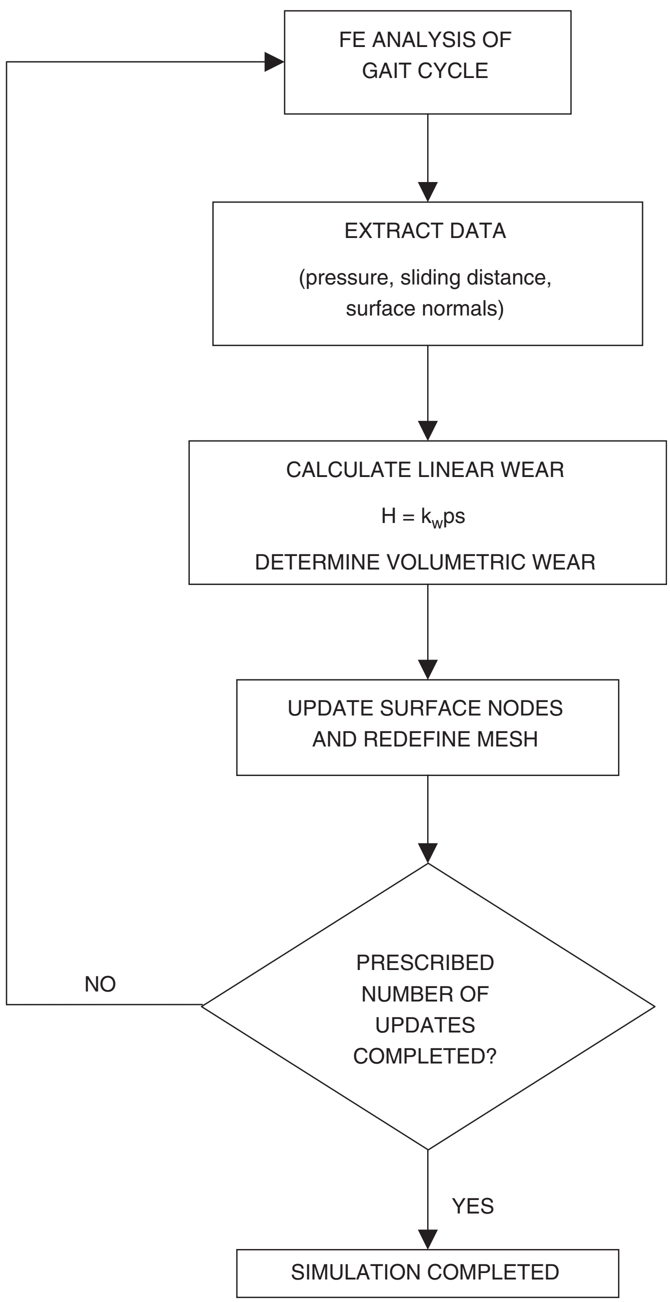

Fig. 3. Flowchart of adaptive remeshing procedure.

3. Results

3.1. Wear convergence study

Convergence analysis of the numerical wear simulation revealed that the predicted wear variables showed different sensitivities to the size of the update interval. Weight loss predictions were not very sensitive to the update interval, differing less than 3% between a single iteration using a 5 million cycle update interval and 10 iterations using a 500,000 cycle update interval. Contact pressure and contact area predictions over the gait cycle were shown to be converged using five iterations of 1 million cycles, since this interval size produced pressure results within 0.5% and area results within 3% of a simulation using 40 iterations of 125,000 cycles. Maximum wear depth was the parameter most sensitive to the update interval, showing initial oscillation and requiring a 500,000-cycle interval to achieve convergence. Reducing the update interval from 500,000 cycles to 125,000 cycles showed less than 2.75% difference in the predicted maximum wear depth (Fig. 4). On the basis

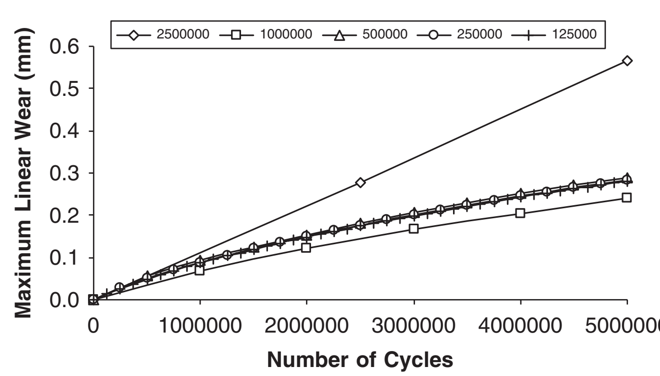

Fig. 4. Temporal convergence of maximum linear wear depth predictions. A 500,000-cycle update interval was chosen for subsequent analyses.

of these results, the update interval for further analyses was chosen to be 500,000 cycles.

3.2. Kinematic prediction

Kinematic predictions from the force-controlled simulation at the beginning of the wear test were in reasonable agreement with the measured experimental data, especially during the stance phase of the gait cycle (Fig. 5). Initial AP translations were in good agreement until peak flexion near 70%, where the model overpredicted the anterior motion by approximately 0.9 mm. During the swing phase, the insert translated posteriorly at approximately the same rate as measured experimentally (Fig. 5). Overall, the mean prediction error (absolute average over the gait cycle) was 0.48 mm. Initial IE rotation predictions also correlated to the experimental results, especially in stance, with a mean prediction error of 0.54°.

Comparison between the experimental and predicted kinematics was also completed following 5 million gait cycles (Fig. 5) to determine how wear would influence the correlation between experiment and simulation. The experimental results showed a small posterior shift (~2.3 mm) in AP translation that was not reflected in the model predictions. During the stance phase, a reduction in IE rotation of 1–2° that was observed in the experiment was accurately predicted in the model (Fig. 5). During the swing phase, however, the agreement was substantially reduced.

3.3. Wear prediction

The predicted weight loss of the displacement-controlled model closely followed the experimental wear results and fell within the bounds of the two implants tested (Fig. 6). The volumetric wear rate of the force-controlled model was within 13% of the displacement-controlled model, although it was approximately 6% less than the lower bound of the experimental result at the end of the test. As measured in the experiment, both force- and displacement-controlled models predicted a volumetric wear which accumulates linearly over time.

The maximum wear depths predicted by the force- and displacement-controlled models were comparable in both trend and final magnitude, although the force-controlled model predicted a somewhat smaller maximum wear depth magnitude at intermediate time points during the simulation (Fig. 7).

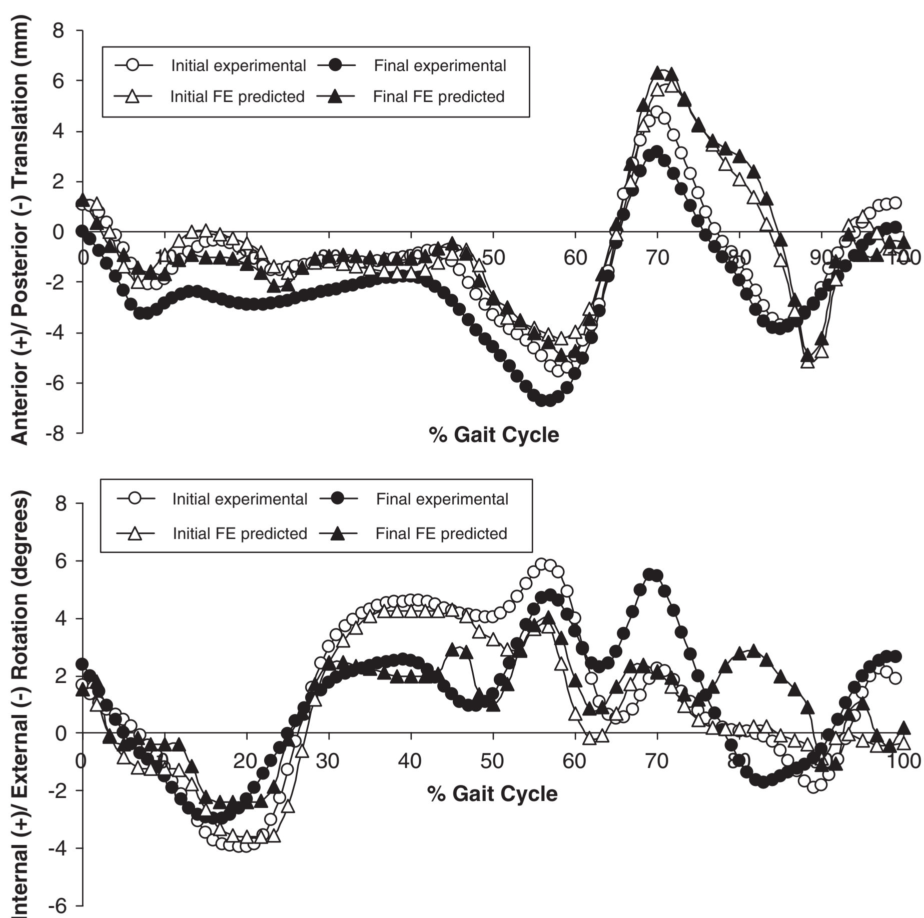

Fig. 5. Comparison between initial and final AP translation and IE rotation of the tibia with respect to the femur for experimental results (displacement control) and predicted kinematics (force control).

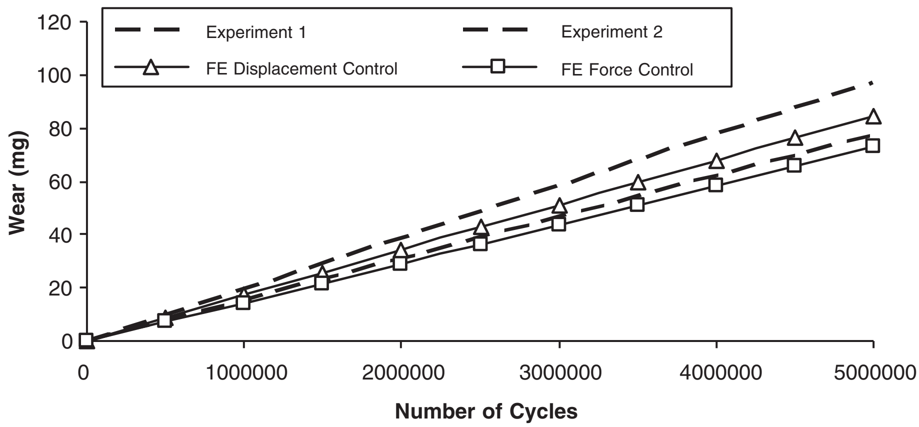

Fig. 6. Weight loss for two stations of the wear simulator and the displacement- and force-controlled FE predictions.

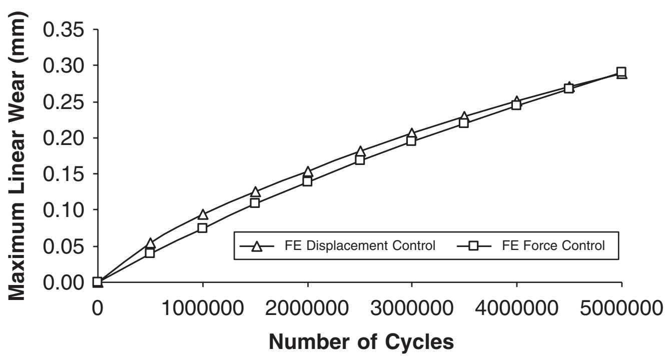

Fig. 7. Maximum linear wear depth for displacement- and force-controlled FE models.

To perform a single wear iteration, a FE simulation was performed, and the single-cycle wear results were calculated from the simulation kinematic and contact pressure results. Each single wear iterations required approximately 10 min of CPU time on a 3.0 GHz Intel PC. The single-cycle wear calculations are then multiplied by 500,000, and the surface geometry updated accordingly. This process is repeated 10 times to predict wear over 5 million cycles, and the entire simulation required approximately 2 h of CPU time.

3.4. Comparison of experimental and predicted wear contours

Fig. 8 shows a comparison between the wear contours predicted by the displacement- and force-controlled models and the surface profilometry plots obtained from the experimental tibial inserts after 5 million gait cycles. Model results demonstrate good qualitative agreement with the profilometry data, showing similar extents, periphery, and peaks. Both the model predictions and the in vitro tibial components show a deep wear scar near the center on the medial side, and a large, more shallow patch on the lateral condyle. Unfortunately, no quantitative data for wear depth was taken at the time of the experiment. There is also good agreement between the force- and displacement-controlled FE models, both exhibiting a deep wear region near the center of the medial condyle, and a shallower wear scar on the lateral condyle. The force-control simulation, however, more accurately predicted that the lateral wear scar would fully extend to the posterior edge, as shown in the experimental contours.

4. Discussion

This is the first study to develop an adaptively remeshing model of TKR wear and compare it with an experimental knee wear simulator. Although there have been several studies that model wear in total hip replacements (Maxian et al., 1996a–c), there has been only one other study to computationally model TKR surface wear (Fregly et al., 2005). This study was not adaptively remeshed and compared the predicted results with retrieved tibial components.

Currently, significant parametric studies on the effects of geometry, alignment, or loading on implant wear are typically not performed during the design phase due to the testing time required. In an effort to develop efficient computational tools to evaluate TKR wear, the aim of this study was to provide a baseline comparison of model-predicted insert wear based on simple wear theory with carefully controlled wear experiments. A displacement-controlled model was utilized to evaluate the predictive power of the simple wear theory, while a force-controlled model provided an evaluation of a numerical experiment such as would be used in design-phase iteration with no a priori knowledge of potential kinematics.

As kinematic prediction is paramount to a successful wear simulation, kinematics at the beginning and end of the wear simulation were compared with the experimentally measured motions. In each case, predicted kinematics were in reasonable agreement with measurements, particularly during the stance phase. Prediction quality diminished during the swing phase, likely to be a result of the lack of constraint provided by the spring gap. While the spring gap represents the toe in region of a typical joint laxity curve, it creates a more challenging situation for kinematic

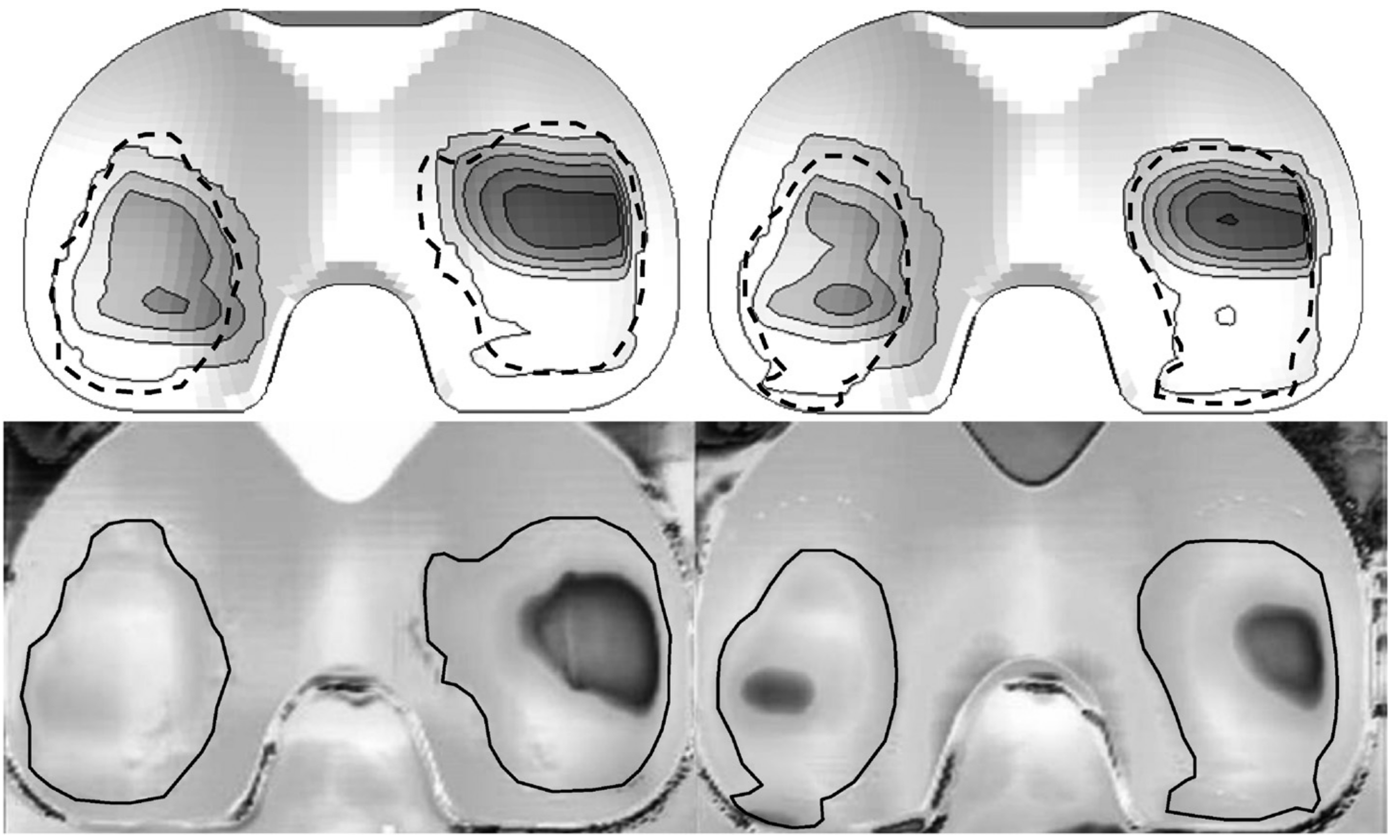

Fig. 8. Predicted wear depth contours from displacement- (top left) and force-controlled (top right) analyses with experimental surface profilometry for both implants (bottom) after 5 million cycles.

prediction due to the potential for the insert to have, at times, very little restraint to motion. In this case, during the 60–80% region of the gait cycle, axial, AP, and IE loading is very small and the change in flexion angle is rapid. For this reason, experimental kinematics are likely quite sensitive to friction or damping present in the simulator that are not included in the simplified FE model. Kinematics at the end of the wear test are likely affected by a change in surface geometry (due not only to wear, but also plasticity and creep) as well as frictional characteristics. The models account only for wear-induced changes in geometry and assume a constant coefficient of friction.

Like prior hip wear simulation studies (Maxian et al., 1996a–c), it was confirmed that the update interval should be evaluated for convergence. Although weight loss was reasonably predicted based on one cycle, converged calculation of wear depth required an iterative approach. The predicted peak linear wear oscillated before convergence at an update interval of 500,000 cycles. The slight oscillation was due to the fact that with the surface geometry updates, both the relative kinematics and contact pressures can change. In this case, when using an update interval of 1 million cycles, the node at which peak penetration occurred experienced slightly less slip (relative sliding distance) and hence smaller linear wear. The results suggest that, under these loading conditions, an update interval of 500,000 cycles provides consistent results for wear volume and maximum wear depth. This interval is consistent with numerical hip wear studies (Maxian et al., 1996a–c).

The predicted weight loss was found to be related by a linear function to the number of loading cycles, and in good agreement in magnitude and trend with gravimetric data. Using the experimentally measured kinematics as inputs to the model led to a weight loss prediction which was bounded by the two experimental specimens, thus indicating that the chosen wear factor was reasonable. The force-controlled model, displaying small differences in kinematic behavior, predicted a similar trend in insert weight loss, but an approximately 13% lower value than the displacement-controlled model. The predicted wear contours were in reasonable agreement with the extents, periphery, and qualitative location of peak penetration shown in the profilometry patterns. However, the experimental profilometry is an indication of total surface deformation, which includes components of plasticity and creep as well as wear.

The initial results of this baseline study are encouraging, but it must be noted that there are several significant assumptions associated with the numerical wear evaluation. The displacement-controlled model used only the initial kinematics as inputs, despite the changing kinematics observed in the experiment. However, only the initial and final kinematic curves were recorded during the experiment, and although there were some small changes, no significant differences were observed, so this was deemed an acceptable assumption. The FE model was based on the CAD implant geometry, and not the as-manufactured part utilized in the experiment. Any differences in geometry due to manufacturing tolerances may contribute to the discrepancy between the simulation and experimental results. In addition, the numerical model does not account for material plasticity or creep (Bevill et al., 2005; Fregly et al., 2005). It is likely that the inclusion of creep would affect the linear wear, and therefore the wear scar, in the initial stages of the wear simulation (Fregly et al., 2005), and its exclusion may contribute to the disparities observed between the measured and predicted wear. However, the extent to which creep will affect the weight loss predictions is currently unclear. Only an abrasive/adhesive wear mode was incorporated and there was no inclusion of fatigue damage and pitting. However, there were no observations of gross delamination, fracture, or failure of the experimental inserts, and these wear phenomena are less likely to occur with current polyethylene processing techniques. More complex wear theories may need to be incorporated in future wear simulations to accurately predict those damage mechanisms. The wear factor was spatially and temporally constant, and taken from the literature rather than from the experimental work. This could account for the differences between the predicted and measured wear volumes. This was a retrospective study and so a wear factor was not determined directly from the experiment, however, every effort was made to use a wear factor that was representative of knee contact conditions. The wear factor did not accommodate for local cross shear effects, hence, variations in relative kinematics are only able to produce variations in predicted wear according to total sliding distance, regardless of sliding direction. Ongoing work aims to include cross shear effects in the wear calculation in order to provide a wear factor that is not constant across the surface and therefore, a more accurate representation of tibial surface wear (Knight et al., 2006).

In its current form, the numerical wear simulation seems most appropriate for evaluation of changes in geometry or loading than material, as more detailed wear data for specific bearing material combinations have yet to be experimentally quantified. However, despite the limitations outlined above, the applicability of the method has been demonstrated and this numerical approach is capable of generating kinematic, weight loss, and wear contour predictions which are for the most part, in reasonable agreement with experimental data. The simulations are also much more time- and cost-efficient than physical wear testing, allowing a wider variety of design parameters to be investigated that was previously feasible. To be used properly in design phase evaluation, further work should evaluate the ability to appropriately differentiate and rank implant designs, loading conditions, and surgical placement for wear performance.

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