The effect of plate design, bridging span, and fracture healing on the performance of high tibial osteotomy plates

AN EXPERIMENTAL AND FINITE ELEMENT STUDY

A. R. MacLeod, G. Serrancoli, B. J. Fregly, A. D. Toms, H. S. Gill

University of Bath, Bath, United Kingdom

Objectives
Opening wedge high tibial osteotomy (HTO) is an established surgical procedure for the treatment of early-stage knee arthritis. Other than infection, the majority of complications are related to mechanical factors — in particular, stimulation of healing at the osteotomy site. This study used finite element (FE) analysis to investigate the effect of plate design and bridging span on interfragmentary movement (IFM) and the influence of fracture healing on plate stress and potential failure.

Materials and Methods
A 10° opening wedge HTO was created in a composite tibia. Imaging and strain gauge data were used to create and validate FE models. Models of an intact tibia and a tibia implanted with a custom HTO plate using two different bridging spans were validated against experimental data. Physiological muscle forces and different stages of osteotomy gap healing simulating up to six weeks postoperatively were then incorporated. Predictions of plate stress and IFM for the custom plate were compared against predictions for an industry standard plate (TomoFix).

Results
For both plate types, long spans increased IFM but did not substantially alter peak plate stress. The custom plate increased axial and shear IFM values by up to 24% and 47%, respectively, compared with the TomoFix. In all cases, a callus stiffness of 528 MPa was required to reduce plate stress below the fatigue strength of titanium alloy.

Conclusion
We demonstrate that larger bridging spans in opening wedge HTO increase IFM without substantially increasing plate stress. The results indicate, however, that callus healing is required to prevent fatigue failure.

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Keywords: Osteotomy, Plate, Stress

Article focus
- The influence of working length when using different opening wedge designs of high tibial osteotomy (HTO) plates.
- The influence of healing on HTO plate stress reduction when using different plate designs.

Key messages
- Larger bridging spans in opening wedge HTO increase interfragmentary movement (IFM) without substantially increasing plate stress.
- Osteotomy gap healing is required to prevent plate fatigue failure.

Strengths and limitations
- This is an experimentally validated computational study.
- Several situations were evaluated involving different screw configurations and stages of healing.

Introduction
The lifetime risk of knee osteoarthritis (OA) is estimated to be as high as 45% and is...
becoming more common. Although the demand for knee arthroplasty is predicted to double by 2030, it is only suitable for end-stage disease. High tibial osteotomy (HTO) is an established treatment for early-stage knee arthritis. The procedure uses an opening or closing wedge osteotomy to change the varus alignment to a more valgus alignment, thereby altering the mechanical axis of the leg and reducing the load in the painful compartment. The osteotomy is then stabilized using an osteosynthesis plate. Plates with angularly stable ‘locking’ screws are generally used to ensure the osteotomy angle is maintained, since conventional screws have been shown to subside after the operation. Opening wedge HTO procedures are more popular due to the simpler surgical approach, lower risk of peroneal nerve damage, and better postoperative flexion scores. Unfortunately, the overall complication rate is still relatively high at 31%. Infection is generally the most common complication (approximately 10%), while reported rates of nonunion and delayed union range from 4.3% to 8.2%, and rates of implant breakage range from 4.4% to 10%. Patients with delayed consolidation of the osteotomy have been reported to have a significantly higher rate of complications. Therefore, due to the importance of preventing nonunions and delayed unions from occurring, promotion of fracture healing should be a priority in HTO surgery. In studies by Spahn and Nelissen et al., more plate breakages occurred in the non- or restricted-weight-bearing groups than in the fully weight-bearing groups. This observation implies that the loads imposed on the plate are not the only factor influencing plate breakage. While weight-bearing loads cause plate deformation, they also induce interfragmentary movement (IFM), stimulating healing at the fracture site, which in turn reduces stress on the plate. It is well known that wider fracture gaps require larger movements to achieve the same strain environment, and, unsurprisingly, the likelihood of HTO complications is related to the size of the osteotomy angle. Therefore, with the exception of infection, the most common complications in opening wedge HTO are related to mechanical factors and, in particular, the stimulation of healing at the osteotomy site.

Previous biomechanical studies have investigated the static strength and fatigue resistance of HTO plates. TomoFix HTO plates (DePuy Synthes, Zuchwil, Switzerland) have been shown to have lower stiffness and migrate more with cyclical loading than iBalance (Arthrex GmbH, Munich, Germany), Contour Lock (Arthrex), and PEEKPower (Arthrex). Despite these findings, TomoFix implants (without wedge blocks) are still the gold standard in HTO treatment and have lower complication rates than shorter spacer plates designed by competitors. In laboratory-based studies, implants that produce higher construct stiffness also generally have greater fatigue resistance. Clinically, however, more flexible implants can perform better in terms of healing, indicating that mechanical properties such as strength and, more importantly, stiffness do not necessarily translate into improved clinical performance.

It is difficult to recreate the in vivo biomechanical environment in vitro. For example, it is not practical to incorporate muscle forces in laboratory experiments. Additionally, fatigue tests in a laboratory setting ignore the fact that the osteotomy will heal over time, making it difficult to predict medium- or long-term in vivo performance. Computer simulation allows the prediction of variables that would be difficult or impossible to measure experimentally. However, the model must be thoroughly validated to ensure it represents reality, and minimum reporting guidelines aid reproduction and improve standards. Some previous computational studies have investigated implant stress, IFM within the osteotomy, and the influence of plate design. A major limitation of previous computational studies using locking plates, however, is the use of bonded plate-screw interfaces. This assumption dramatically increases the construct stiffness while also lowering predictions of plate stress. Additionally, previous HTO computational studies have not considered the influence of healing at the fracture site.

The aim of this study was to create and validate a computational model of a tibia following an HTO procedure in order to compare predictions of IFM, plate stresses, and bone strains for different plate spans and designs. The three main goals of the investigation were: 1) to evaluate the influence of different screw arrangements – in particular, to assess whether a longer bridging span using fewer screws could be used safely; 2) to compare a custom HTO plate against an existing design; and 3) to investigate the influence of callus healing on the predictions of plate stress and bone strain.

Materials and Methods

The study was performed in three stages: 1) experimental testing; 2) creation of a finite element (FE) model validated for the intact condition and for an HTO procedure performed using a custom plate design and two different screw configurations; and 3) investigation of the influence of different plate configurations under physiological loading and including healing fracture callus.

Experimental testing. A composite surrogate tibia was selected for use in the study (third-generation Sawbones model 3302, Biomechanical Test Materials; Pacific Research Laboratories, Malmö, Sweden). The specimen was instrumented with tri-axial strain gauges (SGD-2/350-RY63; Omega Engineering Ltd, Manchester, UK) at five locations of interest (Fig. 1a) based on a preliminary computational study. The composite tibia was potted using a low melting point alloy (Wood’s Metal (70°C);
Lowden Metals Ltd, Halesowen, United Kingdom) and mounted in a materials testing machine (Series 5965; Instron, Norwood, Massachusetts). The distal tibia was restrained using a 20 mm diameter spherical bearing positioned at the centre of the ankle joint. A unicortaneous knee arthroplasty femoral component (Medium Oxford Partial Knee; Zimmer Biomet, Swindon, United Kingdom) with a radius of 23 mm was used to secure the specimen proximally and apply the loading to the medial condyle (Fig. 1a). The load was applied along the axis from the medial condyle to the ankle (Fig. 1a) in increments of 50 N up to 500 N. Strain readings were recorded using data acquisition software (LabVIEW 2010; National Instruments Corporation (UK) Ltd, Newbury, United Kingdom). Maximum and minimum principal strains were calculated from the measured directional surface strains (Matlab R2015b; The MathWorks Inc., Natick, Massachusetts). Each test was repeated ten times and the average was calculated.

After testing in the intact condition, a wedge osteotomy was created (AO classification: 31-A1 / 31-A2). The proximal plane of the resulting osteotomy was inclined 15° to the joint line at the knee, and the gap angle was 10° (Figs 1b and 1c). A custom HTO locking plate was designed to match the surface profile of the composite tibia and manufactured using titanium alloy (Ti-6Al-4V) (EOS M:280 laser sintering; Mario Nava S.p.A., Bosio Parini, Italy). The profile of the plate used locations on the surface of the specimen obtained from a micro-CT scan of the composite tibia (Nikon XTH225 ST CT-scanning unit; Nikon UK Ltd, Kingston upon Thames, United Kingdom). Locking screws were also manufactured; the threads were tapped/threaded as post-processing operations. The plate was located against the bone using a custom positioning jig to ensure it was placed exactly as planned in the virtual environment. Tri-axial strain gauges were also applied to the top and bottom surfaces of the locking plate (Fig. 1b). The plated, osteotomized tibia was then tested in an identical manner to the intact specimen. In total, three scenarios were examined: A) an intact tibia; B) an osteotomized tibia stabilized with the custom plate using a short bridging span with four distal locking screws; and C) an osteotomized tibia stabilized with the custom plate using a long bridging span with three distal locking screws (Fig. 2). The working lengths for the short and long bridging spans considered in our study were 33 mm and 50 mm, respectively, for both plate types. As we were limited to recording 15 strain readings simultaneously (three per strain gauge), during the testing of conditions B and C, two of the bone strain gauges were unused during testing of the implanted tibia. In order to measure IFM in a non-contact manner, the net displacement of two tracking markers was determined using digital image correlation software (Ncorr v1.2; Georgia Institute of Technology, Atlanta, Georgia). The tracking markers were applied to the surface of the bone above and below the osteotomy, and photographs were taken every 100 N during loading (Matlab R2016b; The MathWorks, Inc.) using a digital camera (DFK 23GP031, Resolution: 2592×1944 (5 MP); Imaging Source Europe GmbH, Bremen, Germany).

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![Fig. 1a](image1.png) ![Fig. 1b](image2.png) ![Fig. 1c](image3.png)

a) The location of the five strain gauges used on the intact sawbones specimen in the testing machine. b) The osteotomized and plated sawbones specimen. c) The geometry of the osteotomy.
Finite element modelling: validation. For validation purposes, FE models were created to replicate the three experimental cases A, B, and C (Abaqus 2017, Simulia; Dassault Systèmes, Paris, France). The Sawbones tibia geometry was created from the previously acquired CT scan using image processing software (ScanIp version M-2017.06; Synopsys, Mountain View, California). Material properties of the surrogate cortical and cancellous bone were assigned based on the manufacturer’s information (Pacific Research Laboratories). Linear elastic isotropic material properties were used with Young’s moduli (Poisson’s ratio) equal to 7.6 GPa (0.24) and 157 MPa (0.3) for cortical and cancellous bone, respectively.

Screws were modelled as cylinders with an outer diameter of 5.0 mm. As the interest of the study was primarily the IFM and plate stress, the screw-bone interaction was modelled as tied for simplification. The screw-plate interface was modelled as a penalty-based interaction with a tabulated contact stiffness (Table I) and a standard Coulomb friction model with coefficient of friction equal to 0.8 to represent the macro-interlock of the plate and screw threads mating. Similar values of friction have been used previously to represent rough screw-bone interfaces.

Reference points representing the experimental proximal and distal centres of rotation (knee and ankle) were defined. These reference points were rigidly connected to the bone surface using a multi-point constraint. The proximal reference point was restrained against transverse movement but allowed to move along the mechanical axis. The distal reference point was restrained against all translations. The load was applied proximally in the same manner as the experimental tests.

The FE models were solved using geometric non-linearity with an implicit solver and automatic incrementalization (Abaqus 2017). The models were meshed using fully integrated quadratic tetrahedral elements (C3D10), with an average element edge length of 0.8 mm for the plate, 0.8 mm for the screws, 1.4 mm for the cancellous foam, 1.2 mm for the cortical shell, and 2.0 mm for the callus. In contacting regions, the average element edge length was refined to 0.5 mm. The total number of nodes and elements was 1,050,282 and 769,227, respectively. A mesh convergence study was conducted (convergence criteria, 5%). For the selected mesh resolution, doubling the mesh size altered the prediction of IFM by 3.20%, plate stress by 0.54%, and principal strain in the bone around screw holes by 3.89%.

The predicted principal strains and IFM from the FE models were compared against the experimentally measured values using linear regression.

Finite element modelling: incorporating physiological loading. After validation, a physiological loading case was incorporated. We chose to use the instance of the gait cycle where the medial knee joint contact forces are highest, corresponding to 19% of the gait cycle. Muscle and joint contact forces for a 70 kg patient from a previous study were registered to our geometry (Matlab R2016b). All forces were applied at reference points and distributed over the surface of the bone within a 10 mm to 12 mm radius (Fig. 3). In these models, the most distal part of the tibial shaft was constrained in all directions and no other restraint was applied (Fig. 3). The muscle force and joint reaction vectors are given in Table II.

Finite element modelling: structural optimization of plate. The shape of the custom plate was structurally optimized for the short and long configurations under physiological loading (Tosca Structure 2017, Simulia; Dassault Systèmes). The optimization process iteratively modified the envelope of the geometry in order to satisfy prescribed optimization functions. Our intention was to reduce the maximum plate stress while increasing the average plate strain to generate IFM. Two equally weighted design optimization functions were used: 1) the maximum strain energy density (SED) anywhere in the plate; and 2) the negative of the minimum SED in the bridging span of the plate. The maximum values of both

| Pressure (MPa) | Overclosure (mm) |
|---------------|-----------------|
| 0             | 0.01            |
| 10            | 0.05            |
| 100           | 0.1             |
| 1000          | 1               |

The different scenarios examined experimentally, showing: a) the intact tibia; b) the first screw configuration with a short bridging span annotated (plate superimposed for clarity); and c) the second screw configuration with a longer bridging span annotated.

![Fig. 2a](image1.png) ![Fig. 2b](image2.png) ![Fig. 2c](image3.png)
functions were minimized until the combined SED function was deemed to have converged; at 20 design cycles there was 0.12% change from the previous iteration. Geometric restrictions were placed on the screw holes, and the volume of the plate was constrained to a fraction of 0.9 of the original plate to help reduce underused material. The resulting shape of the optimized plate mesh was smoothed and re-exported.

**Finite element modelling: comparison of different plate types with callus healing.** The FE model was modified to replace the custom plate with a widely used commercial HTO device (TomoFix; Depuy Synthes). The geometry of the TomoFix plate was obtained by reconstructing micro-CT scan data in a similar manner to the bone geometry. Other than the plate geometry, all aspects of the TomoFix model were identical to the custom plate model. For both plate types, predictions were made for short and long bridging spans. The fracture callus was represented by filling the osteotomy gap with a linearly elastic isotropic homogeneous material. Three stages of early callus healing were incorporated into the models: connective tissue formation; fibrous tissue formation; and immature woven bone formation. Young’s modulus and Poisson’s ratios for each stage are given in Table III.34,35

**Results**

**Evaluation of finite element model.** The principal strain predictions of the FE models closely correlated with the experimentally measured data (Fig. 4). The linear regression slope (and $R^2$) values for cases A, B, and C were 0.77 ($R^2 = 0.96$), 1.13 ($R^2 = 0.98$), and 1.06 ($R^2 = 0.98$), respectively. The values of measured and predicted axial interfragmentary displacement (IFM) were within 5% at 500 N with a root mean square error (RMSE) of 0.03 mm and 0.02 mm for the short and long bridging spans, respectively (Fig. 5a). Linear regression of the IFM predictions against measured values gave slope (and $R^2$) values of 0.99 (0.996) and 0.94 (0.995) for cases B and C, respectively (Fig. 5b).

**Comparison of TomoFix and optimized custom plate.** The shape optimization process resulted in the custom plate becoming thicker and wider over the bridging span and around the two screw holes on either side of the osteotomy. Compared with the pre-optimized design, the...
average plate stress increased while the maximum stress decreased. The underside of the plate also became concave.

Comparing the optimized custom plate against the TomoFix plate, the pattern of IFM produced by the two devices was similar (Fig. 6). The custom plate produced a 4.8% smaller axial IFM than did the TomoFix device in the short bridging span but a 24.1% larger axial IFM in the long bridging span. In the custom plate, shear IFM was increased by 52.1% and 54.8% for short and long bridging spans, respectively (Fig. 6a). The values of IFM underneath the plate followed the same trends as those at the IFM measurement locations (Fig. 6b).

In both configurations, the TomoFix plate experienced peak plate stresses at the screw hole proximal to the osteotomy and the plate T-junction, whereas the custom plate distributed the stresses more evenly throughout the span bridging the osteotomy (Fig. 7). For the TomoFix plate, the peak von Mises stress was 656 MPa and 654 MPa for short and long bridging configurations, respectively. For the custom plate, the peak stress was 693 MPa for the short span and 715 MPa for the long span. The peak von Mises stress within the custom plate was 5% larger than with the TomoFix plate for the short span and 10% larger for the long span.

**Strains in the bone around screw holes.** The distribution of strain in the bone around screw locations was different for the two plate types (Fig. 8). For both bridging spans, the TomoFix plates produced larger εVOL around the metaphyseal screw holes compared with the custom plate (Fig. 9), particularly screw holes 1 and 3. The TomoFix plate also produced larger εVOL at the most distal screw (number 8). Compared with the TomoFix, the custom plate produced slightly larger εVOL around the screws adjacent to the osteotomy (numbers 4 and 5) compared with the TomoFix plate. Compared with the short span, the longer span produced 44.1% and
61.1% larger εVOL for the custom plate and the TomoFix, respectively. The εVOL for the custom plate was 32.5% smaller and 39.6% smaller than that of the TomoFix for the short and long spans, respectively. Considering only the screws in metaphyseal bone, the custom plate reduced εVOL by 61.2% and by 51.9% for the short and long spans, respectively.

**Influence of fracture callus.** As the stiffness of the callus increased, the stress in both plate types decreased (Fig. 10). Peak von Mises stress in the TomoFix plate was very similar for both span lengths. The custom short span plate produced lower peak stress compared with the custom long span plate. The peak stress was higher than the fatigue strength of titanium for all cases, except when a callus stiffness of 528 MPa was used (corresponding to immature woven bone).

**Discussion**

This study investigated the biomechanical environment produced by two HTO plate types under physiological loading. Different bridging spans were considered by changing the number of screws used. The study was performed using a validated FE model and demonstrated that longer bridging spans did not necessarily increase plate stress. When early callus formation was included in the models, the plate stress was similar or lower for the longer bridging spans compared with the shorter bridging spans.

By using an optimization algorithm to increase the average plate strain while minimizing the maximum plate strain, we designed a custom plate that produced higher IFM and smaller regions of high strain in the bone compared with the TomoFix design with similar levels of peak strain.
stress. It should be noted that if plate stress was the only specified optimization goal, the resulting plate stress could be substantially lower; however, the stiffness would be increased. To ensure the final plate design was flexible, we also included plate deformation as an equally weighted optimization goal.

Our study confirmed the findings of previous studies that increasing the bridging span increased the axial and transverse IFM.38,39 Augat et al40 demonstrated that transverse motion in the absence of axial motion can delay or prevent fracture healing. Park et al41 showed that oblique motion (a combination of axial and transverse motion) produced the best healing in terms of callus formation. It is also known that intramedullary nails perform very well despite producing larger transverse movements than axial movements at the fracture site.42 Therefore, provided that the device adequately stabilizes the fracture, the combination of transverse and axial IFM may be superior to predominantly or exclusively axial motion.

In previous studies that looked at locking plates, bridging span had a more pronounced influence on IFM than found in our study,38,39 and we attributed this difference to two factors: 1) the orientation of the HTO plate with respect to the loading axis; and 2) the presence of the intact cortical hinge laterally. From the deformation observed in our models, it was clear that the plate...
The study predicted peak plate stresses in both plate designs to be lower than the yield strength of Ti-6Al-4V (approximately 789 MPa to 1013 MPa) in our gap defect models, the plate stresses were much higher than the fatigue strength of titanium (approximately 150 MPa for fatigue fretting in phosphate-buffered saline), indicating that callus healing is required to prevent fatigue failure.

Strain levels around screws can be indicative of loosening risk. When using the TomoFix plate, our study predicted the strains within the bone to be largest in the proximal tibial fragment. This is the location where screw loosening and breakage have been reported to occur clinically, giving us confidence in the clinical relevance of the modelling predictions. The optimized custom plate reduced strain levels in the proximal tibia where the bone stock is weaker. We attributed this finding to the increased amount of material surrounding the screws compared with the TomoFix plate — higher rigidity screws are known to reduce strain at the screw-bone interface. One previous study reported a 6.7% rate of tibial plateau fracture; we did not investigate this aspect in our models, however, as the authors attributed it to a technical error of leaving too much lateral cortex intact.

Martinez de Albornoz et al conducted biomechanical experiments to evaluate IFM using different plate types and found larger values than the present study. Their study used low density synthetic bone and measured IFM at the loading actuator, making their results difficult to interpret. Other authors have measured much smaller bending was mainly around its major axis (the stiffer direction). Additionally, the intact lateral hinge resisted plate bending in the weak axis direction. It is likely that a complete osteotomy or a broken lateral hinge would substantially increase IFM.

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This study has several important limitations. The operative technique for TomoFix includes the use of a compression screw to pull the plate closer to the bone, thus generating tensile preload in the plate and compressive preload in the lateral cortical hinge. We did not incorporate this phenomenon as it would be highly variable depending on the size of the hinge and the position of the plate. It would, however, alter predictions of plate stress, IFM, and VOL. Validation was only conducted using the custom plate, and we assumed that the screw-plate contact parameters would be the same for the TomoFix plate. The experimental tests were primarily for validation purposes, with Sawbones selected for their consistent material properties. Nevertheless, we only used a single specimen. It is difficult to compare our results with those of previous studies due to differences in loading regimes and geometrical variation such as the size of the lateral hinge, specimen dimensions, methods of restraining the specimens, and material properties. The material properties of third-generation Sawbones are different in compression and tension. We chose to use the compressive Young’s moduli as our test was a compression test, and implementing differing compressive and tensile moduli in FE codes is not straightforward. Moreover, the material properties of surrogate cortical bone may not be representative of human bone, particularly in osteoporotic cases. We considered the joint contact and muscle forces at 19% of the gait cycle. We also examined the influence of using forces corresponding to 44% of the gait cycle (second contact force peak) but found that it was not as critical for plate stress predictions. In our models, we evaluated the different plate types and configurations using predefined callus properties to represent different healing stages. It is known, however, that plates with larger IFM promote faster callus formation and thus also reduce plate stress more quickly.

In conclusion, we found that larger bridging spans, particularly transverse movements, have the potential to increase substantially the IFM produced at the fracture site. Our optimized plate design produced similar plate stress to the TomoFix plate for both short and long bridging spans while substantially reducing high strain regions within the bone. This study has shown that both plate types and spans produced very large plate stresses in the
gap defect model. Substantial callus formation was required in all cases to reduce stress below the fatigue strength of titanium.

References

1. Murphy L, Schwartz TA, Helmick CG, et al. Lifetime risk of symptomatic knee osteoarthritis. Arthritis Rheum 2008;59:1207–1213.

2. Nguyen USDT, Zhang Y, Zhu Y, et al. Increasing prevalence of knee pain and symptomatic knee osteoarthritis: survey and cohort data. Ann Intern Med 2011;155:725–732.

3. Patel A, Pavlov G, Mújica-Mota RE, Toms AD. The epidemiology of revision total knee and hip arthroplasty in England and Wales: a comparative analysis with projections for the United States. a study using the National Joint Registry dataset. Bone Joint J 2015;97-B:1079–1081.

4. Ruiz D Jr, Koenig L, Dall TM, et al. The direct and indirect costs to society of treatment for end-stage knee osteoarthritis. J Bone Joint Surg [Am] 2013;95-A:1475–1480.

5. Akizuki S, Shikabawa A, Takizawa T, Yamazaki I, Horuchi H. The long-term outcome of high tibial osteotomy: a ten- to 20-year follow-up. J Bone Joint Surg [Br] 2008;90-B:592–596.

6. Kim MK, Ha JK, Lee DW, et al. No correlation angle with stable plates in open-wedge high tibial osteotomy. Knee Surg Sports Traumatol Arthrosc 2015;23:1999–2006.

7. Poronnattamaneewong C, Harmroongj T, Charanancholvanich K. Loss of correction after medial opening wedge high tibial osteotomy: a comparison of locking plates without bone grafts and non-locking compression plates with bone grafts. J Med Assoc Thai 2012;95(Suppl 9):S21–S28.

8. Bonasia DE, Dettoni F, Sito G, et al. Medial opening wedge high tibial osteotomy for medial compartment overload/arthritis in the varus knee: prognostic factors. Am J Sports Med 2014;42:690–698.

9. Hankemeier S, Mommens P, Krettek C, et al. Accuracy of high tibial osteotomy: comparison between open- and closed-wedge technique. Knee Surg Sports Traumatol Arthrosc 2010;18:1328–1333.

10. Wu L, Lin J, Jin Z, Cai X, Gao W. Comparison of clinical and radiological outcomes between opening-wedge and closing-wedge high tibial osteotomy: A comprehensive meta-analysis. PLoS One 2017;12:e0171700.

11. Woodacre T, Ricketts M, Evans JT, et al. Complications associated with opening wedge high tibial osteotomy — A review of the literature and of 15 years of experience. Knee 2016;23:278–282.

12. Warden SJ, Morris HG, Crossley KM, Brukner PD, Bennell KL. Pre-operative planning for fracture fixation using locking plate systems: device configuration and other considerations. Injury 2018;49(Suppl 1):S12-18.

13. Diffo Kaze A, Maas S, Waldmann D, et al. Stability of medial opening wedge high tibial osteotomy using 2 different implants. A biomechanical study. J Orthop Trauma Relat Res 2013;4:2798–2814.

14. Hoffmeier KL, Hofmann GO, Mückley T. Choosing a proper working length of screws placement effect on locking plate fracture fixation devices. Bone Joint J 2017;9:111–120.

15. MacLeod AR, Simpson AHRW, Pankaj P. Experimental and numerical investigation into the influence of loading conditions in biomechanical testing of locking plate fracture fixation devices. Bone Joint Res 2016;5:143–159.

16. MacLeod AR, Simpson AHRW, Pankaj P. Screw-bone interface modelling matter in finite element analysis? J Biomech 2012;45:1712–1716.

17. Damm NB, Morfcock MM, Bishop NE. Friction coefficient and effective interface at the implant-bone interface. J Biomech 2015;48:3517–3521.

18. Scottell Gi, Kinney AL, Frey BJ, Font-Llugués JM. Neumusculoskeletal model calibration significantly affects predicted knee contact forces for walking. J Biomech Eng 2016;138:81001.

19. Isaksson H, Wilson W, van Dokaal Sch, Huiskes R, It K. Comparison of biophysical stimuli for mechano-regulation of tissue differentiation during fracture healing. J Biomech 2006;39:1507–1516.

20. Steiner M, Claes L, Ignatius A, et al. Prediction of fracture healing under axial loading, shear loading and bending is possible using distorsional and dilatational strains as determining mechanical stimuli. J R Soc Interface 2013;10:20130399.

21. Ebacher T, Tang C, McKay H, et al. Strain redistribution and cracking behavior of human bone during bending. Bone 2007;40:1265–1275.

22. Stöffel K, Dieter U, Schachowiak G, Gächter A, Kuster MS. Biomechanical testing of the LCP — how can stability in locked internal fixators be controlled? Injury 2009;34(Suppl 2):B11–B19.

23. MacLeod AR, Simpson AHRW, Pankaj P. Age-related optimization of screw placement for reduced loosening risk in locked plating. J Orthop Res 2016;34:1856–1864.

24. Augat P, Burger J, Schorlemmer S, et al. Shear movement at the fracture site delays healing in a diaphyseal fracture model. J Orthop Res 2002;20:1011–1017.

25. Park SH, O’Connor K, McKellop H, Sarmiento A. The influence of active shear or compressive motion on fracture-healing. J Bone Joint Surg [Am] 1998;80-A:868–878.

26. Duda GN, Mandruzzato F, Heller M, et al. Mechanical boundary conditions of fracture healing: borderline conditions in the treatment of unreamed tibiaal nailing. J Biomech 2001;34:639–650.

27. Hosseini S. Fatigue of Ti-6Al-4V. In: Hudak R, Penhaker M, Majernik J, eds. Biomedical Engineering - Technical Applications in Medicine. Rijeka, Croatia: InTech, 2012:75–92.

28. Giannoudis PV, Schneider E. Principles of fixation of osteoporotic fractures. J Bone Joint Surg [Br] 2006;88-B:1272–1278.

29. Turner CH, Anne V, Pidaparti RM. A uniform strain criterion for trabecular bone adaptation: do continuum-level strain gradients drive adaptation? J Bone Joint Surg [Am] 1997;79:555–563.

30. Ming TS, Koon WM. Autologous bone grafting and revision plating in a case of persistent high tibial osteotomy non-union. J Orthop Case Rep 2016;6:91–93.

31. Donaldson FE, Pankaj P, Simpson AHRW. Bone properties affect loosening of half-pin external fixators at the pin-bone interface. Injury 2012;43:1764–1770.

32. Song EK, Seen JK, Park SJ, Jeong MS. The complications of high tibial osteotomy: closing- versus opening-wedge methods. J Bone Joint Surg [Br] 2010;92-B:1245–1252.

33. Martinez de Albornoz P, Leynes M, Forriol F, Del Buono A, Maffulli N. Opening wedge high tibial osteotomy: plate position and biomechanics of the medial tibial plateau. Knee Surg Sports Traumatol Arthrosc 2014;22:2641–2647.

34. Han SB, Bae JH, Lee SJ, et al. Biomechanical properties of a new anatomical locking metal block plate for opening wedge high tibial osteotomy: unilateral osteotomy. Knee Surg Relat Res 2014;26:155–161.
51. Röderer G, Gebhard F, Duerselein L, Ignatius A, Claes L. Delayed bone healing following high tibial osteotomy related to increased implant stiffness in locked plating. Injury 2014;45:1648-1652.

52. MacLeod AR, Rose H, Gill HS. A validated open-source multi-solver fourth-generation composite femur model. J Biomech Eng 2016;138:124501.

53. Comiskey DP, MacDonald BJ, McCartney WT, Synott K, O’Byrne J. The role of interfragmentary strain on the rate of bone healing—a new interpretation and mathematical model. J Biomech 2010;43:2830–2834.

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Author Contributions

A. R. MacLeod: Designed the study, Collected and analyzed the data, Wrote the manuscript.

G. Serrancoli: Analyzed previous data (used by the study), Wrote the manuscript.

B. J. Fregly: Analyzed previous data (used by the study), Critically evaluated the data, Wrote the manuscript.

A. D. Toms: Designed the study, Critically evaluated the data, Wrote the manuscript.

H. S. Gill: Designed the study, Analyzed data, Wrote the manuscript.

Conflict of Interest Statement

None declared

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