Finite element assessment of block augmented total knee arthroplasty

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Abstract:

Loosening and migration of tibial prostheses have been identified as causes of early total knee replacement (TKR) failure. The problem is made more complex when defects occur in the proximal tibia compromising fixation and alignment. Clinical studies using metal augments have shown these to be an alternative to other means of defect treatment. Finite element (FE) analysis can be used to identify regions that may be prone to loosening and migration. In the current work 3D FE models of TKR uncontained type-2 defects treated with block augments have been constructed and analysed. It has been shown that a metal augment is the most suitable. The use of bone cement (PMMA) to fill proximal defects is not considered suitable as stresses carried by the cement block exceed those of the fatigue limit of bone cement. It has been shown that the stresses in the proximal cancellous bone of block augmented models are significantly below levels likely to cause damage due to overloading. Further, the use of stem extensions have been shown to reduce the cancellous bone stresses in the proximal region thus increasing the likelihood of bone resorption. Given this it is recommended that stem extensions are not require unless necessary to mitigate some other problem.

Keywords: Metal augment, stem extension, finite element analysis, total knee replacement, cement augment

1. Introduction

With the number of total knee replacements (TKRs) increasing every year (over 55,000 in the UK in 2005), the number of revision replacements has also increased and a prosthesis success rate of approximately 90% at 15 years has been reported [1]. Aseptic loosening and migration of prostheses have been identified as primary factors for early failure of TKR and thus a study of their initial stability is necessary to address this problem. Initial stability analysis is based on the hypothesis that migration occurs soon after implantation and is a result of mechanical failure of the cancellous bone surrounding the implant due to high stresses [2].

Obtaining stability of the prosthesis and bone is often made extremely difficult due to defected bone in the proximal regions of the tibia. Stability can be compromised by failure of the surgical team in obtaining correct implant alignment and sufficient bone-implant interface. Augmented features (i.e. wedge, block and conical augments) are commonly used in conjunction with TKR to treat defected bone with the choice of augment being determined by the type of defect presented at the time of surgery. Conditions for the use of prosthetic augments in primary TKR include when $\geq 40\%$ of the interface between the bone and implant is not supported sufficiently by the host bone or when there are indications that the trial implant will displace [3].

Augments are attached to the tibial or femoral components through the use of cement or bolts. Some bone resection of the defect area is often necessary to conform the region to the size of the augment (particularly in the case of block augments) and to decrease the load at

the bone-implant interface [3]. Metal augments have been shown to be a viable alternative to other defect treatments available (i.e. use of bone autograft/allograft or resection of the tibia to the defect level). The use of bone cement to fill proximal defects may be sufficient when defects are small. However their use in larger defects (above 5 mm) may increase the risk of failure when compared with metal augments [4] due to fracture or loosening of the cement mantle leading to instability of the tibial tray. Previous 2D finite element (FE) research has shown that the use of cement blocks to fill proximal moderate uncontained Anderson Orthopedic Research Institute (AORI) type-2 defects results in increased equivalent stresses in the cancellous bone when compared with the use of metal blocks [5] while previous in-vitro studies showed no significant difference in stiffness values between cement block and metal block models [6]. However these results have yet to be verified using a 3D FE analysis.

While the usual approach in the treatment of defects is to combine augmented features with stem extensions of increased lengths, this is often considered over-conservative, with added removal difficulties should future revision be required. The stem extension can reduce or offload stresses from the defected proximal regions to stronger distal cortical regions (i.e. transfer body loads along the stem) particularly in patients with weak bone stock, thus contributing to overall stability of the TKR [7]. However these reduced proximal bone stresses may lead to increased bone resorption due to stress shielding thus compromising stability. Previous 2D FE analysis has shown that the use of stem extensions in conjunction with cement and metal block augments resulted in reduced equivalent stresses particularly in TKR models with large defects (i.e. 15 mm depth) in the proximal tibia [5]. Although previous research in TKR initial stability using FE has been undertaken, little consideration has been given to examining the stability and potential of augmented features using 3D FE

models with (i.e. a tibial tray with a press-fit stem extension) and without (i.e. a tibial tray with a standard stem) the use of stem extensions.

The objective of this study is to consider the effect of augment material and the use of stem extensions on the stability of TKA that treat moderate uncontained type-2 defects of 15 mm. This is achieved using 3D FE models of the implanted tibia. High and low threshold stresses in the cancellous bone have been used to assess potential loosening through cancellous bone failure or resorption respectively. The maximum stresses in the augment have been used to assess potential instability due to augment component damage.

2. Materials and methods

Three dimensional, static, FE models of the proximal tibia implanted with a mobile bearing revision tray were created using ABAQUS 6.7 (Simulia, Warrington, UK). This configuration was selected as it is an implant that is frequently used by the senior clinical author. Conclusions drawn using this configuration may well be applicable to other implant configurations.

The first set of models is representative of a non-defect TKR (see Figure 1) where the implant has a stem extension. The stem extension component was modelled with either tied or press-fit contact conditions (i.e. two models in this first set). These represent clinically cemented and press-fit stem extension constructions respectively. The purpose of this first set of models was to establish the effect of stem extension fixation on the resulting bone stresses.

The second set of models represents a TKR with moderate uncontained type-2 defects on the medial side. The defect was 15 mm deep and was treated using either cement or metal blocks

(see Figure 2). Both augment types were analysed with and without a press fit stem extension (see Fig 3). Thus there were 4 models in this second set. A press fit fixation was used for the stem extension in this second set of models as this reflected the actual technique used in implantation by the senior clinical author. The purpose of this second set of models was to examine the role of a) augment type and b) stem extension on the bone and augment stresses and hence the implant stability.

A 1.5 mm thick cement layer was used to affix the tibial tray and stem to the cancellous bone. Additionally in the cases of the metal block models a 1.5 mm thick cement layer was also placed under the block component. This reflects the practice of the senior clinical author. The geometry of the tibial bone was acquired from the International Society of Biomechanics Finite Element Repository managed by the Istituti Ortopedici Rizzoli, Bologna, Italy [8]. It was assumed that all parts were isotropic, homogenous and linear elastic, although the analysis was non-linear due to the contact modelling. The material properties used for each part are shown in Table 1. A number of simplifications have been made in carrying out this study. These include a uniform thickness of cortical bone and the homogeneity of the cancellous bone. These factors are the subject of current research where the FE model has been reconstructed from CT images and the effect of these simplifications will be assessed when this ongoing research is completed. The modulus of the cortical bone has been used by Nyman et al [9] and is within the range of moduli cited and used by other authors. The value of the cancellous modulus used is within the range of values (389-1132 MPa) cited by Au et al [10] for cancellous bone and obtained experimentally. The value of the cortical bone thickness (2 mm) was based on advice from the senior clinical author and is consistent with an image of a tibia sectioned for a TKA. The anatomical thickness increases distally but it was felt important to reflect the thickness in the proximal region of the tibia.

The block models discussed are hereby referred to as the standard stem metal (SM), press-fit stem extension metal (PM), standard stem cement (SC) and press-fit stem extension cement (PC) block models. All elements used were quadratic solid 10 node tetrahedral elements. A mesh study was undertaken to ensure a sufficient number of elements were used in the models to ensure convergence of displacement and stress distributions. The numbers of elements used in the two block models are listed in Table 2. A typical mesh can be seen in Figure 4. Although a direct comparison was not made between predicted and measured bone strains in this study such a comparison has been undertaken by Completo et al [12]. The mesh used in the current study, employing quadratic elements, is more accurate than that used by Completo et al [10] who only used linear elements. This provides further validation of the mesh used in the current study.

The contact between the mobile bearing and tibial tray was modelled using a surface-to-surface contact algorithm and a constant coefficient of friction of 0.12 in all models as used by Villa et al. [11] on exactly the same mobile bearing tray implant system. In their study they undertook detailed experimental in-vitro measurements of the contact area and pressure distribution which showed excellent correlation with the associated FE modelling with a coefficient of friction of 0.1. Relative movement was allowed between the press-fit stem extension contacting surfaces (i.e. stem extension and bone) and this was also modelled using a surface-to-surface contact algorithm and a constant coefficient of friction of 0.25 as used by Completo et al. [12]. To support this value Completo et al. [12] cite experimental measurements of this value and further show an excellent regression fit between in-vitro measured and FE calculated tibial strain distribution. No other relative motion was allowed between adjoining parts as tied contact was applied to these. Additionally tied contact was

employed between the stem extension and bone in the case of the tied stem extension model only i.e. in the first set of models.

Loading was applied as a uniform pressure load to selected surfaces of the bearing where the medial and lateral femur condyles would make contact. The load application region (see Figure 1) and values used were based on the conditions that occur in late stance phase of gait where maximum joint reaction occurs [13]. This is the phase of the gait that will produce highest stresses in the proximal bone and the augment. The gastrocnemius muscle is the only active muscle at this late stance phase of gait. As the gastrocnemius muscle does not attach to any region of the proximal tibia, it was not necessary to include any ligaments or muscles in the models. The effect of the gastrocnemius however is represented in the applied joint reaction force. Traditionally the loads applied to the knee to represent level gait in FE modelling have been between 2.5 to 3 times body weight [2, 10, 12] These data have been based on developments of the knee biomechanics developed by Morrison [13]. More recently, somewhat lower levels of loading (2.2 times body weight) have been actually measured in-vivo [14] and it seemed more appropriate to use these loads. Thus a total load of 1618 N (representing a 75 kg person) was used in this study. In all models the distal end of the tibia was assumed to be constrained in all directions (see Figure 1). The pressure load application areas were determined from the work of Villa et al. [11] who evaluated contact locations using Fuji Prescale pressure-sensitive films and in-vitro TKR models. These pressure distributions were fairly consistent with the distributions computed from the in-vivo data by Zhao et al [14] at the same phase of the gait cycle.

In order to verify the load transfer between the different components, the axial loads carried by each component were plotted at 4 mm increments from the proximal bone cut. These

values were plotted for each component in all models. The equivalent stress in the cancellous bone was examined for all models to determine if the use of cement to fill proximal defects was likely to result in the increased likelihood of bone failure due to increased stresses. A critical level for this stress was taken to be 2.8 MPa (equivalent to around 4000 $\mu\epsilon$) [15]. This is a very conservative value, being amongst the lowest cited in the literature. Cancellous bone stresses were also examined to ensure that reduced stresses do not lead to severe bone resorption with a resorption threshold of 0.1 MPa (equivalent to around 150 $\mu\epsilon$) adopted [15]. Also the stresses in the augment parts (cement and metal) were examined to determine how these stresses compare with those of their material fatigue strength values as shown in Table 3.

3. Results

Figures 5 and 6 show the load transfer in each component in the press-fit and tied stem extension models respectively. Results show that the cement carries the least amount of load in both models and this is not significantly affected by the tied contact conditions. The loads in the cortical and cancellous bone are more significantly reduced with the use of the tied stem extension (increasing the load taken by the metal stem) as the load is shifted from the cortical bone to the stem. Moving distally the reduced level of load in the cancellous bone with the use of a tied stem extension is indicative of the increased stress-shielding effect compared with the press fit stem extension. This increased stress shielding will have an adverse effect on implant stability and thus further studies with the tied contact extension were discontinued. The use of a press-fit stem extension with tied contact represents the contact conditions that would occur with the use of a cemented stem extension. Based on this work it is likely that the use of a cemented stem extension will have adverse implications on load transfer and stress-shielding in the cancellous bone. This provides a further reason for

press-fit stem extensions being the preferred option if an extension is required. Thus it was decided to continue only with the more acceptable press-fit stem extension for the block augmented models.

Figures 7 and 8 show the axial loads carried by each component in the 4 configurations of the block models analysed in the second group of analyses. Proximally, the load taken by the stem is approximately three times that of the cancellous and cortical bone which carry similar loading (e.g. Figure 7). The main reason for the low level of bone stress proximally is because the bone load carrying area is reduced by the presence of the augment (compare with the non-augmented model in Figure 5). Moving distally in the non-stem extension models (Figure 7) more load is taken by the cortical bone, being transferred from the block and the stem. The level of loading taken by the cancellous bone gradually decreases and then increases (as the load in the stem tip transfers to the bone) but remains significantly less than the cortical bone.

Proximally the use of the stem extension reduces the load taken by the cortical and cancellous bone a little, while increasing the load taken by the stem (compare Figures 7 and 8). This compares favourably with the work of Completo et al., 2008 [18]. When comparing the SM model with the SC model (see Figure 7) it can be seen that the load taken by the metal block is significantly larger than that carried by the cement block. This is because load transfers preferentially along a stiffer path. The loads taken by the cortical bone, and cancellous bone were also higher, but not significantly, in the SM model, with a commensurate decrease in the load carried by the stem.

Figure 9 shows the von Mises stress distribution for the metal and cement blocks for all four block models. The legends have been adjusted so that the top band shows 10% of the fatigue

strength for the metal augment and 100% of the fatigue strength of the cement augment. It can be seen that the stresses in the metal augment hardly exceed 10% of the fatigue threshold of the titanium alloy (510 MPa) whilst the stresses in parts of the cement augment **are approaching** 100% of the fatigue threshold of the cement (8.1 MPa). Thus the cement augments are likely to undergo fatigue damage during the lifetime of the implant. The use of a stem extension does not significantly affect the stresses in the cement and metal block components. This modelling was repeated with augments of only 10 mm depth. The augment stress distribution remained the same while the level of stresses reduced only marginally. Thus even the reduced thickness cement augments were operating at loads approaching the fatigue threshold of the cement and so would be susceptible to fatigue damage.

Figures 10 and 11 give two views of the cancellous stresses in the 4 models. The regions where the stresses lie in the bottom band contain cancellous bone with stresses of less than 0.1 MPa and are thus likely to be prone to resorption [15]. Looking at Figure 10 it can be seen that more bone resorption (stresses less than the assumed resorption threshold of 0.1 MPa) takes place on the upper resected surface in the cement augment models (**Fig 10 c,d**). However, moving distally and considering the augmented side, Figure 11 shows that the cancellous bone stresses become higher in the cement block models compared with the metal block models. This is possibly because the stem in the cement augment models is more highly loaded (as discussed above and seen in Figure 7) and thus transfers more load through the cancellous bone to the cortical bone moving distally. However, the decrease in the cancellous bone stress when using a metal augment are very modest and so should not significantly affect resorption. The reduction in stresses that occur when using a stem extension are more pronounced, see Figure 11, and occur on both the medial and lateral sides.

Thus, this provides further indication that the use of the stem extension will increase stress shielding and promote bone resorption. As mentioned above, when discussing the augment stresses, this modelling was repeated with a reduced augment thickness of 10 mm. It was found that the cancellous bone stress distribution was similar to the 15 mm augment models but that the stresses were marginally lower. Thus the possibility of a stem extension promoting bone resorption is marginally higher.

As discussed in the “Materials and Methods” section a figure of 4000 $\mu\epsilon$ (equivalent to 2.8 MPa in this work) provides a very conservative measure of possible cancellous bone damage by overload with cited values ranging from this value [15] to values in excess of 20,000 $\mu\epsilon$. It can be seen that the stresses in the cancellous bone of both models without a stem extension do not reach even this conservative value (and will even be marginally lower with the 10 mm augment) . Thus it can be concluded that using augments as modelled here do not give rise to stresses that are likely to cause initial instability due to cancellous bone overloading. With a stem extension, stresses that exceed this (conservative) overloading threshold are noted in a zone of bone at the tip of the extension. This provides yet another reason not to use a stem extension unless required to mitigate some other problem.

4. Discussion

Loosening and migration of tibial prostheses have been identified as primary causes of early TKR failure. The probability of loosening may be increased due to defects in the bone in the proximal tibia as this makes it more difficult to achieve the correct fixation and alignment of the prosthesis. Clinical studies using metal augmented features (i.e. wedge, block and conical) have shown these to be a viable alternative to other means of defect treatment such as bone graft and the use of cement. It is possible that regions of high stress and/or stress

concentrations in the cancellous bone close to the prosthesis region may be prone to loosening in the TKR due to overloading. FE analysis can be used to identify such regions and to advise surgical teams on an appropriate course of action when dealing with defects. A review of current literature on defect treatments shows an absence of research into FE modelling of defect treatments in 3D initial stability studies. The objective of this study was to determine the best course of action when treating uncontained type-2 tibial defects. The above models contained defects on the medial side only. The use of cement and metal augments was examined along with the use of stem extensions.

Using cement to fix a stem extension to the adjacent cancellous bone results in significantly increased load transfer to the stem with commensurate reductions in cancellous and cortical bone loads. This is likely to produce excessive stress shielding and hence such a fixation technique for a stem extension is not recommended.

The load distribution study showed that the use of cement to fill the peripheral defects has only minor effects on cancellous and cortical bone load sharing in all models. The use of metal augments to treat defects have shown better clinical survivorship rates in TKR compared to those treated with cement augments [19]. The use of cement to fill proximal defects of 10 and 15 mm depth is not considered to be suitable as stresses carried by the cement block exceed those of the cement fatigue limit. This is likely to result in failure of the cement augment and thus loosening of the prosthesis. Thus it is recommended that the use of cement to fill proximal defects should be confined to the treatment of shallow defects only. Also, more potential bone resorption was noted on the augmented bone surface in the case of the cement augments; this may further account for the greater stability provided by metal augments.

The use of stem extensions (press-fit stems) in conjunction with uncontained type-2 proximal defects may be an over-conservative approach in TKR surgery. The use of a press-fit stem does not cause any significant reduction in stresses on the upper resected surface of the block augment in all models (Figure 11). However, moving further distally, cancellous stresses are significantly reduced with the use of a press-fit stem extension (Figure 10). Further, it was shown that the proximal cancellous bone stresses were not large enough to cause damage due to overloading and thus the stress reducing effect exhibited by the stem extension would not be required. Thus it is recommended that in cases where defects are less than 15 mm on one side only that a tibial tray with a standard stem is adequate for defect treatment. However, in cases of patients presenting with poor proximal bone stock a press-fit stem may be beneficial.

In clinical practice it is possible that a patient may present with additional defects such as central defects at the time of surgery. Such configurations have not been modelled and thus the results presented above may not be sufficient for all patients receiving a block augment. In such cases patient-specific FE models of TKR treated with block augments would be necessary to establish if a stem extension is necessary.

The 3D models detailed in this paper did not include the effect of ligament attachments and their effect on stress patterns. Models will be expanded in future work to examine the effect of loading at different stages in the gait cycle and include the influence of the patellar ligament (active in other stages of gait) which is considered to have the greatest effect on tibia loading compared with other ligaments [19]. In addition the TKR models will also need to be created in patient specific models as the results found in this study were created on ideal bone geometry (i.e. isotropic, homogenous and elastic). From patient-specific data, moduli of

various regions can be identified and an anisotropic, heterogeneous model produced and analysed. Other assumptions in the models include quasi-static and symmetrical condylar loading and tied contact between some contacting surfaces. However as the focus of this study was a comparison between the different types of augments only, these assumptions are unlikely to have affected the outcome.

5. Conclusion

The use of stem extensions in conjunction with proximal defects may be an over-conservative approach in revision TKR surgery. It is recommended that in cases where defects are less than 15 mm on one side only that a tibial tray with a standard stem is adequate. This is because the proximal cancellous bone stresses do not reach even the very conservative overload threshold used and use of a stem extension would increase the tendency for bone resorption. Additionally cement fixation of a stem extension, if used, may not be appropriate as it may lead to a further increased risk of stress shielding. The use of cement to fill proximal defects of 10 mm and 15 mm is not considered to be suitable as stresses carried by the cement block exceed those of the fatigue limit of the cement. This is likely to lead to instability on the medial (augmented) side of the tibial tray.

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Table 1 Material properties used in TKR model [9].

Material	E (GPa)	ν
PMMA (Cement)	2.27	0.46
Ti6Al4V (Tibial Component and Stem)	110	0.3
Cortical Bone	17	0.3
Cancellous Bone	0.7	0.3
UHMWPE (Bearing)	2.3	0.25

Table 2 Number of elements used in each block model.

Material	SM	PM	SC	PC
Top Cement	5,029	5,029	5,029	5,029
Bearing	19,281	19,281	19,281	19,281
Prosthesis	10,515	10,515	10,515	10,515
Block (Metal or Cement)	15,186	15,186	33,383	33,383
Cancellous	96,359	92,013	96,359	92,013
Cortical	27,189	27,189	27,189	27,189
Lower Cement	22,909	21,964	22,909	21,964

Table 3 Fatigue strength values of materials used

Material	Fatigue Strength Value (@10 ⁷ cycles)
Ti6Al4V	510 MPa [16]
Bone Cement	8.1 MPa [17]

List of Figures

Fig. 1 Tibial Plate / Bone Assembly showing pressure load applied and constraints

Fig. 2 Block Component

Fig. 3 Section cut of TKR block model with (a) standard stem and (b) press-fit stem extension

Fig. 4 TKR meshed block model

Fig. 5 Load distribution along the z-axis in the press fit contact stem extension model

Fig. 6 Load distribution along the z-axis in the tied stem extension model

Fig. 7 Load distribution along the z-axis for each material in the SM and SC models

Fig. 8 Load distribution along the z-axis for each material in the PM and PC models.

Fig. 9 Von Mises stress distribution in the block for (a) SM (b) PM (c) SC and (d) PC models.

Fig. 10 Von Mises stress distribution of the cancellous bone for (a) SM (b) PM (c) SC and (d) PC models.

Fig. 11 Von Mises stress distribution of the cancellous and cortical bone for (a) SM (b) PM (c) SC and (d) PC models.

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(c)

(b)

(d)

Fig. 10 Von Mises stress distribution of the cancellous bone for (a) SM (b) PM (c) SC and (d) PC models.

Fig. 11 Von Mises stress distribution of the cancellous and cortical bone for (a) SM (b) PM (c) SC and (d) PC models.