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RESEARCH ARTICLE

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The effect of periprosthetic bone loss on the failure risk of tibial total knee arthroplasty

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Abstract

The effect of long-term periprosthetic bone loss on the process of aseptic loosening of tibial total knee arthroplasty (TKA) is subject to debate. Contradicting studies can be found in literature, reporting either bone resorption or bone formation before failure of the tibial tray. The aim of the current study was to investigate the effects of bone resorption on failure of tibial TKA, by simulating clinical postoperative bone density changes in finite element analysis (FEA) models and FEA models were created of two tibiae representing cases with good and poor initial bone quality which were subjected to a walking configuration and subsequently to a traumatic stumbling load. Bone failure was simulated using a crushable foam model incorporating progressive yielding. Repetitive loading under a level walking load did not result in failure of the periprosthetic bone in neither the good nor poor bone quality tibia at the baseline bone densities. When applying a stumble load, a collapse of the tibial reconstruction was noticed in the poor bone quality model. Incorporating postoperative bone loss led to a significant increase of the failure risk, particularly for the poor bone quality model in which subsidence of the tibial component was substantial. Our results suggest bone loss can lead to an increased risk of a collapse of the tibial component, particularly in case of poor bone quality at the time of surgery. The study also examined the probability of medial or lateral subsidence of the implant and aimed to improve clinical implications. The FEA model simulated plastic deformation of the bone and implant subsidence, with further validation required via mechanical experiments.

KEYWORDS

bone mineral density, bone resorption, finite element analysis, medial tibial collapse, total knee arthroplasty

1 | INTRODUCTION

While total knee arthroplasty (TKA) is a popular and successful orthopaedic intervention, it does not provide a lifelong solution yet, particularly for younger patients, with aseptic loosening being one of the main reasons for revision.[1] Aseptic loosening is a complex multifactorial process, in which loss of mechanical fixation plays an important role.[2]

The mechanical integrity of the reconstruction is dependent on the quality of the periprosthetic bone supporting the implant. The bone quality can change after TKA due to changes in the bone stresses after placement of the metal components, resulting in changes in bone density, commonly referred to as bone remodeling.[3] Reports have shown that after TKA the bone mineral density (BMD) decreases underneath the tibial implant, particularly in the medial compartment.[4–6] Bone loss in this region may weaken the support of the tibial baseplate, which in extreme cases can lead to tibial collapse.[7] While the overall failure rate of TKA due to tibial collapse is low (0.3%–0.6%),[8–10] problems with bone loss may become more apparent with the expansion of the indication for surgery, with younger patients being operated on that will have their implant for a longer period of time.

Several studies examined the relation between bone quality (changes) and component migration.[4,11–13] Linde et al.[12] found minor changes in migration after an initial settling period as measured using RSA techniques, and a decreasing BMD after an initial phase of bone densification, as measured using DEXA. That study was unable to find a correlation between the maximum total point motion and the BMD changes. The potential impact of periprosthetic bone loss on TKA failure have been recognized in previous literature[4,6,8,10] which illustrates that understanding of this relationship is crucial for improving the long-term success of TKA. Among the previous literature, many studies have shown that the failure risk in the tibial implant following TKA is more associated with bone resorption than other factors and demonstrated that a decrease in BMD led to subsidence of the implant. Conversely, Ritter et al.[10] and Bergink et al.[14,15] by studying more than 1500 fracture cases reported an increase in BMD before failure of TKA. this seems counterintuitive to be accounted for tibial implant failure, as it is more likely that a tibial implant would collapse due to weakened bone strength. This illustrates a need for clarification of the effect of bone resorption on TKA survival. The exact circumstances under which a tibial collapse occurs are largely unknown and not well documented in the literature. While TKA reconstructions are perfectly able to withstand the forces acting on the knee joint during daily activities (e.g., gait, stair climbing, rising from a chair),[16] more excessive loads that occur during sudden impact events such as stumbling may lead to permanent damage and perhaps even a tibial collapse and subsequent gross failure of the reconstruction.[9,16]

Computational modeling using finite element analysis (FEA) can assist in assessing the failure risk of the reconstruction.[17] In designing and evaluating orthopedic implants, it is crucial to take into account the potential for plastic deformation of cancellous bone, which can result in implant migration and loosening. Understanding the bone response to both cyclic and static loading is important, as fatigue and large deformation effects both can contribute to implant failure.[18–20] However, this requires a suitable material model that realistically predicts the plastic failure processes occurring in the supporting bone. Our group recently developed an isotropic crushable foam (ICF) constitutive model capable of simulating the failure of tibial bone that can be used for such analyses.

The aim of this study was to investigate the relationship between bone resorption and the risk of failure of the tibial reconstruction in TKA and to evaluate if FEA can simulate this relationship. Specifically, the study examined the effect of clinically relevant postoperative changes in BMD[21] on the risk of failure of tibial TKA, through the use of FEA models subjected to both physiological loading and a simulated stumbling event. The hypothesis of the study was that bone resorption following TKA increases the risk of tibial reconstruction failure, and that FEA with combination of using a proper material model can be used to predict this risk.

2 | METHODS

2.1 | Three-dimensional models

Two proximal tibias of a male (65 years, 85 kg) and female (90 years, 65 kg) subject representing cases with good and poor initial bone quality, respectively, were QCT scanned (Toshiba Medical Systems, Tokyo, Japan; voxel size $0.6 \times 0.4 \times 0.4$ mm, 120 kV, 260 mA). The tibiae were scanned along with solid calibration phantoms (Image Analysis) to compute local BMD.[22] Three-dimensional models of the proximal tibiae were created in 3D Slicer (Open Source Software[23]). Optical scans of various sizes of the Triathlon tibial component (manufactured by Stryker Orthopaedics) were provided. The appropriate size of the implant for each patient was selected based on anthropometric measures, and a CAD model of the selected implant was created. A cement layer with a thickness of 2 mm was created underneath the tibial tray.[24] The geometry of the proximal tibia was resected following mechanical alignment surgical guidelines, to ensure the tibial component is implanted in a neutral position, meaning the resected plateau of the tibia was perpendicular to its mechanical axis (Figure 1A).

2.2 | Finite element model

Three-dimensional models of the implanted proximal tibias were meshed using four-noded tetrahedral elements in Hypermesh (Altair Engineering) and imported into the FEA software (Marc/Mentat 2021, MSC. Software Corporation). The bone was considered as a heterogeneous elastoplastic material in which the mechanical properties of each element were computed based on the calibrated BMD values. Elastic material properties were assigned to the Cobalt-Chromium (CoCr) tibial tray and the polymethylmethacrylate (PMMA) cement layer (Table 1). A fully bonded interaction was considered between bone and cement, and between the implant and cement, while a frictional touching contact with coefficient friction of 0.4 was assumed between the bone and keel of the tibial component.[24] A mesh convergence study was performed using the intact tibial bone with element sizes in the range of 4.5, 4, 3, 2.5, and 2 mm to define the appropriate element size. For element sizes smaller than 3 mm, the difference in strain energy was less than 10%[26] which was selected as the target size. Numerical simulations were performed

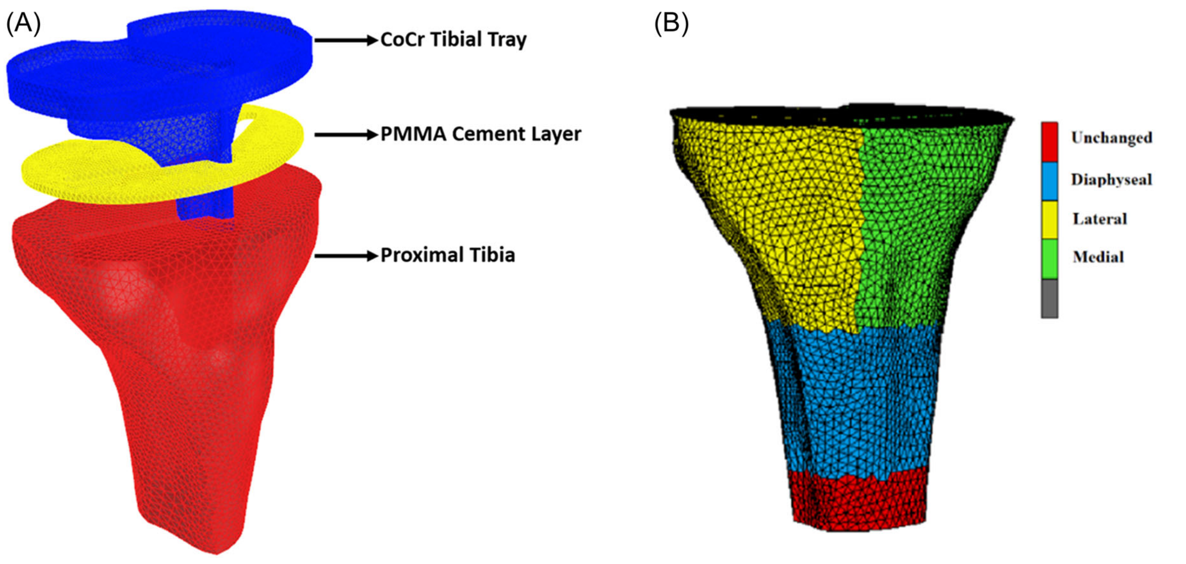


FIGURE 1 A schematic view of the proximal tibia, cement layer and implant (A) and selection of the regions of interest in the computer model (B).

TABLE 1 Mechanical properties of the materials used in the FEA models.[17,25]

Nonlinear elastoplastic behavior dependent on BMD (mg/mm^3)			
Material	ICF Young's modulus	Yield stress	K parameter (strength ratio)
Trabecular bone (BMD ≤ 0.95)	$5113\rho_{BMD}^{1.654}$	$79.36\rho_{BMD}^{1.553}$	$K = 2.993$, BMD ≤ 0.08 $K = 1.361\rho_{BMD}^{-0.312}$, 0.08 < BMD ≤ 0.95
Cortical bone (BMD > 0.95)	$13750\rho_{BMD}^{2.429}$	$111.5\rho_{BMD}^{1.800}$	$K = 1.383$, BMD > 0.95
Linear elastic behavior			
Material	**Young modulus (MPa)**		**Poisson's ratio**
CoCr	210,000		0.3
PMMA	2100		0.3

using a nonlinear elastoplastic ICF model (Table 1; detailed information about this ICF model can be found in study by Soltanihafshejani et al.[17]

To simulate bone resorption in a medium-term follow-up TKA, the equivalent BMD values of the tibiae were adjusted according to clinical follow-up data. The change in BMD of 69 preoperatively varus-aligned knees was reported by Jaroma et al.,[21] during a 7-year follow-up period after TKA (Figure 2). The BMD values were reported for three regions of interest (ROIs) in the proximal tibia (medial, lateral, and diaphyseal), at surgery (0 months), and 3 months, and 1 and 7 years after surgery. The change in BMD was applied accordingly to the trabecular regions of the QCT-based FEA models (BMD ≤ 0.950 mg/mm^3). To represent the worst-case scenario the minimum density value at each time point was selected for analysis. Figure 1B shows the representation of the ROIs in the FEA model.

To simulate the permanent deformation of the tibiae due to a stumbling event, first, the model was preconditioned through cyclic loading with a gait loading regime. The repetitive loading was continued until hysteresis in the model was minimized and a steady state of plastic deformation was reached.[18,27,28] To determine the number of cycles required to achieve a steady state, the load was repeated for 200 cycles. If a change in the number of yielded integration points deviated less than 1%, the number of loading cycles was considered sufficient. After the preconditioning phase a stumble load (biased at the medial side) was applied to the model, representing an incident with excessive but still physiological loads. To confirm the model's ability to accurately account for deviations in incident loads, we adjusted the parameter for stumbling load from medial to lateral. This adjustment was made to represent the scenario where an unexpected event affects the lateral side. The gait and stumble loads were applied to a point located centrally at a level 10 mm above the tibial tray (Figure 3A), following the origin of the Orthoload data set.[29] The central loaded point was linked to the tibial surface tray based on the projection of the distal femoral condyle onto the tibial tray. Table 2 shows the force and moment

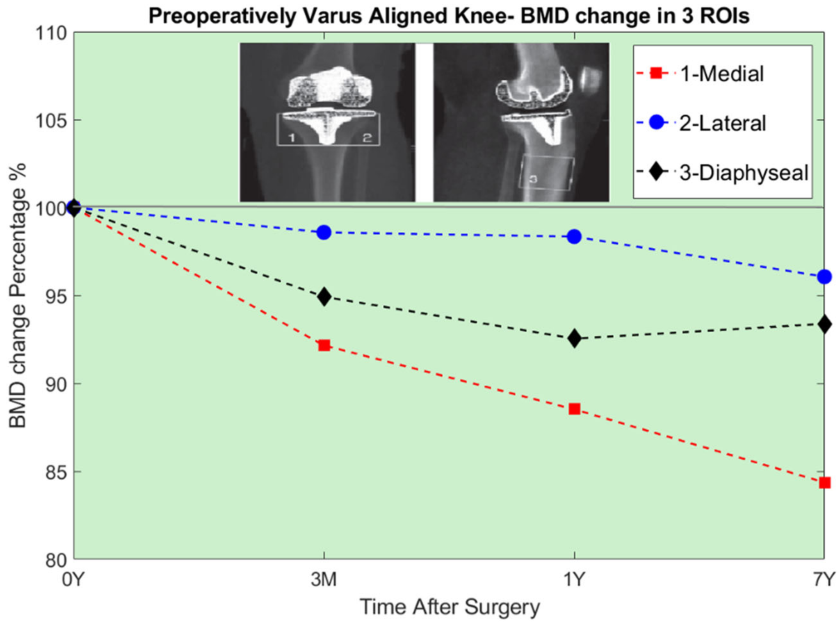


FIGURE 2 Percentage change in BMD in implanted proximal tibia. Adapted from Jaroma et al.[21]

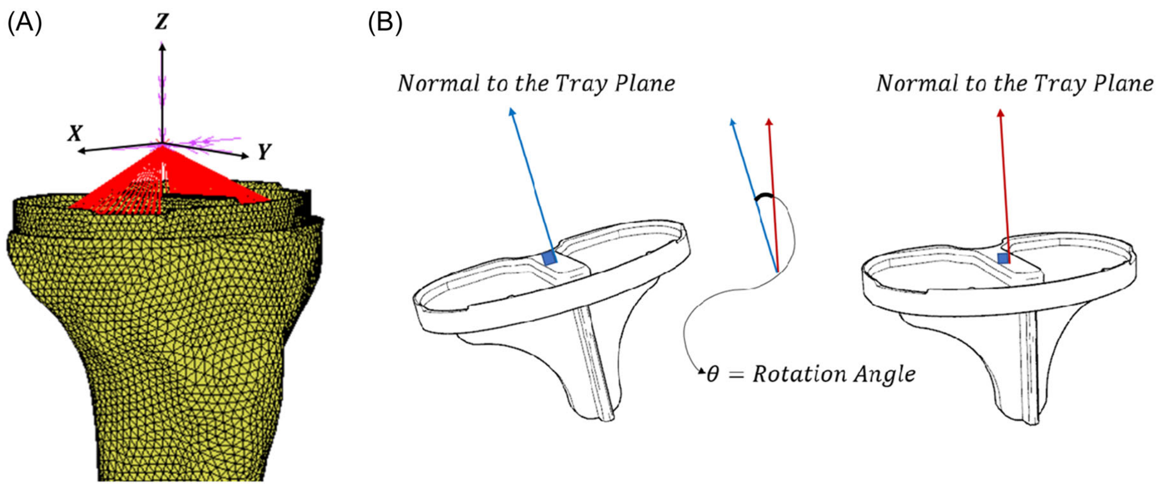


FIGURE 3 (A) Location of the center point above tibial tray in local coordinates and (B) rotation angle of the tibial tray.

TABLE 2 Applied loads to the center point above the tibial tray.[16,29]

Applied loads						
Load	**F_x (BW)**	**F_y (BW)**	**F_z (BW)**	**M_x (BW. mm)**	**M_y (BW. mm)**	**M_z (BW. mm)**
Gait	−0.1	−0.15	−2.61	19	−10	−5
Stumbling (medial biased)	−0.346	−0.519	−9.03	65.74	−34.6	−17.3
Stumbling (lateral biased)	−0.346	−0.519	−9.03	65.74	+34.6	−17.3

components for both loading conditions. For gait, the peak load was taken based on standard loads acting on the knee,[29] while for the stumbling load, these values were multiplied by a factor of 3.46, based on measurements of the hip joint during an actual stumbling event.[16] The distal part of the tibia was fixed in all directions. As a measure of failure of the tibial component, the angular deviation of the normal vector of the tibial tray from the initial orientation was calculated after unloading the model and taken as the tibial migration (Figure 3B). A threshold of 10° was considered a gross failure due to tibial collapse,[9] while a rotation of the tibial component of less than 3° was assumed to be in the safe margin.[8]

3 | RESULTS

Applying the peak load for 200 cycles initially resulted in an increasing number of yielded nodes, followed by a steady state in which the percentage of yielded nodes increased only marginally. In the model with good quality bone (65-year-old, BMD 219 mg/mm^3), after 10 cycles the increase of yielded nodes was less than 1%, which reduced to 0.1% after 60 cycles (Figure 4A), which was similar for all postoperative time points (0, 3 m, 1 year, 7 years). In the model with weak bone (90 years old, BMD 130 mg/mm^3), a significant amount of plastic deformation was noticed. In this model, only after 150 cycles the incremental increase in yielded nodes was less than 1%, which was further reduced to 0.1% after 190 cycles for the weak bone (Figure 4B).

Based on the initial results, both the strong and weak tibia models were preconditioned with 150 gait cycles to ensure a steady state of plastic deformation in both cases before applying the stumbling load.

As shown in Figure 5, for both tibias no prominent migration occurred after the cyclic peak gait loading, at none of the simulated cases of postoperative bone loss. After application of the stumbling load biased toward the medial side, for the strong bone, the rotational migration of the tibial component remained less than 1° for all postoperative time points. However, for the weak bone, the tibial migration increased with postoperative BMD loss, ranging from 3.7° directly postoperatively to 15.2° at 7 years. To explore the effect of variations in the loading configuration, additional simulations were performed in which the stumbling load was biased toward the lateral

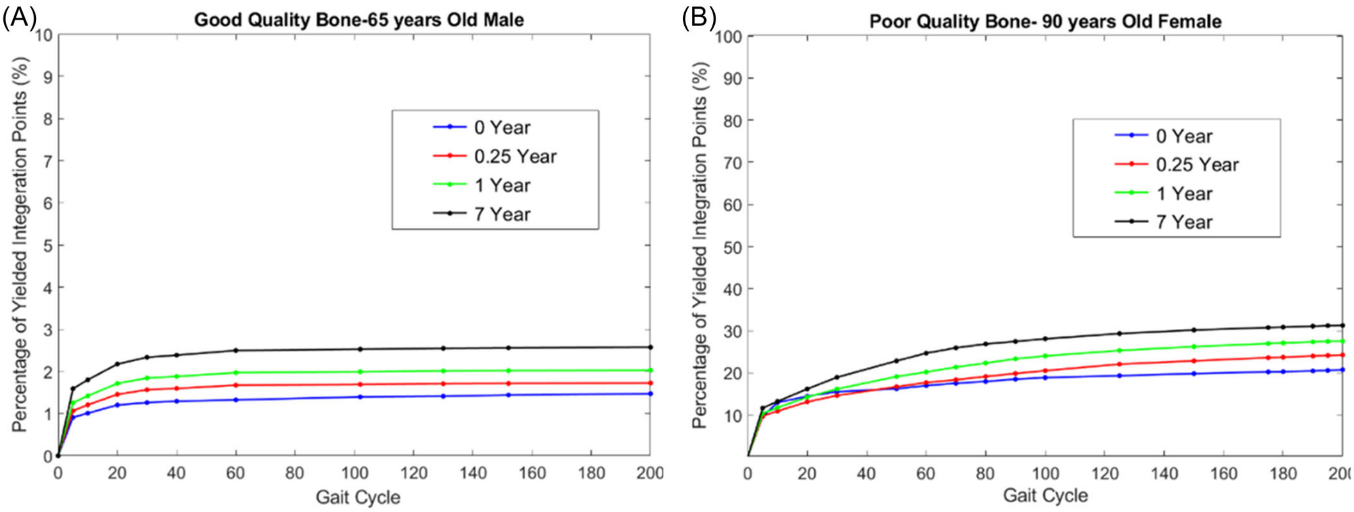


FIGURE 4 Percentage of yielded integration points for 200 cycles of gait peak load in a prealigned varus tibia of a 65-year-old male (A) and a 90-year-old female (B). Note the difference in the scale of the vertical axes.

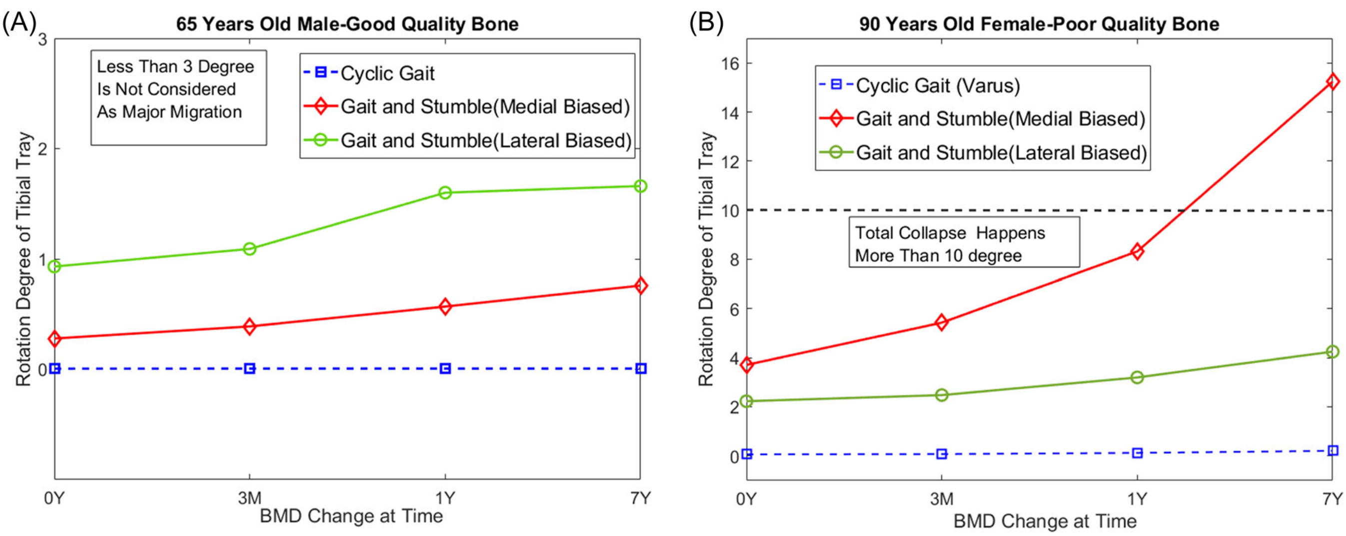
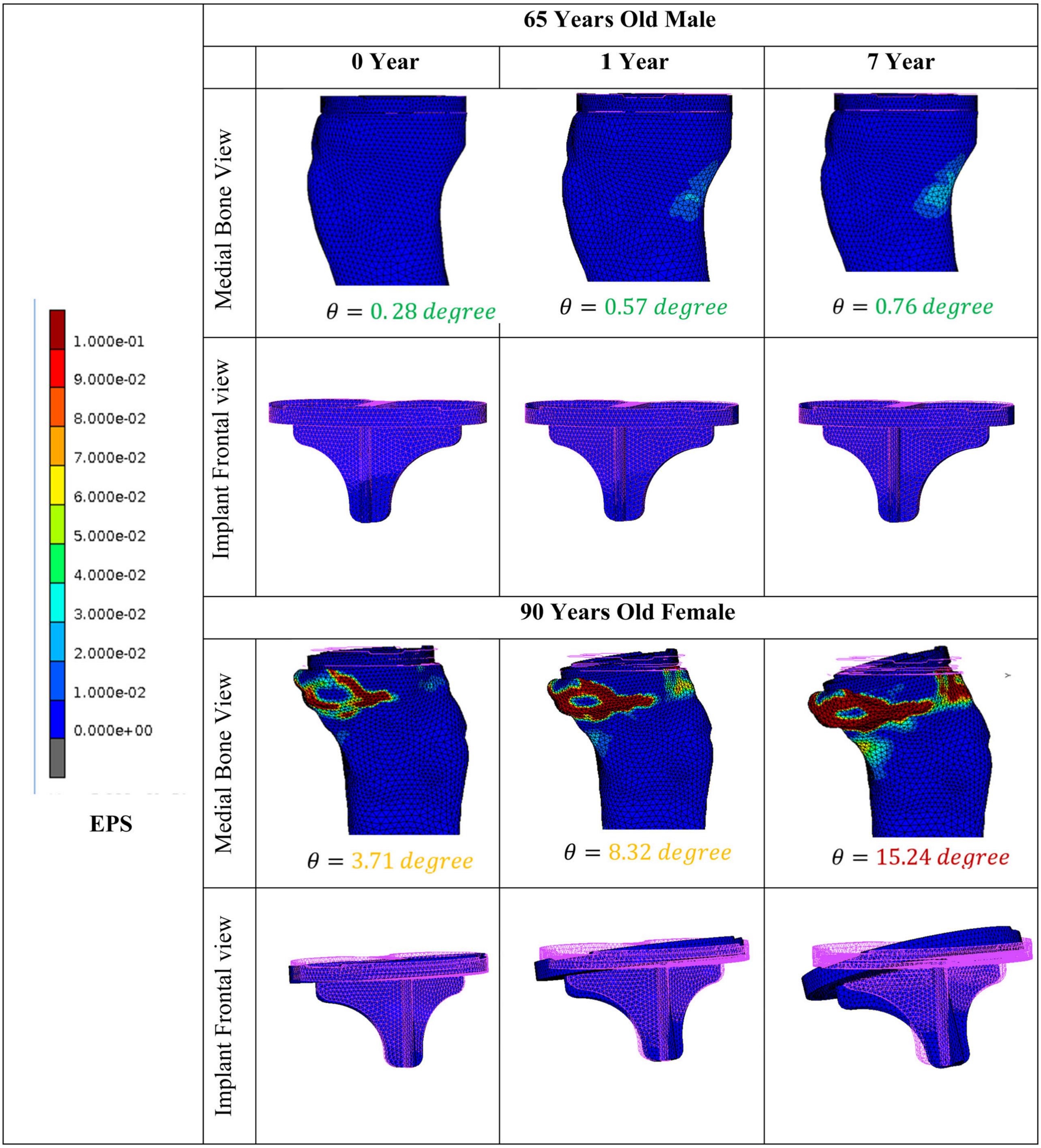


FIGURE 5 The rotation angle of tibial tray of (A) 65 year old male (strong bone) and (B) 90 year old female (weak bone). Note the difference in the scale of the vertical axes.

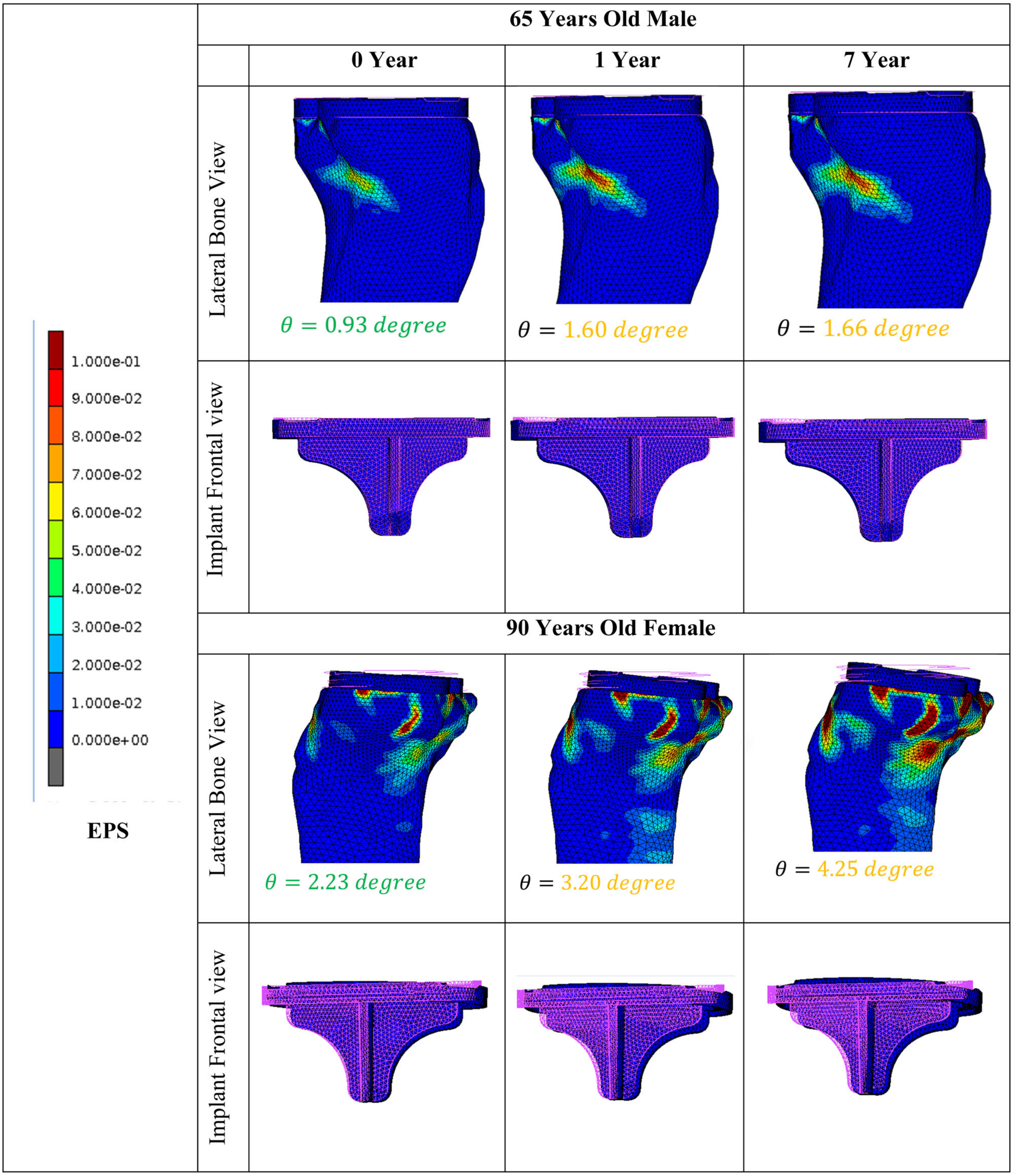
TABLE 3 EPS distribution at the medial compartment of proximal tibiae as demonstration of the permanent deformation in stumbling load biased toward medial side.



side. In the model with the 65-year-old male implant migration was between 1 and 2° for this load, and remained below the major migration threshold of 3°. For the model of the 90-year-old female, the rotational migration was between 2° and 4.3°, which was lower than for the medial load case and also below the threshold for total collapse (10°—Figure 5).

Table 3 shows the distribution of equivalent of plastic strain (EPS) after applying the stumble load, for both tibias with the simulated BMD loss at 0, 1, and 7 years. Peak plastic strains were mainly seen at the medial side. In the strong tibia (good quality bone) no plastic strain was seen directly postoperatively, while with a bone loss at 7 years postoperatively some plastic

TABLE 4 EPS distribution at the Lateral compartment of proximal tibiae as a demonstration of the permanent deformation after stumbling load biased on the lateral side.



deformation was noticed in the medial cortex. For the weak bone, substantial plastic strains were seen already directly post-operatively, which increased significantly with the postoperative time (and decreasing BMD).

Table 4 illustrates the distribution of EPS resulting from the application of a lateral-biased stumbling load to both tibias with simulated BMD loss at 0, 1, and 7 years. The largest plastic strains were observed at the lateral side. For the strong tibia with good-quality bone, some plastic strain was noticed right after the surgery, but there was no significant implant migration. In the case of weak bone, substantial plastic strains were observed directly after surgery, which increased as the postoperative time progressed (and BMD decreased). Although implant migration was seen, a total collapse (rotation larger than 10°) was not observed at any of the time points.

4 | DISCUSSION

The aim of the current paper was to investigate the effect of bone resorption on the failure of the tibial reconstruction in TKA. Bone density changes representing the clinical situation between 0 and 7 years postoperatively were simulated in two FEA models with good and poor initial bone quality. While in a good quality bone the periprosthetic bone loss did not lead to a substantial increase in implant migration, initial poor bone quality and additional simulated postoperative bone loss had a substantial effect on the stability of the tibial implant.

The distribution of plastic strains at medially biased stumbling load, showed that in the poor bone quality model the deformations mainly occurred in the medial compartment, with a migration that clinically would be interpreted as a medial tibial collapse. On the lateral side, bone failure was less pronounced, which may have been related to the reduced bone loss that was simulated in that compartment based on the clinical data,[21] and the reduced load on the lateral compartment. The stumbling load biased toward the lateral compartment compromised the reduced load on the lateral side. However, the results indicate that even if the loads are more biased toward the lateral side, total collapse may not occur laterally. This finding is consistent with tibial failure situations, which mostly occur on the medial side. The extent of tibial migration was maximal at 7 years postoperatively, in line with the progression of bone loss.

Periprosthetic changes to bone density after TKA have been investigated widely in clinical studies.[4–6,10,30] The study of Jaroma et al.,[21] provided a detailed description of these changes, allowing to investigate bone changes at different time points postoperatively. Although in the simulations the BMD change was only applied to the trabecular regions, the actual change may also partially involve the cortical bone. Moreover, the measured bone density changes were applied to relatively large regions of interest, while the density changes may be distributed differently over the ROIs.

While the current study shows that failure of the tibial tray may be related to periprosthetic bone resorption, Ritter et al.,[10] actually showed an increase in bone density in patients that experienced a medial tibial collapse, based on knee radiographs. Other studies have suggested that an increase in BMD may occur before the failure of TKA, independent of medial tibial collapse.[4,14,15] Those findings are also in contradiction with numerous DEXA studies that reported decreasing values of BMD before the collapse.[3,4,6,8,13,30,31] This discrepancy may be due to differences between radiographs and DEXA scans in the ability to measure BMD changes. One potential explanation could be that the participants in the study conducted by Ritter et al. had already undergone a gradual collapse of the tibia, which could have stimulated localized fracture repair reactions and led to increased levels of osteoblast activity. Alternatively, other studies have specifically examined the effects of osteoarthritis and osteoporotic bone which could have potential impact on the results.

The current simulations used an ICF model to capture the post-yield behavior of the tibiae.[17] As the ICF model incorporates an updating yield criterion, to avoid overestimation of the bone strength, a repetitive loading configuration was applied to model damage accumulation in the bone as a preconditioning treatment.[32,33] To ensure model equilibrium before the analysis of the stumbling event, a steady state of yielded bone was defined after cyclic loading using repetitive gait cycles. This allowed us to capture progressive yielding of the tibiae behavior, resulting in accumulated damage and failure with large plastic strains.[18,34]

To determine the appropriate number of repetitive cycles, both tibiae were subjected to 200 cycles of normal gait loading. While decreasing the BMD resulted in larger yield areas for both models, the level of steady-state plasticity remained almost unchanged for all the cases, both for the strong (around 10 cycles, Figure 4A) and the weak bone (150 cycles, Figure 4B). As the weak bone required 150 gait cycles to reach a steady state of plasticity, this number was used for both tibiae before applying the stumbling load. No distinct migration of the tibial tray was seen directly after the cyclic gait loading (for all postoperative time points), confirming the models predicted no risk of failure due to level walking.[16]

Applying the stumbling load biased toward medial side resulted in tibial migration in both the strong (0.3°–0.8°) and weak bone (3.7°–15.2°) (Table 3). Hence, for the strong bone the subsidence remained below 1° for all cases with no risk of tibial collapse. For the weak bone, however, tibial migration exceeded the 10° threshold at 7 years postoperatively, indicating a tibial collapse. At earlier timepoints the tibial migration was already substantial, as rotations of more than 3° (low-level migration) are considered hazardous for vulnerable patients.[8] To investigate the effect of loading variations on bone failure and implant migration the models were also loaded with a stumbling load with a lateral bias. In both models the migration of the tibial tray did not exceed the 10° threshold, demonstrating the sensitivity of the results to the orientation of the applied loads. This stresses the importance of analyzing the effect of including variability to make the predictions more robust. This may also include variations in muscle forces, as these may also significantly affect the joint loads. As early migration of the tibial implant can be an indication for increased risk of failure,[24] preoperative planning and selection of implant deign based on bone quality is essential, and may be informed by a computational model.

Several clinical studies[11,12,35] have shown an early phase of implant migration, which levels off after 1 and 2 years. In contrast, our FEA study did not reflect this pattern in the low-bone quality model. This discrepancy could be attributed to several factors. First, clinical measurements (e.g., radio stereometric analysis) provide a direct assessment of implant migration in vivo, including an initial settling phase that may be influenced by factors such as level of activity, rehabilitation regime, or biological response to the surgery, which are not included in our mechanical model. Moreover, we simulated a case of extreme bone loss in a subject with an already poor baseline bone quality. As such, our model may be more representative of an individual case that has an elevated risk of failure, rather than a representative case of the majority of patients undergoing a TKA procedure. Therefore, broader investigations with a larger number of models reflecting the actual patient population (including its variability) are required to further explore the mechanisms underlying early migration of tibial implants and the role of bone quality.

Our study has several limitations that should be considered. First, the FEA models were not validated against experimental results. The material model used in our study, however, was previously validated against experimental findings of human trabecular bone[17] and now extended with the cortical bone data adapted from Kaneko et al.[25] Second, the change in BMD applied to the models was not related to the implant used in the current study but was based on clinical data of 86 patients implanted with four different prostheses. Hence, BMD changes could be different for this particular implant and tibiae used in the current study. Third, we reported the angular deviation of the tibial components, which may not be fully representative of tibial migration. However, angular measurements in the anatomical planes are widely accepted in clinical studies.[9] Finally in this study, a constant percentage of BMD change was used for all regions of interest (RIOs) since the data source did not provide localized information. While this may not capture localized changes in BMD, the selected RIOs in the study of Jaroma et al.,[21] were chosen as representative areas for a general overview of BMD changes. In future work, we intend to extend the simulations to a larger population of models to obtain more robust insights into the failure risk of tibial TKA.

5 | CONCLUSION

In conclusion, the current study investigated the effect of periprosthetic bone resorption on failure of a tibial TKA using a bone material model simulating progressive yielding. The study demonstrates the feasibility of simulating the collapse of the tibial reconstruction in TKA after long-term periprosthetic bone loss. A stumbling load triggered the failure process of the reconstruction in case of poor initial bone quality and was even more pronounced with additional simulated postoperative bone loss.

While the findings of our study cannot be directly extrapolated to other implant designs, the approach presented here can be used to evaluate TKA components. Our study provides insights into the potential failure mode of TKA implants due to periprosthetic bone loss, which can inform the implant design process and the selection of implants tailored to specific patient groups. From a clinical perspective, our study highlights the importance of monitoring and managing periprosthetic bone loss in TKA patients to ensure long-term implant survival. Ultimately, our research aims to improve clinical outcomes and enhance the quality of life of TKA patients.

AUTHOR CONTRIBUTIONS

Navid Soltanihafshejani: Conceptualization, Methodology, Software, Validation, Formal analysis, Investigation, Data Curation, Writing—Original Draft, Review and Editing, Visualization. **Thom Bitter**: Conceptualization, Methodology, Software, Review, and Editing. **Nico Verdonschot**: Conceptualization, Methodology, Resources, Writing—Review and Editing, Supervision. **Dennis Janssen**: Conceptualization, Methodology, Writing—Review and Editing, Supervision, and Project administration. All authors have read and approved the final submitted manuscript.

CONFLICT OF INTEREST STATEMENT

The authors declare no conflict of interest.

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