Stress Shielding in the Proximal Tibia after Total Knee Arthroplasty: A Finite Element Analysis of 2- and 4-mm-thick Tibia Prosthesis Models

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Abstract

Objective: Proximal tibial bone resorption and osteolysis after total knee arthroplasty occur despite improved design and manufacturing processes. We utilized finite element analysis to study these phenomena.

Methods: We use SOLIDWORKS 2018 software to study stress and displacement of normal tibias and tibias implanted with a 4-mm-thick CoCr tibial tray (4mm-tray) or 2-mm-thick titanium alloy (2mm-tray). Under vertical loads of 1000 or 2000 N, the stress and displacement of both tibia tray models were analyzed. Stress on the supported proximal tibia 1 and 2 cm beneath the surface was analyzed and compared to stress in a normal tibia.

Results: Stress concentrated around the central region compared to the peripheral region in all models, which caused more deformation of the material in the central region. However, the 4mm-tray exhibited a more rigid construct compared to the 2mm-tray. Under any load, the 2mm-tray exhibited more tray deformation, with a central–peripheral deformation difference of approximately five times more than the deformation difference for the 4mm-tray. Moreover, stress on the peripheral region of the supported proximal tibia was only 18–22% of that of a normal bone for the 4mm-tray compared to 54–66% for the 2mm-tray.

Conclusion: Both tibial tray implant models exhibited some degree of stress shielding on the peripheral region of the supported proximal tibia. However, the greater modulus and thicker baseplate construct of the 4-mm CoCr tray exhibited a profound stress shielding effect. This stress shielding may correlate with a higher incidence of proximal tibia bone loss.

Keywords: finite element analysis, total knee arthroplasty, deformation of tibial tray, stress shielding

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การลดลงของความเครียดในกระดูกภายหลังการผ่าตัดเปลี่ยนข้อเข่าเทียม; การศึกษาโดยใช้ระเบียบวิธีไฟไนต์เอเลเมนต์ โดยใช้ข้อเข่าเทียมส่วนแผ่นทิเบียหนา 2 และ 4 มิลลิเมตร

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บทคัดย่อ

วัตถุประสงค์: ภาวะการณ์สลายและดูดซึมกลับของกระดูกทิเบียส่วนต้น (proximal tibial bone resorption and osteolysis) ภายหลังการผ่าตัดเปลี่ยนข้อเข่าเทียม เป็นผลมาจากการลดลงของความเครียดในกระดูก (stress shielding) โดยภาวะนี้ยังคงเกิดขึ้นแม้ว่าจะมีการพัฒนาเครื่องมือและกระบวนการผลิต ผู้ประพันธ์จึงนำโปรแกรมวิเคราะห์วิธีระเบียบวิธีไฟไนต์เอเลเมนต์ (finite element analysis; FEA) เพื่อทำการศึกษาถึงปรากฏการณ์ดังกล่าว

วิธีดำเนินการวิจัย: ผู้ประพันธ์ใช้โปรแกรม SOLIODWORKS 2018 ในการศึกษาภาวะเค้นและการกระจัดที่เกิดขึ้นทั้งในกระดูกทิเบียและข้อเข่าเทียมส่วนแผ่นทิเบีย (tibial tray) ชนิดโคบอลต์โครเมียมซึ่งมีความหนา 4 มิลลิเมตร และชนิดไทเทเนียมซึ่งมีความหนา 2 มิลลิเมตร ภายใต้แรงดันที่กระทำต่อแผ่นรองข้อเข่าเทียมซึ่งมีขนาด 1,000 และ 2,000 นิวตัน โดยวิเคราะห์การเปลี่ยนแปลงของความเค้นและการกระจัดของแผ่นรองข้อเข่าเทียมทั้งสองชนิด โดยวัดความเค้นที่ส่งผ่านไปยังกระดูกทิเบียที่ความลึก 1 และ 2 เซนติเมตร

ผลการวิจัย: ความเค้นที่เกิดขึ้นบริเวณศูนย์กลางเทียบกับบริเวณขอบ มีการเปลี่ยนรูปวัสดุของแผ่นรองมากในบริเวณศูนย์กลาง แต่อย่างไรก็ตาม แผ่นรองข้อเข่าเทียมชนิด 4 มิลลิเมตร แสดงถึงความมอดุลัสของสภาพยืดหยุ่น (Modulus of elasticity) มากกว่า และแผ่นรองข้อเข่าเทียมชนิด 2 มิลลิเมตร มีการเปลี่ยนรูปวัสดุมากกว่า โดยวัดความแตกต่างเปรียบเทียบระหว่างจุดศูนย์กลางและขอบของวัสดุมากกว่าอีกชนิดหนึ่งถึง 5 เท่า และความเค้นที่เกิดขึ้นผ่านแผ่นรองข้อเข่าเทียมชนิด 4 มิลลิเมตร ต่อกระดูกทิเบียที่บริเวณขอบเกิดขึ้นเพียงร้อยละ 18-22 เมื่อเทียบกับชนิด 2 มิลลิเมตร ซึ่งเกิดขึ้นถึงร้อยละ 54-66

สรุป: ข้อเข่าเทียมส่วนแผ่นทิเบียทั้งสองรูปแบบแสดงว่าให้เห็นว่ามีการลดลงของความเครียดในกระดูกของกระดูกที่บริเวณขอบของกระดูกทิเบีย อย่างไรก็ตามระดับความมอดุลัสของสภาพยืดหยุ่นและความหนาของแผ่นรองข้อเข่าเทียมชนิด 4 มิลลิเมตร แสดงให้เห็นภาวะส่งผลให้เกิดการลดลงของความเครียดในกระดูกที่มากกว่าอย่างมีนัยสำคัญ ซึ่งนำไปสู่การเกิดภาวะการณ์สลายและดูดซึมกลับของกระดูกทิเบียส่วนต้น

คำสำคัญ: การผ่าตัดเปลี่ยนข้อเข่าเทียม การลดลงของความเครียดในกระดูก ระเบียบวิธีไฟไนต์เอเลเมนต์ การเปลี่ยนแปลงรูปร่างของข้อเข่าเทียมส่วนแผ่นทิเบีย

วันที่รับบทความ 19 กันยายน 2565 **วันแก้ไขบทความ** 9 ธันวาคม 2565 **วันตอบรับบทความ** 13 ธันวาคม 2565

Introduction

The number of primary total knee arthroplasty (TKA) operations has increased markedly because of the aging society. Consequently, the number of revision TKA operations has also increased[1-2]. The major causes of TKA failure include infection, periprosthetic fractures, osteolysis, and aseptic loosening[1]. Despite advances in TKA technology, aseptic loosening and osteolysis are still major causes of revision[1,3]. In a retrospective study[4], medial proximal tibial bone loss after TKA was significantly higher in prostheses with thicker tibial baseplates compared with bone loss in prostheses with thinner tibial baseplates. These results correspond to recent reports[3,5-6] suggesting a high incidence of proximal tibial bone resorption for thick tibial baseplate designs. In our institute, we observed similar results with higher incidence of proximal tibial bone resorption in 4-mm-thick cobalt–chromium alloy (CoCr) tibial trays (figure 1) compared to 2-mm-thick titanium alloy (Ti) tibial trays. This may cause from differences between these two prostheses included the materials, designs, and tibial baseplate thicknesses. We hypothesized that these factors can change the pattern of deformation of the tibial tray under load, which may affect stress transfer to the supported proximal tibial bone.

Figure 1 Plain radiograph of 72-year-old female shown bone resorption of proximal Tibia after total knee replacement with 4-mm-thick cobalt–chromium alloy for 5 years

Finite element analysis (FEA) is used to study patterns of load distribution, contact surface, and microdeformation of knee prostheses[7-9]. To understand proximal tibial bone resorption better, we used FEA to analyze the stress and deformation of different tibial trays and stress on the supported proximal tibia beneath the trays under load.

Methods

This research was approved by the Ethics Committee of Navamindradhiraj University (COE 07/2020). We used a computer-generated finite element (FE) model with SOLIDWORKS 2018 software to analyze stress and deformation of the normal tibia and two tibial models: a 4-mm-thick CoCr tibial tray (4mm-tray) and 2-mm-thick Ti tray (2mm-tray). The three-dimensional drawings of both tibial tray models are shown in Figure 2 from SOLIDWORKS 2018 software. The materials used for the tibial trays were as follows: cobalt–chromium alloy (ASTM F75 CoCr alloy) in the CoCr tray and titanium alloy (Ti–6Al–4V alloy) in the Ti tibial tray. The properties of the CoCr alloy and Ti trays are shown in Table 1. The material properties of each part of the tibia and bone cement from the previous work of Enab et al.[10] were used in the model, and they are shown in Table 2.

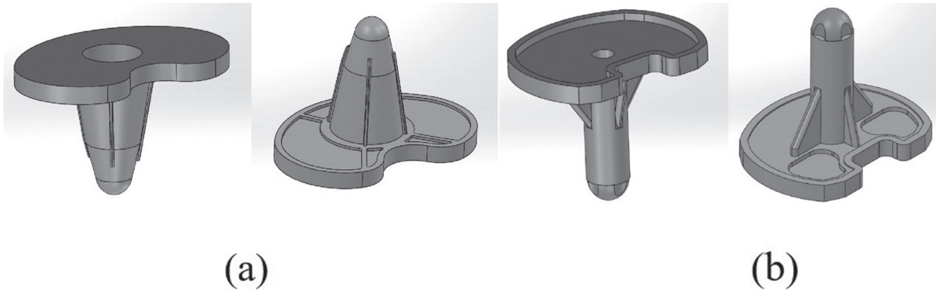


Figure 2 Three-dimension model of the tibial tray developed in FE simulation (a) Cobalt chromium alloy with 4 mm -thickness (b) Titanium based alloy (Ti-6Al-4V alloy) and Cobalt chromium alloy with 2 mm-thickness

Table 1 Material properties of alloy to be used in FE simulation

Properties	Material value		Units
	Ti-6A-4V alloy	ASTM F75 CoCr alloy	
Elastic Modulus	105	230	GPa
Poisson's Ratio	0.31	0.29	N/A
Mass Density	4430	8300	Kg/m^3
Tensile Strength	1050	480	MPa
Yield Strength	827	480	MPa
Thermal Expansion Coefficient	0.000009	12	/K
Thermal Conductivity	17	13	W/(m.K)

Table 2 Material properties represented in the FE model

Material	Modulus of elasticity (GPa)	Poisson's ratio
Cortical bone		
a. Top	14	0.3
b. Bottom	7	0.3
Cancellous bone		
a. Top	0.3	0.2
b. Top-middle	0.15	0.2
c. Middle	0.1	0.2
d. Bottom	0.05	0.2
PMMA cement	2.150	0.46

We matched the assigned tibial tray to the bone surface. A 2–3 mm layer of bone cement was created to simulate the cemented fixation of the tibial tray. The boundary conditions were set by clamping total support of the tibia tray and proximal tibial bone. Vertical loads of 1000 and 2000 N, which reflect the estimated normal and peak internal forces within the tibia during stance and the swing phase of subjects weighing 60 kg[11], were applied on the top surface of the tibial tray with equal force distribution on both sides of the plateau. The stress distribution of both tibia tray models was analyzed. Figure 3 showed examples of FE models under 1000N load. The stress on the supported proximal tibia at 1 and 2 cm beneath the surface was also analyzed for all models and compared to stress on the normal tibia and showed stress distribution on bone deep from surface 2 cm on FE models. We implemented a static analysis to calculate the stress distribution and deformation of the models. Central area deformation was determined at the closest part to the cone or stem in each design. Peripheral deformation is represented as a mean value of four measured points, as shown in Figure 4. Stress distribution on sagittal view of bone surface were shown in Figure 5.

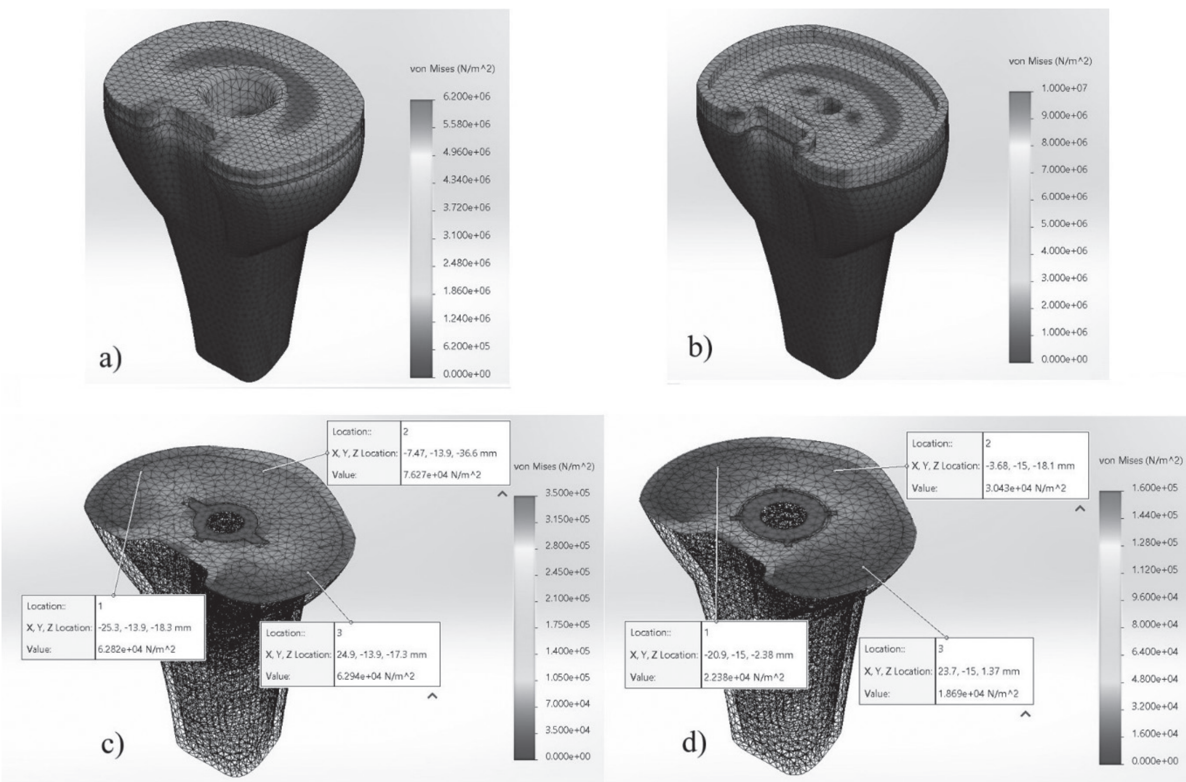


Figure 3 Stress distribution when force of 1000N applied to 4-mm CoCr based tibial trays (a) and 2-mm Ti based tibial tray (b) simulated using SOLIDWORKS. Stress distribution on bone deep from surface 2 cm of Peripheral area of 2-mm Ti based tibial tray (c) and 4-mm CoCr based tibial tray (d)

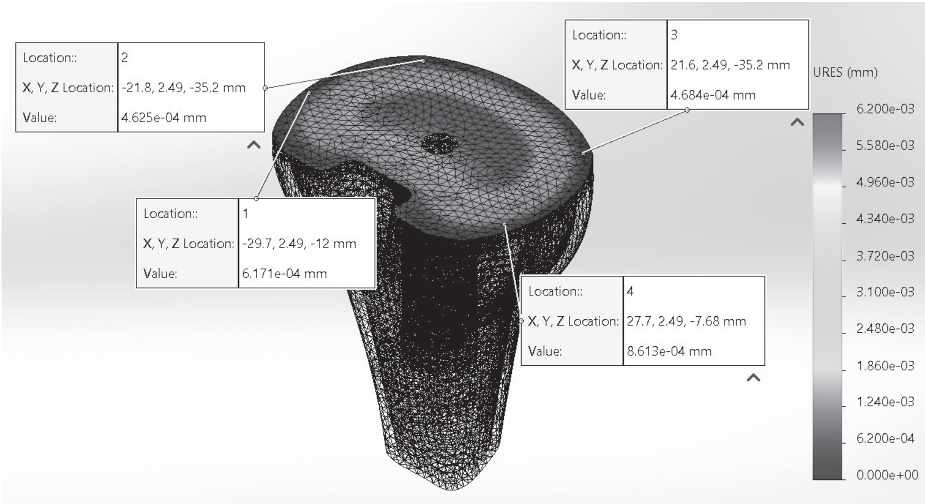
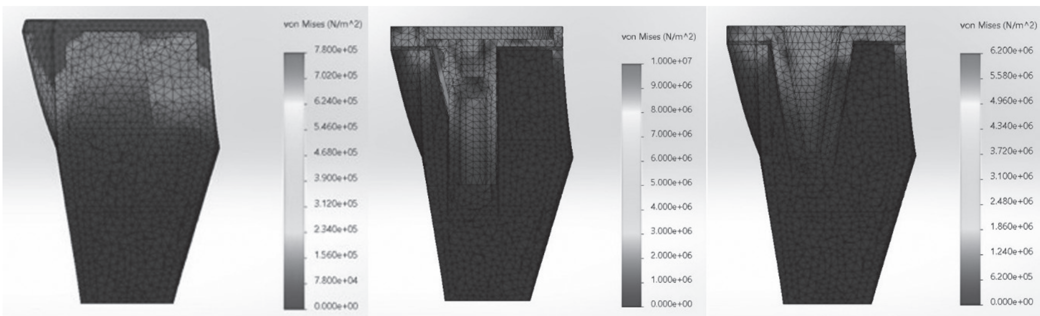


Figure 4 Displacement at 4 positions of Peripheral area of 2-mm Ti based tibial tray was measured after apply force of 1000 N



(a) (b) (c)

Figure 5 Sagittal view of stress distribution on normal bone (a), 2-mm Ti based tibial tray (b), and 4-mm CoCr based tibial tray (c)

Results

FE modeling revealed the same force patterns for the different tibial trays, with force concentrated more over the central region, as shown in Figure 2. When we increased force from 1000 to 2000 N, stress increased on the surface in both tibial tray models, as expected. Although the represented point may be slightly different between trays because of different designs, the pattern of stress concentration was the same, with more force concentrated over the central region for both tibial trays, as shown in Table 3.

Higher stress over the central region caused more deformation in the central region in both types of tibia trays, as shown in Figure 2. However, the degree of deformation was different. Under any load, the 4mm-tray showed less deformation compared to the 2mm-tray. Under a 1000 N load, the deformation difference between the central and peripheral regions for the 4-mm-tray was 13.05 microns and the difference for the 2mm-tray was 55.83 microns. The deformations of the trays under both loads are presented in Table 4.

Table 3 Stress on Ti Tibial tray and CoCr Tibial tray with force applied 1000 N and 2000 N (N/m^2)

	1000N		2000N	
	Peripheral	Central	Peripheral	Central
Ti Tibial tray 2 mm	2.78 MPa	10.10 MPa	5.50 MPa	20.19 MPa
CoCr Tibial tray 4 mm	2.33 MPa	6.22 MPa	4.71 MPa	12.11 MPa

Table 4 Detailed summary of deformation of tibia tray prosthesis under different load (Microns)

	Peripheral (micron)		Central (micron)		Peripheral-Central deformation (micron)	
	1,000 N	2,000 N	1,000 N	2,000 N	1,000 N	2,000 N
2 mm Ti tibial tray	6.02	13.36	61.85	123.90	55.83	110.54
4 mm CoCr tibial tray	6.275	14.08	19.33	38.50	13.055	24.42

Stress distribution underneath the tibial tray was also different between the two trays. At any load, the 4mm-tray had much less stress transfer to the peripheral part of the supported proximal tibia compared to the stress transfer for the 2mm-tray. Under a 1000 N load, stress on the peripheral region of the supported proximal tibia at 1 cm below the surface in a normal bone was 0.18 MPa. However, the stress values on the 4mm-tray and 2mm-tray were 0.04 and 0.12 MPa, respectively. If the load was increased to 2000 N, stress increased to 0.36 in a normal bone, 0.07 MPa in the 4mm-tray, and 0.23 MPa in the 2mm-tray. Load transfer to the supported proximal tibia was significantly reduced in both models but was more profound for the 4mm-tray. A detailed summary of stress underneath the tibial tray in all models is presented in Table 5.

Table 5 Detailed summary of stress underneath surface at different level with force of 1,000 N and 2,000 N (N/m^2)

	Normal Tibial bone (MPa)		2 mm Ti based tibial tray (MPa)		4 mm CoCr based tibial tray (MPa)	
	1,000 N	2,000 N	1000 N	2,000 N	1,000 N	2,000 N
Peripheral area at depth 1 cm	0.18	0.36	0.12	0.23	0.04	0.07
Peripheral area at depth 2 cm	0.11	0.21	0.06	0.13	0.02	0.05
Central area at depth 1 cm	0.13	0.27	0.05	0.1	0.03	0.06
Central area at depth 2 cm	0.06	0.12	0.09	0.18	0.03	0.06

Discussion

Loosening of the tibia prosthesis after TKA is one of the unsolved problems in knee arthroplasty. Many hypotheses explain early loosening, including inadequate fixation, poor cement technique, stress shielding, or implant malposition[6, 12]. Implant design, such as a cement pocket under the tibial tray and the roughness of the surface, are possible causes leading to the modification of the design. Despite improvements in the design, tibial loosening is still widely reported[6,12-13]. Proximal tibia resorption after TKA may aggravate this problem. A higher incidence of proximal tibial resorption in thicker tibial tray designs was reported, which aligned with our unreported data. When comparing the 4-mm-thick CoCr tray to the 2-mm-thick Ti tray, we found four main differences: 1) bearing mobility, which can be fixed or mobile, 2) tray design, 3) material, and 4) tray thickness. The 4-mm-thick CoCr trays used at our institution are mobile-bearing tibial trays, whereas the 2-mm-thick Ti trays are fixed-bearing tibial trays. The mobile-bearing knee has theoretical advantages, including less volume of wear particle creation from lower contact stress, which should lessen the wear particle-related problems. However, more recent reports showed that the amount and size of wear particles were similar in both mobile-bearing and fixed-bearing designs[14-15]. No clinical results suggest differently[3,6,13,16]. Therefore, we excluded the possibility of wear particle-related causes for proximal tibia bone loss or early loosening. Thus, these phenomena can be explained by the differences in the material, design, or thickness of tibial baseplates. Hence, the thickness of the tibial tray and a higher Young's modulus material may explain the differences, leading to the present study.

Our results showed the same pattern of stress and deformation for both tibial tray models. Stress was more concentrated around the central region of the trays, resulting in more deformation in the central region in both models. However, the degree of deformation was much lower in the 4mm-tray model. When we analyzed stress on the supported proximal tibia for the normal tibia and post-TKA models, we found that less stress was transferred to the supported proximal tibia in the post-TKA models, suggesting a stress shielding pattern. However, the stress transfer was much less in the 4mm-tray model, indicating a profound pattern of stress shielding. These results may be explained by the higher modulus material combined with the thicker construct in the 4mm-tray model. According to Wolff's law, stress shielding will result in some degree of bone loss, which may occur in the supported proximal tibia in our model.

In addition to the material and thickness, the design of the tibial tray may affect stress shielding. Two prosthesises built with almost equal thicknesses and the same material were shown to induce significantly different rates of proximal bone resorption[17]. The 2mm-tray in our study was designed with a reinforced peripheral ring, which theoretically improves the stiffness of the peripheral region compared to the 4mm-tray. However, the overall effects of the material and thickness overcome the effect of the design in our study.

There are several limitations to our study. Our study shows FEA results only; clinical application in a real situation depends on each surgeon. The load applied to the tibia in our study was in one direction. However, the normal knee is subjected to rotational force also, which was not applied in our FEA model. Further studies with more variables may show different results.

Conclusion

Because 4-mm-thick CoCr trays were made with a stronger material and thicker baseplate, the degree of deformation under load was less compared with the deformation in the 2-mm-thick Ti tibial tray. The FEA model also demonstrated that less stress was transferred to the supported proximal tibia underneath the tray, which may correlate with the resorption of bone in this region. Further studies with different tibial tray models or more complex load patterns can help us understand the complex nature of failure in knee arthroplasty.

Conflict of Interest

We declare no conflict of interest in this study.

Acknowledgement

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