Original article

Capability of auxetic femoral stems to reduce stress shielding after total hip arthroplasty

Bolun Liua, Huizhi Wangc, Min Zhanb, Junwei Lia, Ningze Zhanb, Yichao Luanb, Chaohua Fangb,c, Cheng-Kung Chenga,c,*

a Key Laboratory of Biomechanics and Mechanobiology, Ministry of Education, Beijing Advanced Innovation Center for Biomedical Engineering, School of Biological Science and Medical Engineering, Beihang University, Beijing, China
b Department of Joint Surgery, The 6th Hospital of Ningbo, No.1059 Zhongshan Road, Yinzhou District, Ningbo, 315000, Zhejiang, China
c School of Biomedical Engineering & Med-X Research Institute, Shanghai Jiao Tong University, Engineering Research Center of Digital Medicine, Ministry of Education, Shanghai, China

ARTICLE INFO

Keywords:
Auxetic structure
Femoral stem
Negative Poisson's ratio
Stress shielding
Total hip arthroplasty

ABSTRACT

Background: Stress shielding (SS) is considered the main mechanical cause of femoral stem loosening after total hip arthroplasty (THA). This study introduces an auxetic lattice femoral stem structure with negative Poisson's ratio that can expand laterally, with the intent of transferring more load to surrounding bone and thereby reducing SS. This study aims to evaluate how the geometry profile of different femoral stems with auxetic structures affects the level of SS. Different re-entrant angles for the auxetic unit cells were also evaluated.

Methods: This study assessed three commercial femoral stem designs (Mayo, CLS and Fitmore) and three re-entrant angles for the auxetic structures (60°, 70° and 80°). Nine auxetic femoral stems (three M-type, three C-type and three F-type) and three solid femoral stems (control group) were designed. All femoral stems were implanted into a finite element model of the human femur to compare levels of SS between the auxetic stems and their traditional solid counterparts.

Results: The results showed that incorporating an auxetic structure into the stem design caused less SS of the surrounding bone than the control models. The M-type stems had the lowest level of SS, followed by the C-type and F-type stems. A re-entrant angle of 70° for the M-type stem, 80° for the C-type stem and 60° for the F-type stem were the designs most capable of reducing SS.

Conclusions: This study found that femoral stems with an auxetic lattice structure caused less SS after THA than comparable solid femoral stems. A femoral stem based on the M-type geometry profile is recommended when designing auxetic femoral stems to minimize SS of surrounding bone.

The translational potential of this article: The novel solution provided in this study may serve to increase the survival rate of femoral stems by reducing SS after THA.

1. Introduction

Total hip arthroplasty (THA) is considered the gold standard for treating severe hip disease when more conservative approaches offer little relief to the patient. However, as younger patients undergo hip replacement, it is becoming increasingly important to design implants that can withstand an active lifestyle and minimize complications after long-term implantation. Aseptic loosening of the femoral stem is one of the most common causes of implant failure [1], with stress shielding (SS) being the main mechanical factor leading to aseptic loosening [2].

In the intact femur, load transfers vertically from the femoral head through the femoral neck to the femur shaft, resulting in a tensile force in the lateral proximal femur and compressive force in the medial proximal femur. Replacing a section of the femur with a hip implant disrupts the natural load transmission down the bone shaft and introduces a new interface between the implant and bone through which the load must be transmitted [2]. Femoral stems with higher stiffness bear greater loads than the surrounding bone, leading to a reduction in stress transferred to the proximal femur, which is termed SS. SS can cause periprosthetic bone resorption due to insufficient mechanical stimulation at the stem-bone interface.
fixation of the implant, often leading to aseptic loosening of the implant.

Previous studies introduced a wide variety of femoral stem designs and features aimed at reducing the level of SS from the implant. Among them, short stems have been shown to increase the magnitude of load transferred to the proximal femur [3], while the use of a conical profile can convert shear forces at the bone-implant interface into normal forces to increase the stress on the femur [4]. Alternatively, incorporating a collar on the proximal part of the stem allows for direct loading of the calcane of the femur which increases the axial stress on the bone. For the stem structure, using a slotted or grooved design or hollow and porous structures can reduce the local stiffness of the stem by reducing the material volume, which can increase the load transmitted to the bone [5]. Although such designs have been shown to reduce SS [4,6–10], several studies found that the stress on the femur after implantation is still less than on the intact femur, especially around the proximal lateral side near the greater trochanter [6,7,11].

An auxetic structure is one with a negative Poisson’s ratio, where the overall structure expands radially under axial tension (or retracts under compressive stress) on the tension side of the femoral stem and using the tensile structure inside the stem to expand the stem against the cavity walls [12]. Kolken et al. designed a femoral stem that combined both an auxetic structure and porous honeycomb structure. Kolken’s stem could produce compressive stress on both the medial and lateral sides of the femoral cavity to reduce the risk of failure at the fixation interface [13]. Ghavdelnia et al. reported that femoral stems with an auxetic structure have a more smooth and even stress distribution within the implant and produce less micromotion at the bone-implant interface [14]. Eldessouky designed a femoral stem with auxetic structures to promote bone ingrowth and reduce implant stiffness [15]. Although some previous studies developed femoral stems with auxetic structures, they did not investigate the ideal location for the structure. This study proposes placing the auxetic structure on the tension side of the femoral stem and using the tensile stress to expand the auxetic structure laterally to reduce SS. The geometric shape of the femoral stem, such as the stem length, fixation method and the stem’s offset, can also have a considerable impact on the mechanical transmission between the femoral stem and femur [3,6]. To the author’s knowledge, there is a lack of corresponding research on what geometric shape of the femoral stem with an auxetic structure best facilitates load transmission. In addition, the re-entrant angle has a prominent influence on the mechanical properties of the auxetic structure. The impact of different re-entrant angles on SS with different stem types was investigated in this study.

The aim of this study was to develop a finite element model of a femur and implant it with various femoral stems to determine how the shape of the stem and location of the auxetic structure affect SS of the surrounding bone. There are three hypotheses to this study. (1) Placing the auxetic structure on the tensile region of the femoral stem can reduce SS in comparison to its solid counterpart; (2) The Mayo femoral stem design (Zimmer Biomet, Indiana, USA), termed M-type in this study, causes less SS than other femoral stem designs; (3) A re-entrant angle of 60° is suitable to reduce SS compared with the other two angles. The results of this study may be used to improve the design of femoral stems with auxetic structures to reduce SS after THA.

2. Method

2.1. Development and validation of implanted femoral model

The 3D geometry of an intact femur was reconstructed from CT images of a healthy male (40 years, 80 kg) using Mimics 17 (Materialise N.V., Leuven, Belgium). Computer-aided design (CAD) models of three common commercial femoral stems with different design features were constructed in NX 12.0 (Siemens, Germany): Fitmore (Zimmer Biomet, Indiana, USA), CLS (Zimmer Biomet, Indiana, USA) and Mayo (Zimmer Biomet, Indiana, USA). The stem size chosen for each was determined from the size of the femoral model. These stems were chosen because of reports of SS from using these products [7,16,17]. The Mayo and Fitmore prostheses are classified as short stems which rely on a proximal press-fit to achieve stability [18]. The CLS prosthesis is a cementless stem of standard length that uses a press fit insertion at metaphyseal-diaphyseal junction to achieve stability. The loading axis of the CLS and the Fitmore stems coincides with the femoral axis, whereas the tip of the Mayo stem contacts the lateral cortex, thus increasing the stem’s offset (malalignment with the femoral axis). Each of the three stem models was virtually implanted into the femur using Mimics 17 under the guidance of an experienced orthopedist.

The femur and femoral stems were assumed to have isotropic, linear and elastic material properties [19]. All stems were modelled as a titanium alloy with an elastic modulus of 110 GPa and Poisson’s ratio of 0.3 [6]. The Poisson’s ratio of the whole femur was set as 0.3 [6] and the local Young’s modulus of the femoral bone was mapped in Mimics 17 using equations (1) and (2) [20]:

\[ \rho = -13.4 + 1017 \times \text{HU} \]  
\[ E = 8346 \times \rho^{1.5} \]

where \( \rho \) (g/cm³) is the density of the bone, HU is the Hounsfeld value from the CT images, and E (Pa) is the elastic modulus of the bone. A sliding surface-to-surface contact was applied between the stem and medullary cavity with a friction coefficient of 0.1 [21]. Using HyperMesh 12.0 (Altair Engineering, Tokyo, Japan), the femur was meshed with C3D4 elements (4-node tetrahedron linear solid elements) and the three dense stem models were meshed with C3D8 elements (8-node hexahedral linear solid elements). Mesh sensitivity analyses were conducted until there was a negligible change (relative error less than 5%) in the maximum von Mises stress. The final model had an element size of 2 mm for the femur and 1 mm for the femoral stems.

Bieger et al. investigated SS after in vitro implantation with Fitmore, CLS and Mayo femoral stems placed under a physiological load [7]. The boundary and loading conditions of the finite element model in this study were adopted from Bieger’s study, as follows. The anatomical axis of the femoral shaft (a line connecting the center of the marrow cavity and the center of the femoral condyle) was abducted 6° from the sagittal plane and had a flexion angle of 8° from the coronal plane. The distal end of the femur was fixed, and a static load of 1600 N (approximately 2 times body weight of an 80 kg man) was applied to the center of the femoral head (Fig. 1a). After loading, the average major strain (strain with the largest absolute value) was recorded at six pre-defined points and the values were compared with those reported by Bieger et al. to verify the accuracy of the FEA model.

2.2. Auxetic femoral stem design

Femoral stems with an auxetic structure were modelled from the profiles of the same commercial prostheses as the solid models, namely the Mayo, CLS and Fitmore femoral prostheses. The auxetic stems were then modelled with three different re-entrant angles for the unit cell, resulting in nine unique auxetic femoral stems (three M-type, three C-type and three F-type) and three solid femoral stems, which acted as a
2.2.1. Unit cell selection

The most common auxetic structures used in the design of hip replacements are re-entrant structures, rotating polygonal structures and chiral structures [22]. A re-entrant honeycomb is a typical re-entrant auxetic structure. Due to its simple configuration, a pattern can be easily expanded from a 2D to 3D profile and it is relatively easy to fabricate with additive manufacturing. Therefore, the re-entrant honeycomb structure was selected for this study. There are four primary design parameters for this auxetic unit cell, as shown in Fig. 1b: length of the vertical struts (H), length of the re-entrant struts (L), re-entrant angle (θ) and strut thickness (t). Considering that the fabrication accuracy of most metal additive manufacturing equipment is about 400 µm, the strut thickness (t) in this study was set as 0.4 mm [23]. Also, since the pore size of the unit cell needs to be within the range suitable for bone ingrowth (300 µm–800 µm) [24], H and L were set as 1.6 mm and 0.5 mm, respectively. The re-entrant angle (θ) is defined as the angle between the vertical strut and the inclined strut and changes in this angle have a greater influence on the mechanical properties than the other parameters (H, L, t) [25]. When modeling the 3D model of auxetic structure through NX 12.0 software, we found that if the re-entrant angle is less than 54°, the auxetic structure would not be able to expand. If the re-entrant angle is greater than or equal to 90°, the structure would no longer have a negative Poisson’s ratio. Therefore, the value should be within 54°–90°. As such, the femoral stems in this study were modeled with auxetic structures with re-entrant angles of 60°, 70° and 80°.

2.2.2. Location of the auxetic structures in the stem

Given that auxetic structures expand in response to a tensile load, the auxetic unit cells were only located on the tension side of the femoral stem. The neutral axis of the three solid femoral stems (Ms, Fs and Cs) was determined from the stress distribution on the finite element models simulated per section 2.1 of this manuscript. The auxetic structure would be placed in the region under tensile stress (gray region in Fig. 1d), with the region being further refined as follows.

(2) Since the femoral neck does not contact bone, no auxetic unit cells were placed around the neck of the stem, as shown by the shaded area 1 in Fig. 1e.

(3) The shoulder of the stem was designed as solid titanium to facilitate the fixation of auxetic structures. In addition, since the auxetic unit cells are cubic, any region intended to incorporate an auxetic structure should be regular without sharp changes in curvature to facilitate the arrangement and stacking of unit cells. Fig. 1d and e show a sudden change in the path of the neutral axis at the proximal end of the Ms stem, similar turning points also occur on the other two femoral stems. The region above a horizontal line passing through this turning point was modeled as solid titanium (shaded area 2 in Fig. 1e).

(4) The three femoral stems assessed in this study were designed to rely on metaphyseal or metaphyseal-diaphyseal load transmission for stability, and the distal end of the stems was not designed for implant fixation [26,27]. Therefore, no auxetic unit cells were placed around the distal end of the femoral stems. In addition to the proximal turning point identified previously, the distal end of the Ms stem also showed a sudden change in the path of the neutral axis, as shown in Fig. 1e. As such, the region below this turning point was modeled as solid titanium (shaded area 3 in Fig. 1e). As there were no evident sharp changes in the neutral axis around the distal end of the Cs and Fs stems, to unify the number of auxetic unit cells in the longitudinal direction, the distance between the upper and lower boundaries of the auxetic region on the Cs and Fs stems was set to be consistent with the Ms stem.

(5) Incorporating an auxetic structure on the tensile side of the femoral stem would reduce the stiffness of this region, causing the neutral axis to shift towards the medial side (Fig. 1f). Therefore, the medial boundary of the auxetic region was expanded to the medial side. We found that the new neutral axis reaches near the central line between the original neutral axis (black dotted line in Fig. 1e) and the medial curve of the stem (Fig. 1f). To facilitate the arrangement of unit cells and to maximize the area under tension, the region on the lateral side of this central line was defined as auxetic structures and the medial side was defined as solid titanium (shaded area 4 in Fig. 1f).

Fig. 1h shows the final design of the femoral stem including the auxetic lattice structure. The upper border of the auxetic region is designated by the green line, which was set to pass through the proximal turning point of the neutral axis. The blue line shows the medial border of the auxetic region, being a central line between the original neutral axis and the medial curve of the femoral stem. The lateral boundary is marked by the red line which follows the lateral curve of the femoral stem, and the yellow line defines the lower border and was set to pass through the distal turning point of the neutral axis. The method described above applies to all three stems.

Figure 1. (a) Loading and boundary conditions of finite element models, (b) design parameters of the auxetic unit cell, (c) illustration of femoral stem design features, (d) tensile region and neutral axis on the stem, (e) illustration of the region that need to be designed as solid titanium, (f) illustration of the tensile region after incorporating auxetic structure based on the previous step, (h) location of boundary lines between the auxetic and solid sections of the femoral stem. Original neutral axis (black dotted line); upper border (green line); lower border (yellow line); lateral boundary (red line); medial border (blue line). (For interpretation of the references to colour in this figure legend, the reader is referred to the Web version of this article.)
2.3. Biomechanical function of the femoral stems

Finite element models of the auxetic stem and femur system were developed with the auxetic unit cells being composed of 0.4 mm C3D8 elements. Other parameters, including the surgical implantation, mechanical properties of the implant and femur, contact conditions, and loading and boundary conditions were set up as described in section 2.1.

The average von Mises stress within each Gruen zone [28] on the femur was compared among the 12 femoral stems. The level of SS from each femoral stem was calculated using equation (3), with a positive SS value indicating greater SS.

\[
SS = \frac{(S_{int} - S_{stem})}{S_{int}} \times 100\%
\]  

(3)

\( S_{int} \) is the average von Mises stress on the intact femur, \( S_{stem} \) is the average von Mises stress on the implanted femur.

The difference in SS value between the auxetic stem and its solid counterpart showed how the auxetic structure influenced the level of SS.

As the flexibility of the stem body influences the load transferred from the femoral stem to the bone [5], finite element analysis was used to calculate the bending stiffness of the three solid stems and three auxetic femoral stems with a 70° re-entrant angle for the unit cells. Due to its porous nature, in theory, an auxetic stem should have a lower stiffness than its solid counterpart. The finite element models were constructed with the material properties, element type and element size being adopted from similar studies measuring the stiffness of femoral stems [29,30]. As shown in Fig. 2a, the stem was aligned at 10° in adduction and 9° in flexion, and the distal end of the femoral stem (80 mm below the center of the femoral head) was fixed. A uniform vertical displacement load of 1 mm was applied downward at the center of the femoral head.

3. Results

3.1. Validation of the finite element model

Fig. 3 shows the six measuring points on each femur, which were set according to the location of strain gauges in Bieger's study [7]. Gauge 1 and Gauge 4 were located at the level of lesser trochanter and below the greater trochanter, respectively. Gauge 2 and Gauge 5 were located medially and laterally at a distance 40 mm proximal to the tip of the stem. Gauge 3 and Gauge 6 were located medially and laterally 20 mm below the tip of the stem. The bone strain was measured after loading the implanted femoral stems, and the results were comparable to experimental measurements reported by Bieger et al. [7]. Specifically, tensile strains were observed on the lateral side of the femur and compressive strains on the medial side, while strain on the proximal femur at measuring points 1 and 4 was lower than at the other regions. The magnitude of strains in this study was also similar to those report by Bieger.

3.2. Biomechanical evaluation of auxetic femoral stems

Fig. 4 shows the von Mises stress distribution on the intact femur and on the femur after implanting different femoral stems. High stress was mainly observed around the medial zone of the femur after implantation. Fig. 5b–d quantitatively compares the average von Mises stress on the femur in each Gruen zone for the intact femur and implanted femurs. Figs. 4 and 5 show obvious SS at the lateral side (Gruen zone 1) and medial side (Gruen zone 7) of the proximal femur in all models implanted with a dense femoral stem. For the Mayo femoral stem, incorporating auxetic structures on the lateral side of the stem (M1, M2, M3) was found to increase the stress on the bone at all Gruen zones, and the stress values were higher than the threshold of bone resorption (2 MPa) [31]. Similarly, with the CLS stem, the stress on surrounding bone increased at all Gruen zones (except zone 3) when using an auxetic stem (C1, C2, C3) rather than a solid stem (Cs). Also, the stress values at all Gruen zones were higher than the threshold of bone resorption (2 MPa), except for C1 and C2 at Gruen zone 1. Compared with the solid Fitmore stem (Fs), the stress at all Gruen zones increased after implantation with auxetic stems (F1, F2, F3). However, the stress values at Gruen zone 1 with all three F-type stems were below the threshold of bone resorption.

3.2.1. Influence of stem geometry on SS

To evaluate the effect of stem geometry on SS, the difference in SS values between the auxetic stems and their solid counterparts was calculated (Table 2). In the lateral zones (zones 1, 2 and 3), the M-type stems (M1, M2 and M3) demonstrated the greatest reduction in SS in Gruen zone 1 (76%, 96% and 71% lower than Ms) and Gruen zone 3.

Figure 2. (a) Illustration of bending stiffness test setup, (b) load-displacement curve for auxetic stems and their solid counterparts.
The F-type stem showed the greatest reduction in Gruen zone 2 (52%, 62% and 63% lower than F$s$), while around the medial zones (zone 5, 6 and 7), the greatest reduction in SS was found at zone 6 (33%, 35% and 42% lower than F$s$) and zone 7 (39%, 40% and 42% lower than F$s$). The C2 and C3 stems showed the second-largest reduction in SS in zones 1 and 2, but overall, the C-type

Figure 3. Location of six measuring points on the medial and lateral sides of the femur after implanting (a) Mayo, (b) CLS and (c) Fitmore femoral stems. Histogram showing a comparison of strain measurements between this study and Bieger et al. [7] at the same measuring point.

Figure 4. Distribution of von Mises stress on the (a) intact femur, (b) femur after implanting a solid Mayo stem and M1, M2, M3, (c) femur after implanting a solid CLS stem and C1, C2, C3, and (d) femur after implanting a solid Fitmore stem and F1, F2, F3.
stems were the least effective at reducing SS in the other zones (except for zone 5). Of note, the SS values at zones 2 and 3 were found to be 5% and 2% higher than their solid counterpart.

3.2.2. Influence of re-entrant angle on SS
Table 3 shows the average von Mises stress on the femur implanted with different femoral stems and the ratio of these values as a percentage of the values recorded for the intact femur. For the M-type femoral stems, M2 demonstrated the greatest reduction in SS at all Gruen zones, especially at zones 1 and 2, with the load transmitted to these regions reaching 134% and 92% of the intact femur, respectively. Among the C-type femoral stems, C2 and C3 offered a similar reduction in SS, while the load transmitted to Gruen zone 1 by C1 was only 42% of the intact femur. For the F-type stems, F3 was best able to reduce SS of the bone. At zone 2, the von Mises stress recorded was 153% of the intact femur.

Table 3
| Implant | Zone 1 | Zone 2 | Zone 3 | Zone 5 | Zone 6 | Zone 7 |
|---------|--------|--------|--------|--------|--------|--------|
| M1      | –76    | –36    | –53    | –23    | –23    | –18    |
| C1      | –7     | 5      | 2      | –12    | –17    | –5     |
| F1      | –6     | –52    | –9     | –10    | –33    | –39    |
| M2      | –96    | –51    | –53    | –24    | –27    | –22    |
| C2      | –41    | 0      | 7      | –27    | –16    | –15    |
| F2      | –22    | –52    | –9     | –11    | –35    | –40    |
| M3      | –71    | –27    | –47    | –24    | –26    | –20    |
| C3      | –47    | –2     | 7      | –27    | –16    | –15    |
| F3      | –25    | –63    | –14    | –10    | –36    | –42    |

3.2.3. Stiffness of femoral stems
Fig. 2b shows the load-displacement curves for the auxetic stems and solid stems. The bending stiffness of the solid stems (Ms, Cs and Fs) was recorded as 4639 N/mm, 5667 N/mm and 7490 N/mm. For the auxetic stems, the bending stiffness of M1, C2 and F2 was calculated as 514 N/mm, 1224 N/mm and 1918 N/mm, which is 89%, 78% and 74% lower than their solid counterparts.

4. Discussion
This study assessed the ability of femoral stems designed with a partial auxetic structure to reduce SS in comparison to similar commercial femoral prostheses. The results showed that auxetic stems cause less SS than their solid counterparts. The auxetic femoral stems based on the Mayo stem (M-type) caused less SS, in general, than the stems modeled from the CLS (C-type) and Fitmore (F-type) prostheses. There is no regularity in the effect of the re-entrant angle of the auxetic unit cell on the SS for all prostheses.

The strain distribution and magnitude observed on the FEA models in this study were similar to the strains reported in an in-vitro study by Bieger et al [7]. However, individual differences such as the femoral stem size, medullary cavity shape, bone mass, anteversion angle, and neck-shaft angle of the femur did lead to some differences between the results.

The results showed that SS occurred around the proximal region of the femur (zone 1 and 7). This is in agreement with clinical data [26,32]. As expected, the bone surrounding the auxetic stems experienced greater stress than the femurs implanted with solid stems, indicating less SS. The increased stress around the lateral region of the femur may be attributed to the auxetic structure moving outward under tensile load. The increased stress on the medial side of the femur might be due to a
et al. [12] aimed to strengthen implant survivorship. An auxetic modular stem introduced by Alderson literature for different purposes, but with the overall aim of improving auxetic structure (Fig. 2b). Reduction in the stem's bending stiffness due to the presence of the cavity walls. Because more stress was transferred to the bone, this design producing compressive stress on both the medial and lateral sides of the proximal lateral femur. In Gruen zone 1, the M1, M2 and M3 stems were not considered [36]. Kolken et al. reported that the Poisson's ratio and Young's modulus of a re-entrant honeycomb structure. The greater the expansion (Δw) of the auxetic structure, the greater the corresponding stress transmitted to the bone. According to the definition of Young's modulus (eq. (4)) and Poisson's ratio (eq. (5)), it can be deduced from equation (6) that Δw may be calculated from a material's Poisson's ratio (v), Young's modulus (E), the force acting on it (F), and the width of the auxetic unit cell (w). Yang et al. reported that the Poisson's ratio and Young's modulus of a re-entrant honeycomb increase as the re-entrant angle increases [25,39]. Therefore, no correlation is anticipated between the re-entrant angle of the auxetic unit cell and resulting SS, which was confirmed by the results of this study (Table 3).

Table 3
Average von Mises stress on the femur implanted with different femoral stems, and the ratio of these values (%) with respect to the intact femur.

| Model | Average Stress (MPa) | % Intact | Gruen zone 1 | Average Stress (MPa) | % Intact | Gruen zone 2 | Average Stress (MPa) | % Intact | Gruen zone 3 | Average Stress (MPa) | % Intact | Gruen zone 4 | Average Stress (MPa) | % Intact | Gruen zone 5 | Average Stress (MPa) | % Intact | Gruen zone 6 | Average Stress (MPa) | % Intact | Gruen zone 7 | Average Stress (MPa) | % Intact |
|-------|----------------------|---------|--------------|----------------------|---------|--------------|----------------------|---------|--------------|----------------------|---------|--------------|----------------------|---------|--------------|----------------------|---------|--------------|----------------------|---------|--------------|----------------------|---------|
| Intact Femur | 2.53 | 100 | 5.61 | 100 | 8.49 | 100 | 14.13 | 100 | 10.15 | 100 | 9.91 | 100 |
| Ms | 0.96 | 38 | 2.33 | 42 | 8.95 | 105 | 11.82 | 84 | 7.01 | 69 | 5.27 | 53 |
| M1 | 2.89 | 114 | 4.35 | 77 | 13.46 | 159 | 15.06 | 107 | 9.30 | 92 | 7.10 | 72 |
| M2 | 3.40 | 134 | 5.19 | 92 | 13.49 | 159 | 15.22 | 108 | 9.77 | 96 | 7.41 | 75 |
| M3 | 2.76 | 109 | 3.87 | 69 | 12.94 | 152 | 15.25 | 108 | 9.68 | 95 | 7.24 | 73 |
| Cx | 0.87 | 35 | 5.96 | 106 | 3.32 | 39 | 5.06 | 36 | 6.05 | 60 | 8.96 | 90 |
| C1 | 1.05 | 42 | 5.68 | 101 | 3.18 | 37 | 6.70 | 47 | 7.77 | 77 | 9.43 | 95 |
| C2 | 1.92 | 76 | 5.98 | 107 | 2.69 | 32 | 8.82 | 62 | 7.71 | 76 | 10.49 | 106 |
| C3 | 2.08 | 82 | 6.06 | 108 | 2.74 | 32 | 8.85 | 63 | 7.70 | 76 | 10.44 | 105 |
| Fs | 0.49 | 19 | 5.06 | 90 | 9.31 | 110 | 12.43 | 88 | 7.89 | 78 | 4.75 | 48 |
| F1 | 1.09 | 43 | 8.00 | 143 | 10.05 | 118 | 13.82 | 98 | 11.29 | 111 | 8.57 | 87 |
| F2 | 1.05 | 41 | 7.97 | 142 | 10.07 | 119 | 13.93 | 99 | 11.41 | 112 | 8.70 | 88 |
| F3 | 1.13 | 45 | 8.57 | 153 | 10.48 | 123 | 13.87 | 98 | 11.54 | 114 | 8.92 | 90 |

Table 3 shows that a re-entrant angle of 70° for the M-type stem, 80° for the C-type and 60° for the F-type were the designs most capable of transferring load to the surrounding bone. Theoretically, the re-entrant angle of the unit cell may affect the mechanical properties of the re-entrant honeycomb structure. The greater the expansion (Δw) of the auxetic structure, the greater the corresponding stress transmitted to the bone. According to the definition of Young's modulus (eq. (4)) and Poisson's ratio (eq. (5)), it can be deduced from equation (6) that Δw may be calculated from a material's Poisson's ratio (v), Young's modulus (E), the force acting on it (F), and the width of the auxetic unit cell (w). Yang et al. reported that the Poisson's ratio and Young's modulus of a re-entrant honeycomb increase as the re-entrant angle increases [25,39]. Therefore, no correlation is anticipated between the re-entrant angle of the auxetic unit cell and resulting SS, which was confirmed by the results of this study (Table 3).

There are some limitations to this study. (1) The models did not include soft tissues such as muscles, ligaments and cartilage, or consider forces exerted on the limbs from surrounding soft tissues. (2) A static force was applied to the hip joint, which does not represent physiological loading during human gait. This study also did not simulate other common loading conditions, such as climbing stairs, running and squatting. (3) This study developed FEA models based on an adult male without considering the effect of medullary cavity morphology on the femoral stem's biomechanical function. Bone ingrowth into the surface of a femoral stem is helpful for improving long-term stability [40]. Although the interconnected pores of auxetic structures are theoretically suitable for bone ingrowth, this needs to be further validated through in vivo experiments.

 diversos en el esqueleto humano, y su estudio puede proporcionar una guía valiosa para el diseño de prótesis femorales. La reforma de la estructura auxética puede ser beneficiosa en la reducción de la rigidez del hueso y la transferencia más eficiente de la fuerza a través del hueso. Se ha comunicado que las prótesis femorales con estructuras auxéticas pueden reducir la fractura y la osteólisis, lo que podría mejorar aún más la supervivencia del implante.

La tabla mostrada en el texto muestra los esfuerzos promedio von Mises en diferentes zonas del fémur para diferentes modelos de prótesis femorales. Se observa que las prótesis auxéticas pueden reducir significativamente los esfuerzos comparados con los especímenes intactos. Sin embargo, se deben tener en cuenta las limitaciones mencionadas en el texto, como el modelo estático de carga y la ausencia de consideración de los tejidos blandos circundantes.

En conclusión, la incorporación de estructuras auxéticas en los prótesis femorales puede contribuir a mejorar la supervivencia del implante, pero se requieren más estudios experimentales en humanos para confirmar estos hallazgos.
5. Conclusion

This study introduced femoral stem designs incorporating auxetic lattice structures as a potential solution for reducing SS following THA. The results showed that the use of auxetic structures caused less SS of bone around the femoral stem in comparison to their solid (non-auxetic) counterparts. The auxetic femoral stems based on the Mayo stem (M-type) were generally better able to reduce SS than the GLS (C-type) and Fitmore (F-type) femoral stems. There was no predictable effect of the re-entrant angle on SS, with the most effective re-entrant angles being different for each stem type. The findings from this study may serve as a basis for designing auxetic femoral stems with better survivorship than traditional solid femoral stems due to their ability to transfer greater load to the surrounding bone.

Author contributions

BLL and CKC contributed in the conception and design of the study. BLL, JWL, NZZ contributed in acquisition and analysis of data. BLL contributed in original draft preparation. All authors contributed towards revising the manuscript. All authors have read and approved the final submitted manuscript.

Ethics Statement

Written informed consent was obtained from the individual(s) for the publication of any potentially identifiable images or data included in this article.

Declaration of competing interest

The authors declare that they have no known competing financial interests or personal relationships that could have appeared to influence the work reported in this paper.

Acknowledgments

This work was supported by the National Key Research and Development Program, China (Grant number: 2016YFC1101904) and Medical Health Science and Technology Project of Zhejiang Provincial Health Commission (Grant number: 2022RC248). We would like to thank Colin McClean for his assistance with editing this manuscript.

References

[1] Ulrich Silf D, Seyler TM, Bennett Derek, Delanois Ronald E, Saleh Khadel J, Thongtrakun Isada, et al. Total hip arthroplasties: what are the reasons for revision? Int Orthop 2008;32(5):597–604.
[2] Huiskes R, Weins J, Van RB. The relationship between stress shielding and bone resorption around total hip stems and the effects of flexible materials. Clin Orthop Relat Res 1992;274(274):124–34.
[3] Souza A, Van J, Dosins N, Neves P, Ramos J, Coelho R. Comparison of short-stem versus conventional stem for hip arthroplasty in patients younger than 60 years: 7–14 years follow-up. Eur J Orthop Surg Traumatol 2022;32(4):693–700.
[4] Maji PK, Roychowdhury A, Datta M. Minimizing stress shielding effect of femoral stem—a review. J Med Imag Health Inform 2013;3:171–8.
[5] Liu B, Wang H, Zhang N, Zhang M, Cheng C-K. Femoral stems with porous lattice structures as a potential solution for reducing SS following THA. J Mech Behav Biomed Mater 2018;77:520–50.
[6] Eldessouky I, El-Hofy H. Design and prototyping of a novel low stiffness cementless hip stem. Int J Biomed Eng Technol 2020;32:228–44.
[7] Martins LG, Garcia FL, Picado CH. Aective loosening rate of the Mayo femoral stem with medium-term follow up. J Arthroplasty 2014;29(11):2122–6.
[8] Nam D, Sali B, Barrass RL, Nuney RM. An analysis of proximal femur bone density in young, active patients undergoing total hip arthroplasty at one year postoperatively. Hip Int : the journal of clinical and experimental research on hip pathology and therapy 2019;29(1):51–7.
[9] Feyen H, Shimmin A. Is the length of the femoral component important in primary total hip replacement? The Bone & Joint Journal 2014;96-B:442–8.
[10] Wang S, Zhou L, Liu L, Shi H, Yao Y. On the design and properties of porous femoral stems with adjustable stiffness gradient. Med Eng Phys 2020;81:50–8.
[11] Sun C, Wang L, Kang J, Li J, Jin Z. Biomechanical optimization of elastic modulus distribution in porous femoral stem for artificial hip joints. J Biomech 2018;15(4):693–702.
[12] Matsuyma K, Ishidou Y, Guo YM, Kakoi H, Setoguchi T, Nagano S, et al. Finite element analysis of cementless femoral stems based on mid- and long-term radiological evaluation. BMC Musculoskelet Disord 2016;17(1):397.
[13] Ren X, Dai R, Tran P, Ngo TD, Xie YM. Auxetic metamaterials and structures: a review. Smart Mater Struct 2018;27:023001.
[14] Arabnejad S, Johnston RB, Pura JA, Singh B, Tanzer M, Pasi M, Fazli D. High-strength porous biomaterials for bone replacement: a strategy to assess the interplay between cell morphology, mechanical properties, bone ingrowth and manufacturing constraints. Acta Biomater 2016;30:345–56.
[15] Tarlochan F, Mebboob H, Mebboob A, Chang SH. Influence of functionally graded pores on bone ingrowth in cementless hip prosthesis: a finite element study using mechano-regulatory algorithm. Biomimetic Model Mechanobiol 2018;17(3):701–16.
[16] Yang L, Harryson O, West H, Cormier D. Mechanical properties of 3D-renter honeyscomb auxetic structures realized via additive manufacturing. Int J Solid Struct 2015;69:70:475–90.
[17] Brodt S, Matziolis G, Buckwitz B, Zippelius T, Strube P, Roth A. Auxetic metamaterials and structures: a review. J Med Imag Health Inform 2013;3:171–8.
[18] Pekpe W, Nador J, Everbeek J, Streib MR, Kinkel S, Gottharm T, et al. Primary stability of the Fitmore stem: biomechanical comparison. Int Orthop 2014;38(3):483–8.
[19] Gruen TA, McNeice GM, Amatuz HC. Modes of failure of cemented stem-type femoral components: a radiographic analysis of loosening. Clin Orthop Relat Res 1979;141:17–27.
[20] Jette B, Bralovski V, Dumas M, Simonoue C, Terrainu P. Femoral stem incorporating a diamond cubic lattice structure: design, manufacturing, and testing. J Mech Behav Biomed Mater 2018;77:58–72.
[21] Simonoue C, Terrainu P, Jette B, Dumas M, Bralovski V. Development of a porous metallic femoral stem: design, manufacturing, simulation and mechanical testing. Mater Des 2017;114:546–56.
[22] Frost HMA. Update of bone physiology and Wolff’s Law for clinicians. Angle Orthod 2003;74(1):1–5.
[23] Kutzner KP, Pfeil D, Kovacevic MP, Rehbn P, Mal S, Siebert W, et al. Radiographic alterations in short-stem total hip arthroplasty: a 2-year follow-up study of 216 cases. Hip Int : the Journal of clinical and experimental research on hip pathology and therapy 2016;26(2):278–83.
[24] Cansanan OD, Muradov PI, Simpson JB, Incavo SJ, Traumatol Cech 2021;88(1):50.
[25] Riviere C, Grappiolo G, Engh Jr CA, Vidalain JP, Chen AF, Boehler N, et al. Long-term alterations in short-stem total hip arthroplasty: a 2-year follow-up study of 216 cases. Hip Int : the Journal of clinical and experimental research on hip pathology and therapy 2016;26(2):278–83.
[26] Canham CD, Muradov PI, Simpson JB, Incavo SJ. Corrosion and adverse local tissue reaction after total hip arthroplasty with a modular titanium alloy femoral neck. Arthroplasty Today 2017;3(4):7–15.
[27] Cevad J, Ne A, Cabala J, Ne C, Grappiolo G, Engh Jr CA, Vidalain JP, Chen AF, Boehler N, et al. Long-term alterations in short-stem total hip arthroplasty: a 2-year follow-up study of 216 cases. Hip Int : the Journal of clinical and experimental research on hip pathology and therapy 2016;26(2):278–83.
[28] Cansanan OD, Muradov PI, Simpson JB, Incavo SJ. Corrosion and adverse local tissue reaction after total hip arthroplasty with a modular titanium alloy femoral neck. Arthroplasty Today 2017;3(4):7–15.
[29] Riviere C, Grappiolo G, Engh Jr CA, Vidalain JP, Chen AF, Boehler N, et al. Long-term alterations in short-stem total hip arthroplasty: a 2-year follow-up study of 216 cases. Hip Int : the Journal of clinical and experimental research on hip pathology and therapy 2016;26(2):278–83.
[37] Guo J, Tan J, Peng L, Song Q, Kong HR, Wang P, et al. Comparison of tri-lock bone preservation stem and the conventional standard Corail stem in primary total hip arthroplasty. Orthop Surg 2021;13(3):749-57.

[38] McLaughlin JR, Lee KR. Long-term results of uncemented total hip arthroplasty with the Taperloc femoral component in patients with Dorr type C proximal femoral morphology. The Bone & Joint Journal 2016;98-b(5):595-600.

[39] Yang L, Harrysson O, West H, Cormier D. Modeling of uniaxial compression in a 3D periodic re-entrant lattice structure. J Mater Sci 2013;25:1413-22.

[40] Fujibayashi S, Takemoto M, Sasaki K, Otsuki B, Nakamura T, Matsushita T, et al. Effect of pore size on bone ingrowth into porous titanium implants fabricated by additive manufacturing: an in vivo experiment. Mater Sci Eng C 2016;59:690-701.