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The sensitivity of cartilage contact pressures in the knee joint to the size and shape of an anatomically shaped meniscal implant


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Abstract

Since meniscal geometry affects the cartilage contact pressures, it is essential to carefully define the geometry of the synthetic meniscal implant that we developed. Recently, six independent modes of size- and shape-related geometry variation were identified through 3D statistical shape modeling (SSM) of the medial meniscus. However, this model did not provide information on the functional importance of these geometry characteristics. Therefore, in this study finite element simulations were performed to determine the influence of anatomically-based meniscal implant size and shape variations on knee cartilage contact pressures.

Finite element simulations of the knee joint were performed for a total medial meniscectomy, an allograft, the average implant geometry, six implant sizes and ten shape variations. The geometries of the allograft and all implant variations were based on the meniscus SSM. Cartilage contact pressures and implant tensile strains were evaluated in full extension under 1200 N of axial compression.

The average implant induced cartilage peak pressures intermediate between the allograft and meniscectomy and also reduced the cartilage area subjected to pressures >45 MPa compared to the meniscectomy. The smaller implant sizes resulted in lower cartilage peak pressures and compressive strains than the allograft, yet high implant tensile strains were observed. Shape modes 2, 3 and 6 affected the cartilage contact stresses but to a lesser extent than the size variations. Shape modes 4 and 5 did not result in changes of the cartilage stress levels.

The present study indicates that cartilage contact mechanics are more sensitive to implant size than to implant shape. Down-sizing the implant resulted in more favorable contact mechanics, but caused excessive material strains. Further evaluations are necessary to balance cartilage contact pressures and material strains to ensure cartilage protection and longevity of the implant.

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Introduction

The menisci are crucial for load transmission and stability of the knee joint (McDermott and Amis, 2006). Meniscal injuries occur frequently, resulting in approximately 650,000 meniscus-related surgeries annually performed in the USA (Abrams et al., 2013). A majority of these surgeries involves meniscectomy (e.g. partial or total removal of the meniscus). However, it is known that 50% of the meniscectomized patients still develop symptomatic osteoarthritis (Englund et al., 2003). These patients may be treated by meniscal allograft transplantation (i.e. replacement of the native tissue with a donor meniscus) to relieve pain and restore knee function (Rosso et al., 2014). However, problems related to the availability and sizing of allografts, the risk of disease transmission and post-implantation remodeling (Lee et al., 2012; Wada et al., 1998), have driven the search for an alternative treatment. A non-resorbable synthetic prosthesis could potentially overcome the shortcomings of meniscal allograft transplantation.

The native meniscus has a wedge-shaped geometry that assures optimal conformity between the femur and tibia. Meniscal geometry is known to influence the stresses and strains inside the meniscal tissue and the pressures acting on the articulating cartilage surfaces (Haut Donahue et al., 2004; Huang et al., 2002; Meakin et al., 2003). A mismatch of only ten percent between the native meniscus and a meniscal allograft may already disturb the articular cartilage contact pressures (Dienst et al., 2007).
Non-physiological elevation of the contact pressures is believed to lead to damage of the articular cartilage (Lanzer and Komenda, 1990; McDermott and Amis, 2006).

Recently, we have developed an anatomically shaped, polycarbonate urethane implant for total replacement of the medial meniscus. Given the importance of the geometry for native meniscus biomechanical functioning, careful consideration of the implant geometry is necessary to obtain optimal implant functioning. To address this problem, we have developed a statistical shape model (SSM) that describes the variations in 3D meniscal geometry (Vrancken et al., 2014). In this study, six independent modes were identified that together determine the variations in meniscal geometry; one size configuration (predominantly representing linear scaling) and five shape configurations. However, the SSM does not provide information on the functional importance of these geometry parameters. Therefore, the aim of this study was to computationally quantify the sensitivity of the tibiofemoral cartilage contact pressures to the previously identified meniscal implant size and shape variations, using 3D finite element simulations of the knee joint.

2. Methods

2.1. General approach

A 3D finite element model of the knee joint was developed based on the geometries of the knee joint tissues provided by the Open Knee project (Erdemir and Sibole, 2010). First, simulations were performed without the medial meniscus in place, representing the total meniscectomy case. Second, the simulations were repeated for a medial meniscal allograft. Last, the allograft was replaced by the implant and the simulations were repeated for 16 meniscal size and shape variations that were derived from a SSM of the medial meniscus (Vrancken et al., 2014). For each meniscal condition studied, the articular cartilage contact pressures and principal compressive strains were monitored. Additionally, for the allograft and implant cases, the internal meniscal tensile strains were evaluated.

2.2. Allograft and implant geometry

The mesh geometry of the medial meniscal allograft and all implant variations was derived from a SSM of the medial meniscus, which was based on 35 healthy male and female subjects (70 years old) (Erdemir and Sibole, 2010), a 3D finite element model of the knee joint was developed based on the MRI-extracted 3D geometry of the knee substructures from the Open Knee project (female, 70 years old) (Erdemir and Sibole, 2010), a 3D finite element model of the knee was developed in Abaqus v6.13 (Fig. 2). The model included the femur, tibia, anterior and posterior cruciate ligaments (the ACL and PCL, respectively), the medial and lateral collateral ligaments (the MCL and LCL, respectively), the articular cartilage and the medial and lateral menisci. General frictionless contact involving finite sliding was modeled between the femoral cartilage and the menisci, the menisci and the tibial cartilage, and the femoral and tibial cartilage surfaces of both the lateral and medial compartments. The ligaments, cartilage and native lateral meniscus were discretized into hexahedral elements with a full geometrically nonlinear formulation. Due to the geometrical complexity of the allograft and implants, quadratic tetrahedral elements were used for the allograft and all implants to avoid geometrical smoothing. Based on convergence studies, the characteristic element size of ligaments, cartilage surfaces and the lateral meniscus was chosen to be 1 mm and that of the medial meniscus allograft and the implants was chosen to be 0.7 mm. The convergence studies were conducted by doubling the mesh densities to verify that the observed model outcomes were independent of the chosen mesh densities.

The loading condition of interest here was that of short-term gait load of a single leg in full extension. The tibia was fixed and the femur was unconstrained in all translational and rotational degrees of freedom, except in flexion. An axial compressive femoral load of 1200 N was applied.

2.4. Material models

The native lateral meniscus, the medial meniscal allograft and the ligaments were modeled as transversely isotropic nearly-incompressible neo-Hookean materials (Erdemir and Sibole, 2010; Pena et al., 2006; Weiss et al., 1996), where the Cauchy stress was the summation of the stress in the bulk material, σbulk, and in the collagen fibers σfib:

\[
\sigma = \sigma_{\text{bulk}} + \sigma_{\text{fib}}
\]

The local fiber orientation at each voxel was estimated according to the local element geometry. In all ligaments, the fibers were in line with the principal geometrical axis of the ligament. In the meniscus, the fibers were aligned in circumferential direction, according to the well-established fiber orientation described in the literature (Fithian et al., 1990; McDevitt and Webber, 1990). The material parameters for the native lateral meniscus, the meniscal allograft and the ligaments are presented in Table 1. The meniscal horn attachments were modeled as linear springs (10 per horn) with a stiffness of 350 N/mm (Villegas et al., 2007).

The femur and tibia were modeled as rigid bodies given that bone stiffness is substantially higher than that of the included soft tissues. This assumption does not have a substantial effect on the contact variables (Donahue et al., 2002). The cartilage was assumed to behave as a linearly elastic material with an elastic modulus E=15 MPa and a Poisson ratio ν=0.475, based on selected experimental measurements under short loading times, where viscoelastic effects are negligible (Donzelli et al., 1999; Li et al., 2001; Pena et al., 2006). The meniscus implant material (polycarbonate urethane, Bionate® grade II BOA, DSM Biomedical, Geleen, Netherlands) was modeled as an isotropic neo-Hookean material with an initial Young’s modulus E=11 MPa and a Poisson ratio ν=0.49.

3. Results

The total medial meniscectomy resulted in a focal pattern of contact pressures centrally located on both the femoral and tibial cartilage. The meniscal allograft redistributed the pressures over a larger area, loading both the central and peripheral cartilage regions. The center of pressure on the femur was moved in posterior direction.
by the allograft. For the femoral cartilage, the average implant maintained the posterior shift of the center of pressure compared to the meniscectomy case; however, the pressures were higher than for the allograft (Fig. 3a). The cartilage area experiencing high pressures (45 MPa, which is the peak pressure observed for the native medial meniscus loading conditions comparable to those in our study (Allaire et al., 2008; Lee et al., 2006; Verma et al., 2008; Wang et al., 2014)) was considerably lower for the allograft and average implant, particularly for the femur (Fig. 4a and b). Although the allograft and average implant showed a favorable distribution of the pressures compared to the meniscectomy, only on the tibial side this was related to an increase of the contact area (Fig. 4e and f). The peak contact pressures for the average implant (6.0 MPa (femur) and 6.4 MPa (tibia)) were intermediate between those for the meniscectomy (7.1 MPa (femur) and 7.5 MPa (tibia)) and allograft conditions (5.2 MPa (femur) and 5.1 MPa (tibia)) (Table 2). The femoral cartilage experienced similar peak principal compressive strains for the

Table 1

<table>
<thead>
<tr>
<th>Tissue</th>
<th>$C_1$ (MPa)</th>
<th>$C_2$ (MPa)</th>
<th>$C_4$ (–)</th>
<th>$C_5$ (MPa)</th>
<th>$\kappa$ (MPa)</th>
<th>$\lambda^*$</th>
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<tbody>
<tr>
<td>Meniscus/allograft</td>
<td>4.6115</td>
<td>0.12</td>
<td>150</td>
<td>400</td>
<td>227.5</td>
<td>1.02</td>
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<tr>
<td>ACL</td>
<td>1.95</td>
<td>0.0139</td>
<td>116.22</td>
<td>535.039</td>
<td>73.2</td>
<td>1.046</td>
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<tr>
<td>PCL</td>
<td>3.25</td>
<td>0.1196</td>
<td>87.178</td>
<td>431.063</td>
<td>122</td>
<td>1.035</td>
</tr>
<tr>
<td>MCL</td>
<td>1.44</td>
<td>0.57</td>
<td>48</td>
<td>467.1</td>
<td>397</td>
<td>1.063</td>
</tr>
<tr>
<td>LCL</td>
<td>1.44</td>
<td>0.57</td>
<td>48</td>
<td>467.1</td>
<td>397</td>
<td>1.063</td>
</tr>
</tbody>
</table>

by the allograft. For the femoral cartilage, the average implant maintained the posterior shift of the center of pressure compared to the meniscectomy case; however, the pressures were higher than for the allograft (Fig. 3a). The cartilage area experiencing high pressures (> 5 MPa, which is the peak pressure observed for the native medial meniscus loading conditions comparable to those in our study (Allaire et al., 2008; Lee et al., 2006; Verma et al., 2008; Wang et al., 2014)) was considerably lower for the allograft and average implant, particularly for the femur (Fig. 4a and b). Although the allograft and average implant showed a favorable distribution of the pressures compared to the meniscectomy, only on the tibial side this was related to an increase of the contact area (Fig. 4e and f). The peak contact pressures for the average implant (6.0 MPa (femur) and 6.4 MPa (tibia)) were intermediate between those for the meniscectomy (7.1 MPa (femur) and 7.5 MPa (tibia)) and allograft conditions (5.2 MPa (femur) and 5.1 MPa (tibia)) (Table 2). The femoral cartilage experienced similar peak principal compressive strains for the
meniscectomy, allograft and average implant conditions, whereas for
the tibial cartilage, the average implant caused compressive strains
intermediate between the allograft and the meniscectomy conditions
(Table 2). The principal tensile strains generated in the
average implant showed patterns corresponding to those in the allograft;
strains were induced in the anterior horn and inner posterior region,
while the outer posterior region experienced negligible strains
(Fig. 3b). The maximum tensile strain was 23% for both the allograft
and average implant (Table 2).

The size-based variations in implant geometry clearly affected
the articular cartilage contact mechanics. While the larger implants
(mode 1–1SD, 2SD, 3SD) all induced concentrated areas of
high peak pressures on the central femoral and tibial cartilage, all
smaller implant variations (mode 1–1SD, 2SD, 3SD) distributed
the pressures over a larger area of the central and peripheral
cartilage surfaces (Figs. 4b, c and 5a). The pressure distribution
and the cartilage area experiencing peak pressures over 5 MPa for the
smaller implants were closer to that of the allograft than to that of
the average implant (Figs. 4b and 5a). The peak pressure observed
for the mode 1–1SD implant (4.6 MPa for both the tibia and femur)
was considerably lower than that of the allograft meniscus (5.2 MPa
(femur) and 5.1 MPa (tibia)), while the mode 1–3SD implant induced a peak pressure slightly lower than the allograft (5.1 MPa
(femur) and 5.0 MPa (tibia)). The peak pressures of all larger
implant variations resembled that of the meniscectomy condition
(7.1 MPa (femur) and 7.5 MPa (tibia)). Cartilage peak compressive
strains for the larger implants were comparable to those for the
average implant (19–21%), while the peak compressive strains for the
smaller implants were considerably smaller than for the
allograft condition (13–15% versus 18–19%, Table 2). On the other
hand, the smaller implants experienced substantially higher tensile
strains than the larger implants. All three smaller implant variations
showed tensile strains up to 40% in the anterior horn and along the
inner circumference of the posterior horn (Fig. 5b and Table 2).

The influence of implant shape on cartilage contact mechanics was
less pronounced than that of implant size. The pressure distribution
showed minimal changes for the shape variations over the ±35D
range of modes 2, 4 and 5. The implant shapes associated with modes
3 and 6 induced more distinct changes to the tibial and femoral
contact pressures over their ±35D ranges (Fig. 6). All implant shapes
reduced the area of the femoral cartilage experiencing contact
pressures over 5 MPa with respect to the meniscectomy condition,
while this effect was not observed for the tibial cartilage (Fig. 4b). The
peak pressures for modes 2+3SD, 3+3SD, 4+3SD and 5+3SD (6.0–
6.3 MPa) were close to those found for the average implant geometry.
Modes 3–3SD and 6–3SD induced peak pressures that were compar-
able to the meniscectomy condition (7.5 MPa). On the other hand,
mode 6–3SD resulted in peak contact pressures that were inter-
mediate between those found for the allograft and the average
implant (Fig. 6, Table 2). The cartilage peak compressive strains were
rather comparable for all implant shape variations (Table 2).

Implant strains were smaller for the shape variations that
resulted in higher contact pressures. However, the maximum
tensile strain did not exceed 23% for any of the implant shape
variations studied. The strain distribution was similar to that of the
average implant, with higher strains induced in the anterior,
central and inner posterior regions of the implants (Fig. 7, Table 2).

4. Discussion

The ability of a meniscal implant to improve cartilage contact
mechanics after total meniscectomy obviously depends on its geo-
metry. In this study, finite element simulations were employed to
evaluate the influence of anatomically-based geometry variations of a
poly carbonate urethane total medial meniscal implant on cartilage
contact pressures. The most important findings were: (1) cartilage
contact stresses and compressive strains were more sensitive to
variations in implant size than implant shape. (2) The optimally
size-matched, average implant reduced the cartilage contact stresses
to values intermediate between the meniscectomy and the meniscal
allograft cases, whereas the implant just under 10% smaller (mode
1–1SD) than the average implant was able to reduce the peak
contact pressures below the values found for the allograft. (3) The
down-sized implant geometries show maximal tensile strains double
that of the average implant.

Overall, the cartilage peak contact pressures and compressive
strains were lowest for the mode 1–3SD implant geometry. This
implant was able to distribute the pressures on the tibial and
femoral cartilage closely corresponding to that of the allograft and
reduced the peak contact pressures and compressive strains even
further than the allograft. These results are in agreement with experimental findings by Dienst et al. (2007) who showed that
lateral allografts more than 10.5% smaller than the native meniscus
could restore contact pressures close to native levels, whereas
optimally size–matched allografts showed higher contact pressures.
Due to ruptures in the smallest allograft group, they concluded that there is an increased risk of failure for allografts that are more than 10.5% smaller than the native meniscus (Dienst et al., 2007), which is also in concordance with our findings. In a 3D finite element model of the knee joint, Haut Donahue et al. studied the sensitivity of the cartilage contact pressures in response to variations of four meniscal dimensions (length, width, height and cross-sectional width) (Haut Donahue et al., 2004). Contrary to the experimental results by Dienst et al. (Dienst et al., 2007) and the findings from our simulations, Haut Donahue et al. did not report decreased contact pressures for down-sized meniscal geometries. However, rather than by a simultaneous change, they independently varied four geometrical parameters, which may explain these differences. Huang et al. concluded that the width of the meniscus body, within the anatomical range of variation, is an important determinant of the cartilage contact mechanics (Huang et al., 2002). In our simulations, the anatomical variation in meniscal body width is spread over the size mode and the five shape modes, with most of the variation occurring in the former. As the cartilage contact pressures were most sensitive to implant size variation, we do agree that meniscal body width, amongst others, has a considerable influence on the cartilage contact mechanics.

With the exception of modes 2, 3 and 6, the cartilage pressure distribution and the contact pressures were hardly affected by changing implant geometry over the ±3SD range of the SSM meniscal shape variations. However, the main changes in geometry over modes 3 (inverse length–width changes and posterior height changes) and 6 (overall meniscal height changes and a shift in anterior–posterior horn length) were not unique for these specific modes of meniscal geometry variation. Therefore it was not possible to identify meniscal shape parameters that advantageously or adversely affected the tibiofemoral contact mechanics. Most likely, the changes in contact stresses for the implant shape variations resulted from subtle differences in implant and cartilage congruency. Hence, these effects will be different in each knee joint.

In this study we found that although the smaller implants resulted in superior cartilage contact mechanics compared to the average implant, the tensile strains in the implant material increased up to 40% in these cases. This is quite a large strain to be accommodated by the implant material, which will be
subjected to long-term loading. Therefore, advancements to the implant design seem to be necessary in order to combine optimal cartilage contact mechanics with acceptable deformation of the implant material. Implant deformation could be limited by using a material with a stiffness exceeding that of the specific polycarbonate urethane under investigation in this study. Increasing the meniscus circumferential, radial and axial modulus to values that substantially exceed that of the native meniscus and the material under investigation in this study, demonstrated to decrease the cartilage contact pressures (Haut Donahue et al., 2003; Leatherman et al., 2014). Alternatively, reinforcing the bulk material of a disc-shaped meniscal replacement with highly stiff fibers has been shown to reduce the bulk strain and peak contact pressures as well (Elsner et al., 2010). Since any material property changes likely interact with the geometry-dependent response of the cartilage contact mechanics, it is necessary to repeat the simulations performed in this paper for any change to the implant material properties.

Table 2
(I) The cartilage peak contact pressures and (II) the cartilage peak principal compressive strains induced on the femoral and tibial cartilage for the meniscectomy (Mx) and allograft cases, the average implant and all implant size and shape variations. (III) The maximum tensile strain generated in the allograft and all implant variations.

<table>
<thead>
<tr>
<th>Size variations</th>
<th>Mx</th>
<th>Allograft</th>
<th>Average implant</th>
<th>Mode 1 (+3) SD</th>
<th>Mode 1 (+2) SD</th>
<th>Mode 1 (+1) SD</th>
<th>Mode 1 SD</th>
<th>Mode 1 (-1) SD</th>
<th>Mode 1 (-2) SD</th>
<th>Mode 1 (-3) SD</th>
</tr>
</thead>
<tbody>
<tr>
<td><strong>Cartilage peak contact pressure (MPa)</strong></td>
<td>Femur</td>
<td>7.1</td>
<td>5.2</td>
<td>6.0</td>
<td>5.1</td>
<td>5.0</td>
<td>4.6</td>
<td>7.4</td>
<td>7.4</td>
<td>7.3</td>
</tr>
<tr>
<td></td>
<td>Tibia</td>
<td>7.5</td>
<td>5.1</td>
<td>6.4</td>
<td>5.0</td>
<td>5.0</td>
<td>4.6</td>
<td>7.5</td>
<td>7.5</td>
<td>7.5</td>
</tr>
<tr>
<td><strong>Cartilage peak compressive strain (%)</strong></td>
<td>Femur</td>
<td>24</td>
<td>18</td>
<td>21</td>
<td>14</td>
<td>14</td>
<td>13</td>
<td>20</td>
<td>20</td>
<td>20</td>
</tr>
<tr>
<td></td>
<td>Tibia</td>
<td>20</td>
<td>19</td>
<td>19</td>
<td>15</td>
<td>14</td>
<td>13</td>
<td>20</td>
<td>20</td>
<td>20</td>
</tr>
<tr>
<td><strong>Max. tensile strain (%)</strong></td>
<td>Femur</td>
<td>–</td>
<td>23</td>
<td>21</td>
<td>40</td>
<td>40</td>
<td>40</td>
<td>10</td>
<td>10</td>
<td>10</td>
</tr>
<tr>
<td></td>
<td>Tibia</td>
<td>23</td>
<td>21</td>
<td>21</td>
<td>40</td>
<td>40</td>
<td>40</td>
<td>10</td>
<td>10</td>
<td>10</td>
</tr>
</tbody>
</table>

Shape variations

| **Cartilage peak contact pressure (MPa)** | Femur | 7.0 | 6.1 | 7.2 | 6.1 | 6.1 | 6.1 | 6.3 | 7.4 | 5.7 |
| | Tibia | 6.9 | 6.0 | 7.4 | 6.1 | 6.3 | 6.0 | 6.2 | 7.5 | 5.6 |
| **Cartilage peak compressive strain (%)** | Femur | 19 | 18 | 22 | 18 | 19 | 17 | 20 | 19 | 17 |
| | Tibia | 18 | 23 | 10 | 17 | 20 | 21 | 18 | 17 | 23 |
| **Max. tensile strain (%)** | 18 | 23 | 10 | 17 | 20 | 21 | 18 | 17 | 23 |

Fig. 5. (a) The medial femoral and tibial cartilage contact stresses following the variation of implant size and (b) the corresponding deformation and tensile strain in induced in the implants.
The sensitivity of knee cartilage contact stresses to meniscal geometry changes has been addressed using finite element modeling before (Haut Donahue et al., 2004; Meakin et al., 2003). However, this is the first study that uses 3D anatomically based meniscal geometry variations as an input. The SSM allows the definition of continuous 3D meniscal geometries in which all dimensions that are correlated change simultaneously, providing a more realistic representation of meniscal geometry variation than the subset of meniscal dimension variations studied by Haut Donahue et al. (Haut Donahue et al., 2004). As such it was possible to systematically evaluate a large set of geometry variations that may potentially affect functioning of our implant. On the other hand, the simulations presented here only took into account the effects of the separate modes of geometry variation, whereas combining modes may affect the contact mechanics in an alternative fashion. Performing a similar study experimentally would have been problematic as the implant production technique does not allow for easy changes to the implant geometry and degenerative changes to the cadaveric tissue would interfere with testing all geometries under equal conditions.

In order to extrapolate the results of any numerical simulation to the in vivo situation, verification and validation of the model outcomes against experimental data is necessary. In terms of direct validation, we were hampered by the fact that we used the Open Knee model. Hence, the physical knee was not available for validation purposes. We therefore compared our results with those reported in the literature, to assess whether our model predicts appropriate outcomes. The literature provides several studies that experimentally measured the tibial cartilage contact pressures for a total medial meniscectomy and after medial allograft transplantation, which can provide a reference for the cases that we studied (Table 3). The experimentally recorded contact pressures were highly variable, which could not solely be attributed to the differences in loading conditions. Under 1000 N of compression peak contact pressures between 6 and 12.3 MPa were reported after a total medial meniscectomy (Allaire et al., 2008; Fukubayashi and Kurosawa, 1980; Verma et al., 2008), while 1800 N axial compression resulted in peak pressures between 6.0 and 9.2 MPa (Lee et al., 2006; Paletta et al., 1997; Wang et al., 2014). The peak pressure predicted for the meniscectomy condition in the present study (7.5 MPa) is in acceptable agreement with these studies, considering the intermediate loading of 1200 N. Subsequent implantation of an allograft meniscus generally resulted in a decrease of the peak contact pressures. However, Wang et al. did report on a slight increase of the peak pressures for the fully extended knee (Wang et al., 2014). The 31% reduction of the peak contact pressures induced by the allograft in the present study is in good agreement with the observations by Verma et al. (Paletta et al., 1997; Verma et al., 2008). By including both a meniscectomy and an allograft as control conditions for comparison with the implant, we are confident that our simulations

![Fig. 6. The medial femoral and tibial cartilage contact stresses following the variation of implant shape.](image-url)
provide the information necessary to compare implant functioning to the cases that are relevant in the current clinical practice.

Several limitations should be considered when interpreting the results of our study. As every person displays a unique native meniscus and knee joint geometry, each knee may potentially respond different to the implant geometry variations studied here. Repeating the simulations for a larger set of knee models would provide insight whether our results can be extrapolated to a general population or whether patient-specific models are necessary to predict the effect of a meniscal implant on cartilage stresses. In the present study, static simulations were performed under rather limited loading conditions. It is well established that cartilage and menisci are biphasic tissues (Mow et al., 2005). However, when subjected to a step load, the instantaneous response of a biphasic material is equal to that of an elastic material as the fluid flow (and thus the biphasic effect) is not substantial at this time scale (Ateshian et al., 2007). As this study focused on the initial response of the cartilage and meniscus to a compressive load experienced during gait, these tissues can thus be modeled as elastic materials. In addition, the tissue-level complexities (e.g., anisotropy, the swelling properties and the inhomogeneous distribution of the biochemical constituents) in the material descriptions of the cartilage were not accounted for (Wilson et al., 2006). Future numerical studies may address these limitations to more accurately predict the mechanical response of the articular cartilage, preferably using dynamic simulations with loading conditions derived directly from gait data.

5. Conclusion

The numerical simulations conducted in the present study demonstrate that cartilage contact pressures and compressive strains are influenced by the geometry of a soft grade polycarbonate urethane implant for total medial meniscal replacement. Cartilage mechanics were shown to be more sensitive to implant size than to implant shape. Down-sizing the implant resulted in an improvement of the mechanical output parameters compared to the best fitting, average implant. The implant that was approximately 10% smaller than the average was able to reduce the cartilage peak contact pressures and compressive strains below the values found for the allograft. However, the strains in the down-sized implants locally increased beyond the elastic range of the intended implant material. These key findings will be employed to tune the geometry and material properties of our novel meniscal implant. During this process, care should be taken to balance a maximal reduction of the peak contact pressures with material deformations that remain within the elastic range of the implant material.

Table 3

<table>
<thead>
<tr>
<th>Study</th>
<th>Flexion angle</th>
<th>Compressive load (N)</th>
<th>Meniscectomy peak pressure (MPa)</th>
<th>Allograft peak pressure (MPa)</th>
<th>Meniscectomy–Allograft (%)</th>
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</thead>
<tbody>
<tr>
<td>Fukuhayashi and Kurosawa (1980)</td>
<td>0</td>
<td>1000</td>
<td>6.0</td>
<td>–</td>
<td>–</td>
</tr>
<tr>
<td>Allaire et al. (2008)</td>
<td>0</td>
<td>1000</td>
<td>6.4</td>
<td>–</td>
<td>–</td>
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<tr>
<td>Lee et al. (2006)</td>
<td>0</td>
<td>1800</td>
<td>9.2</td>
<td>–</td>
<td>–</td>
</tr>
<tr>
<td>Paletta et al. (1997)</td>
<td>0</td>
<td>1800</td>
<td>8.3</td>
<td>3.6</td>
<td>– 57%</td>
</tr>
<tr>
<td>Verma et al. (2008)</td>
<td>0</td>
<td>1800</td>
<td>12.3</td>
<td>7.8</td>
<td>– 36%</td>
</tr>
<tr>
<td>Wang et al. (2014)</td>
<td>15</td>
<td>2400</td>
<td>6.1</td>
<td>5.0</td>
<td>– 22%</td>
</tr>
<tr>
<td>Present study</td>
<td>0</td>
<td>1200</td>
<td>7.5</td>
<td>5.1</td>
<td>– 31%</td>
</tr>
</tbody>
</table>

Fig. 7. The deformation and tensile strain induced in the implants after variation of the implant shape.
Conflict of interest statement

T.G. van Tienen and P. Buma have published (1) and issued (2) patents related to the implant evaluated in this study.

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References


