Function of the medial meniscus in force transmission and stability

Peter S. Walker, Sally Arno, Christopher Bell, Gaia Salvadore, Ilya Borukhov, Cheongeun Oh

Abstract

We studied the combined role of the medial meniscus in distributing load and providing stability. Ten normal knees were loaded in combinations of compressive and shear loading as the knee was flexed over a full range. A digital camera tracked the motion, from which femoral tibial contacts were determined by computer modelling. Load transmission was determined from the Tekscan for the anterior horn, central body, posterior horn, and the uncovered cartilage in the centre of the meniscus. For the three types of loading; compression only, compression and anterior shear, compression and posterior shear; between 40% and 80% of the total load was transmitted through the meniscus. The overall average was 58%, the remaining 42% being transmitted through the uncovered cartilage. The anterior horn was loaded only up to 30 degrees flexion, but played a role in controlling anterior femoral displacement. The central body was loaded 10–20% which would provide some restraint to medial femoral subluxation. Overall the posterior horn carried the highest percentage of the shear load, especially after 30 degrees flexion when a posterior shear force was applied, where the meniscus was estimated to carry 50% of the shear force. This study added new insights into meniscal function during weight bearing conditions, particularly its role in early flexion, and in transmitting shear forces.

1. Introduction

Various biomechanical functions have been ascribed to the menisci including distribution of the force across the knee, protection of the underlying cartilage, lubrication of the cartilage surfaces, and stability (Fox et al., 2014; Abraham and Donahue, 2013). On the lateral side, force distribution is necessary due to the low conformity between the cartilage bearing surfaces, while medially, some anterior–posterior stability is likely due to the more conforming geometry and the reduced meniscal mobility. The role of the medial meniscus is of particular importance due to the frequency of injuries, the higher forces on the medial side (Halder et al., 2012), and the incidence of medial osteoarthritis (Gale et al., 1999; Badlani et al., 2013).

The contact areas on the meniscal surfaces have been determined in several studies using sectioning or casting techniques (Walker and Erkman, 1975; Takel, 1979; Fukubayashi and Kurosawa, 1980; Kurosawa et al., 1980). Other studies used pressure sensitive films to quantify the pressure distribution and the forces acting (Fukubayashi and Kurosawa, 1980; Ahmed and Burke, 1983; Seitz et al., 2012). In almost all of the studies, the loading times were prolonged and thus the cartilage and menisci were likely to have deformed more than for the short-term loading in functions such as walking (Hosseini et al., 2010; Liu et al., 2010). Also, the applied forces were along the tibial axis but without superimposed shear or torque. Nevertheless, the results have been consistent in showing a substantial area of contact on the meniscus and on the uncovered cartilage, progressively more posterior contact with flexion, and up to 60% of the force being transmitted through the meniscus. More recent studies have used pressure sensitive film in knee specimens in a knee simulating machine in walking or stairclimbing (Bedi et al., 2010; Gilbert et al., 2014; Wang et al., 2014) More variable pressure patterns were measured than for single loading conditions. A summary of these studies is given in Table 1. Regarding the stabilizing role of the meniscus, the anterior–posterior stability provided by the medial meniscus has also been determined, although as a secondary stabiliser compared with the role of the cruciate ligaments (Shoemaker and Markolf, 1986). Musahl et al. (2010) showed the importance of the medial meniscus when the tibia was loaded anteriorly. The stability data by Arno et al. (2012) after partial meniscectomy experiments is consistent with these studies.

Nevertheless, in medial osteoarthritis, there is strong evidence that cartilage degeneration and loss most often occurs in the central region of the tibial plateau uncovered by the meniscus, and on the centre of the distal femur (Arno et al., 2012). This might not be expected if the medial meniscus was indeed carrying over half
of the force. There are however other biomechanical factors which can play a role in causing degeneration including higher medial forces due to varus alignment (Halder et al., 2012), increased can play a role in causing degeneration including higher medial
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The purpose of this study was to investigate the load-bearing and stabilizing role of the medial meniscus under test conditions which represented a range of functions. This would include both compressive and shear forces over different flexion ranges, using input data from instrumented total knees (Heinlein et al., 2009; D’Lima et al., 2011). Motion should also be continuous over the flexion range rather than the tests being conducted under static conditions. The focus of our study was to determine the actual forces transmitted through three meniscal regions; anterior, central and posterior over a full range of flexion and shear as well as compressive forces applied. Our first hypothesis was that the highest loading on the medial meniscus would be at the extremes of motion, rather than in the mid-range. Our second hypothesis was that the medial meniscus would provide the majority of the restraint to anterior–posterior femoral displacement throughout flexion when compressive loads were acting.

2. Methods and materials

Ten fresh knee specimens, 7 male 3 females 55–91 years were used. MRI scans were taken to determine that there was no abnormal pathology of the cartilage surfaces, menisci, or anterior cruciate, which was verified visually after the experiments. The tibia was mounted vertically to the base of the Desktop Knee Machine (Arno et al., in press; Walker et al., 2015) by cementing to an intramedullary rod (Fig. 1). A counterbalanced frame was pinned to the shaft of the femur. Two side axles projected outwards in line with the circular axis. Horizontal and vertical cables were attached to the axles to apply combinations of vertical compressive force, and anterior–posterior (AP) shear force. A cable attached to the proximal femur applied flexion from an extended position in 10–15°. A Tekscan sensor (500 N sensor, 4011 N, Tekscan Inc, Boston MA) was inserted between the medial meniscus and the cartilage surface through an anterior incision, and sutured posteriorly (Fig. 2). The sensor covered all of the cartilage under the meniscus and the central uncovered cartilage, but did not include the slope of the tibial spine.

A sequence of loads was then applied, each for a full range of flexion from 5 degrees hyperextension to 135 degrees flexion, or as close to that as the knee would allow. Tekscan data was obtained for the following sequences of loading: 500 N compressive vertical load, 500 N + 100 N anterior shear (an anterior force on the femur applied simultaneously to the compressive load), 500 N + 100 N posterior shear. The values were scaled down from values determined from instrumented total knees (Heinlein et al., 2009; D’Lima et al., 2011) for durability and strength considerations. The vertical load was applied to the side axles of the femoral frame such that the
lateral and medial forces would be nominally equal. During flexion, a digital camera was used to track targets on the femur and tibia, to enable femoral–tibial contacts to be determined, details of the methodology being described previously (Walker et al., 2015).

After the testing, the capsule was excised and the inner periphery of the meniscus was traced on the Tekscan with the knee at 30 degree flexion. The tracing was recorded by the Tekscan and was used for identification of the 3 regions of the anterior horn, central body, and posterior horn (Fig. 2). These regions would remain relatively unchanged during the flexion–extension ranges based on the analysis of the contact patches of the medial femoral condyle on the tibia from a related study under the same loading conditions (Walker et al., 2015).

The Tekscan software was then used to determine the forces on the 3 regions of the meniscus as well as on the central area of the tibial plateau not covered by the meniscus (uncovered cartilage). The forces were calculated as a percentage of the total force on the medial side taken to be the sum of the 3 meniscal regions and the uncovered cartilage.

An estimate was made of the shear forces between the upper meniscal surfaces on the anterior and posterior horns, and the femoral condyles. This was used as a measure of AP stability. The estimate of shear force was obtained by using a mid-medial sagittal MRI section taken of the intact knees before testing, and the Tekscan data. A tangent was drawn at the midpoint of the interface between the meniscus and the femoral condyle (Fig. 3). It was assumed that the resultant reaction force occurred at this point, and also that the force on the upper surface was the same as on the lower surface, the latter being measured by the Tekscan. The angle of the tangent line to the horizontal was measured. For the posterior horn, the horizontal shear component SP of the reaction force MP is: SP = MP sin P. The net shear force in the anterior direction on the femur is (SP – SA).

The net shear forces were calculated for all 10 knees for all flexion angles. The MRI scan was taken at a close to extension and it was assumed that the angles A and P would remain constant throughout flexion. This assumption was reasonable posteriorly where the sagittal femoral radius is almost constant. Anteriorly, the angle would be reasonably constant up 30 degrees flexion, after which the forces on the anterior horn were small. The resultant shear forces, in Newtons, were averaged for the 10 knees.

3. Statistical analysis

In order to evaluate differences of percentages of load in repeated measurements between meniscal regions, different flexion angles, and different force types, a multi-variate linear mixed-effect model was used. This was an extension of a linear regression model, which included additional random effect terms to incorporate dependence of the repeated or related subjects, that is the knees in our study. In our study we used R for fitting mixed models to data, namely, the nlme (an acronym for non-linear mixed effects), described in detail by Pinheiro and Bates (2000). The statistical model was designed to evaluate the percentage of load as a function of these three variables within specimens (regimens, flexion angles, force types) and their interactions. A random statement with study knees (subjects) as the random effect was included to account for the correlation within knees because of the repeat nature of the data. The mixed model was generated using the nlme package. Statistical significance of \( p < 0.05 \) was considered to be relevant.

4. Results

The average percentages of the total force acting on the 3 meniscal regions are shown in Fig. 4. For vertical compression (Fig. 4A), the meniscus carried between 44% and 78% of the load. Overall, the posterior horn had the highest contribution, even at low flexion angles, but with a peak at 135 degrees flexion, although the latter was for only 6 of the 10 specimens. The central body contributed at a moderate level throughout flexion. The anterior horn however carried force primarily towards extension, with small values after 30 degrees flexion. When an anterior shear force was superimposed (Fig. 4B), the total percentage was little changed from compression only. In early flexion, the anterior horn was dominant, with small participation from the posterior horn. However, again the force on the anterior horn was small after 30 degrees flexion. After this, the force on the posterior horn was similar to that for compression only. The central body loading was similar to that for compression throughout flexion. For a posterior shear force, (Fig. 4C), the total percentage carried by the meniscus increased from 61% to 82%. The anterior horn still carried force up to 15 degrees flexion, but the force on the posterior horn increased substantially throughout flexion. The central body loading was again similar to compression only. The statistical analysis showed
that there was significance at $< 0.05$ for the meniscus region and for the type of loading. However there was no significance overall for the angle of flexion (Table 2).

The overall average for meniscal load-bearing for the 3 loading conditions was 58%. However, there was considerable variation in the pressure patterns and forces between knees. Examples of 3 Tekscan patterns for specific cases are shown in Fig. 5. A typical average case is shown; a case with a pressure bias towards the posterior horn; and a single case with minimal pressure on the central body. In all cases, there was pressure on the tibial surface not covered by the meniscus.

Data of the anterior–posterior stability analysis is shown in Fig. 6. The measure of stability was the net anterior or posterior shear force carried by the meniscus, in relation to the applied shear force.

In early flexion, due to loading of the anterior horn, the meniscus carried 35% the applied shear force. After early flexion, the anterior horn had a minimal role in stability. In contrast, the posterior horn played an increasing stabilizing role when posterior shear was applied. After 30 degrees flexion it carried 50% of the applied shear force.

### Table 2

Statistical analysis of the data in Fig. 4. Mixed linear model on percentage of load as outcome, against the meniscal regions, the angles of flexion, and the force type. 1 is indicated with reference to the anterior horn; 2 is indicated with reference to the anterior shear loading.

<table>
<thead>
<tr>
<th>Variable</th>
<th>Coefficient (SD)</th>
<th>p-value</th>
</tr>
</thead>
<tbody>
<tr>
<td>Meniscus region¹</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Central body</td>
<td>8.2795 (1.42)</td>
<td>&lt; 0.0001</td>
</tr>
<tr>
<td>Posterior horn</td>
<td>32.2016 (1.42)</td>
<td>&lt; 0.0001</td>
</tr>
<tr>
<td>Angles of flexion</td>
<td>−0.0135 (0.0135)</td>
<td>0.2507</td>
</tr>
<tr>
<td>Force type²</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Compression</td>
<td>3.1239 (1.4284)</td>
<td>0.0287</td>
</tr>
<tr>
<td>Posterior shear</td>
<td>9.5680 (1.4123)</td>
<td>&lt; 0.0001</td>
</tr>
</tbody>
</table>

### Fig. 4

The average percentages of the forces on the 3 meniscal zones at the different flexion angles for the 10 knees tested. (A) Vertical compression; (B), compression and anterior shear, and (C), compression and posterior shear. The remaining force (up to 100%) was carried on the uncovered cartilage on the centre of the tibial plateau. Unless otherwise noted, the sample number ($n$) for the data shown was $n = 10$. For clarity, the standard error (SEM) is shown rather than standard deviation (SD). Note: SD = SEM$\sqrt{n}$ i.e. for $n = 10$, SD = 3.2 SEM.

### Fig. 5

Examples of Tekscan pressure patterns at different flexion angles. Top, a typical average distribution; centre, the loads biased posteriorly; bottom, no loading on the central body.
highest percentage of load overall, especially at higher shear force was applied to the femur. The posterior horn carried the horn experienced its highest loads, particularly when an anterior meniscus remained relatively constant throughout force at low for compressive load in high for flexion angle and loading conditions. In early flexion, the anterior horn experienced its highest loads, particularly when an anterior shear force was applied to the femur. The posterior horn carried the highest percentage of load overall, especially at higher flexion angles and when there was a posterior shear force. There would also be stability provided against medial subluxation by the central body of the meniscus, based on the consistent pressures and forces at all flexion angles and loading conditions. However there is little data on applied forces in vivo, to determine the importance of the amount of stability provided.

The results need to be considered with regard to the limitations of the study. The average peak forces during walking have been measured on instrumented knees at about 2000 N (Heinlein et al., 2009; D’Lima et al., 2011) with approximately 1200 N on the medial side (Halder et al., 2012). Our load of 250 N is much less than this, which was based on the strength and durability of the specimens during the prolonged test regime. However, in a previous study using pressure sensitive film (Ahmed and Burke, 1983) loading at 445 N and 1335 N total force on the knee produced close to identical meniscal/total force ratio for their range of flexion. Nevertheless if our tests had been carried out at higher compressive loads, this might have resulted in increased shear force transmission, because the increased cartilage compression would allow increased radial expansion causing meniscal stiffening. This area of the effect of compressive load on stability merits further investigation by experiment and modelling. The tibias were mounted vertically with respect to their proximal intramembraneous canal, but this may have resulted in a few degrees of medial slope in the frontal plane, which may be physiologic but could have introduced a small medial shear force. The flexion motion of the femur on the tibia was continuous in 10–15 s for the full flexion range. Tibial deformation could progress in that time (Hosseini et al., 2010) but much less so on the femur due to the motion. The knee specimens were in an older range, and while the menisci and the cartilage did not show degeneration visually, there would likely be reduced material properties of cartilage and meniscus compared with younger specimens. In testing passive knee specimens, there would be no radially inward forces exerted by the capsular structures and muscles which could potentially provide enhanced stability and load-bearing to the menisci. Finally, we assumed that the regions of the meniscus and uncovered cartilage remained constant throughout flexion, although this may be a reasonable assumption based on the small meniscal contact displacements, as determined in the motion analysis of these knees (Walker et al., 2015). A number of assumptions were made in estimating the shear forces transmitted between the meniscus and the femur, but treating the meniscus as part of the tibial surface, albeit with a flexible attachment, is realistic and the results appear to be reasonable.

Overall, our results were consistent with the previous studies referred to in Table 1 in terms of the contact areas, pressures, and percentage of load carried by the meniscus. As emphasized particularly by Ahmed and Burke (1983), there was considerable variation between specimens. This might imply that some knees are inherently more at risk of cartilage overload and damage than others. In cases where the meniscus carried less force, the uncovered cartilage would experience increased force, and this is the area where osteoarthritic lesions most frequently occur (Arno et al., 2012). However, the cartilage damage also spreads medially. Our data was consistent with medially-biased contacts in most of the specimens, which could exert higher medially-directed forces on the central body explaining meniscal extrusion and medial femoral subluxation (Gale et al., 1999; Khamaisy et al., 2014), especially when the meniscal material became degenerate (Arno et al., 2013a,2013b).

The results of the study have implications to the AP stability of the medial side. Towards extension, the high forces on the anterior horn will play an important role in limiting anterior femoral displacement, particularly because the meniscus would need to move up the extension facet of the anterior tibia (Lankester et al., 2008). Also, due to the high force on the anterior horn, the required forces in the posterior capsule and muscles will be greatly reduced in preventing hyperextension. The posterior horn evidently played an important role after 30 degrees of flexion. However, previous studies have shown only a secondary role in AP stability from the menisci for loaded and unloaded conditions (Shoemaker and Markolf, 1986) although data was obtained only at 20 degrees flexion. In another study (Arno et al., 2013a,2013b), progressive removal of the posterior horn led to a more posterior position of the femur on the tibia, but also increased the AP laxity. Nevertheless, several studies have demonstrated reduced AP laxity when compressive loads are applied (Shoemaker and Markolf, 1986; Luger et al., 1997) but the mechanism itself has not been fully explained. Our studies here suggest that the meniscus may well contribute in part, a possibility that merits further research.

Relating our data to the progression of medial osteoarthritis, in most cases, the area of initial degeneration occurs on the uncovered cartilage. From our data, this area would carry just less than others. In cases where the meniscus carried less force, this would be consistent with the high loads and pressures on the anterior horn towards full extension especially in cases with a high anterior tibial slope.

Overall our study has confirmed several previous studies, but has added additional data of the loading on the different regions of the meniscus, and under conditions of shear loading as well as axial compression. In addition, evidence for the stabilizing role of the medial meniscus; the anterior horn at low flexion angles, and the posterior horn after 30 degrees flexion; has been demonstrated.
Conflict of interest

None.

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References


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