

The sensitivity of tibiofemoral contact pressure to the size and shape of the lateral and medial menisci

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Abstract

In an effort to prevent degeneration of articular cartilage associated with meniscectomies, both meniscal allografts and synthetic replacements have been studied. A number of biomechanical criteria may be important for a meniscal replacement to restore normal tibiofemoral contact pressure in the knee joint and hence be clinically successful. One of these criteria is geometric similarity. The objectives of the current study were to: determine the sensitivity of the contact variables of the tibial plateau to the transverse depth and width of both the lateral and medial menisci; determine the sensitivity of the contact variables of the tibial plateau to the cross-sectional width and height of the lateral and medial menisci; and determine the tolerances on each of the four parameters for both menisci. To satisfy these objectives, a previously developed finite element model of the tibiofemoral joint was used to compute the contact pressure distribution on the tibial plateau. The effect of the above-mentioned geometric parameters on the contact behavior was studied by perturbing the finite element model. Results showed that the contact variables are similarly sensitive to both the transverse and cross-sectional parameters of the menisci. Additionally the medial meniscal parameters have a greater effect on the contact variables than do the lateral meniscal parameters. Finally, less than a 0.5 mm change in the medial meniscal height and greater than a 1 mm change in the lateral meniscal height could be tolerated before the relative difference in the contact variables from those for the original geometry exceeded 10%. Thus in the design or selection of meniscal replacements, each of the four parameters should be measured when sizing a replacement tissue. Also tighter tolerances should be placed on the medial meniscal parameters compared to the lateral meniscal parameters.

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Introduction

Injury to the meniscus is commonplace with complex tears being the most common, accounting for approximately 30% of all tears [27]. Because complex tears usually occur in the avascular region and cannot be repaired with consistent success, the standard treatment is partial meniscectomy [24,30]. This treatment may alleviate clinical symptoms in the short term, but in the longer term partial meniscectomy yields degenerative changes within the joint both in animal models [7,8] and

in humans [6,9,28]. Presumably degenerative changes are caused by an increase in the tibiofemoral contact pressure that accompanies partial meniscectomies [2,17].

Meniscal replacements, and most notably meniscal allografts, are being investigated as a means for restoring the contact pressure distribution to normal and hence possibly either preventing or delaying joint degeneration. To restore the normal contact pressure distribution, however, a number of biomechanical criteria must be satisfied, one of which is geometrical similarity. Geometrical similarity dictates the degree of conformity between the menisci and the femoral condyles. The degree of conformity enables the menisci to distribute the contact force over a greater area, limiting the contact pressure developed on the articular cartilage.

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The size and shape of the menisci should be, therefore, important determinants of the contact pressure distribution [23].

Tissue banks that supply allograft menisci for transplantation recognize the importance of geometrical similarity and typically attempt to match donor tissue to that of the recipient by measuring dimensions from either roentgenograms or MR images of the recipient's knee [21]. Dimensions are usually only measured in the transverse plane, thus assuming implicitly that only the transverse geometric features and not the cross-sectional geometric features are important determinants of the contact pressure distribution. However, recent studies using allografts selected according to this procedure have demonstrated that size and shape are important determinants of the contact pressure [1] and establish the need to improve upon the procedures currently used by tissue banks in selecting donor menisci.

Many independent parameters, including both transverse and cross-sectional measurements, are necessary to quantitatively describe the solid geometry of each meniscus owing to the complexity in their shape [16]. Inasmuch as menisci vary widely in the values of these geometric parameters [16], and tissue banks have a finite inventory of allografts for meniscal replacement, a perfect match in size and shape is impractical. It is important, therefore, to determine the parameters that are most critical to match and the corresponding tolerances that will restore the contact pressure distribution to within some allowable deviation from normal. Recent experimental studies have demonstrated that some cross-sectional parameters, particularly the width and height, are better predictors of contact pressure distribution than others [18]. Therefore, the objectives of the current study were to: determine the sensitivity of contact variables of the tibial plateau to the transverse depth and width of both the lateral and medial menisci; determine the sensitivity of the contact variables of the tibial plateau to the cross-sectional width and height of the lateral and medial menisci; and determine tolerances on each of the four parameters describing the size and shape of each meniscus. To satisfy these objectives, a previously developed finite element model (FEM) of the tibiofemoral joint of a human cadaveric knee [13,14] was used to compute the contact variables on the tibial plateau as the parameters describing the transverse and cross-sectional geometry were varied. The contact variables of interest were the maximum pressure, the mean pressure, the contact area, and the location of the center of pressure.

Methods

One human, fresh-frozen cadaveric knee was obtained from the right leg of a 30-year-old male. An FEM of the cadaveric knee joint was created as previously described [13]. Briefly, the FEM was generated from a 3-D laser coordinate digitizing system [15] that imaged

the cartilage and menisci with an error of less than 8 μm . The model included both the femoral and tibial cartilage, both the medial and lateral menisci and their horn attachments, the anterior cruciate ligament, the transverse ligament, and the deep medial collateral ligament (Table 1). The bones were treated as rigid since a previous study confirmed that this simplification had no substantive effect on the contact variables [13]. The anterior cruciate and deep medial collateral ligaments were modeled as one-dimensional nonlinear springs [5,20,26,33], requiring a nonlinear stiffness parameter (k), and a reference strain (ϵ_r), where reference strain is the initial strain in the reference position (i.e., full extension). Both of these ligaments were modeled with anterior and posterior bundles. The transverse ligament and horn attachments were modeled as linear springs. The cartilage was considered as a linearly elastic and isotropic material while the menisci were linearly elastic and transversely isotropic. These constitutive relations for the cartilage and menisci have been previously justified [13,14].

The material parameter values for the menisci were those that were optimized to give the best match between the contact variables determined from the finite element solution and those determined from direct experimental measurements of contact pressures using pressure sensitive film [14]. The lateral collateral and posterior cruciate ligaments were added to the model for this study and were modeled as one-dimensional nonlinear springs [5,20,26,33]. Previous work indicated that the addition of these ligaments did not substantially affect the computed contact variables at zero degrees of flexion with the native meniscal geometry [14].

With all degrees of freedom unconstrained except flexion angle, the model was compressed to a load level of 1200 N at 0 degrees of flexion, and the contact variables described below were determined from the model solution. Compressive loading was applied according to the coordinate system of Grood and Suntay [11] along the functional tibial rotation axis as determined using the procedures described by Bach and Hull [3]. This axis coincides approximately with the axis of the intramedullary canal. The FEM has been validated for compression loading between 400 and 1200 N and between 0 and 15 degrees of flexion [14].

To determine the range of variation in geometric parameters, the data presented previously for the average and standard deviations of geometrical parameters taken from 10 medial and 10 lateral human menisci were used [15]. Each parameter for the menisci of the cadaver knee in the present study fell within the range of \pm one standard deviation (SD) of the average value for the corresponding parameter from the previous study. Because of this close comparison, all parameters in the current study were varied over a range of \pm one standard deviation. Each parameter was investigated independently. When a change in a parameter resulted in interference between the superior surface of the menisci and the femoral cartilage, the joint in the FEM was distracted to remove this interference. Before the compressive load of 1200 N was applied, however, the joint was returned back to the undistracted position, creating a pre-load on the menisci. This mimicked what would occur when a meniscal replacement is implanted during surgery.

The transverse parameters of width and depth were studied for both the lateral and medial menisci independently. Although a previous study documented the transverse width of the menisci in the anterior and posterior regions [16], we varied the transverse width simultaneously in the two regions. The width was varied \pm 3 mm from starting values of 24.2 and 25.5 mm for the anterior and posterior regions of the medial meniscus, and 27.0 and 28.9 mm for the anterior and posterior regions of the lateral meniscus, respectively (Fig. 1a). The locations of the horn attachments and the cross-sectional geometry were not altered as the width was changed. The transverse depth of the menisci was varied \pm 4 mm from a starting value of 35.6 mm for the medial meniscus and 31.4 mm for the lateral meniscus (Fig. 1b). Again, the cross-sectional geometry and horn attachments were not altered with a change in depth.

The cross-sectional parameters were varied uniformly throughout the entire meniscus, rather than in regions as described previously [16]. The cross-sectional width was varied \pm 3 mm while the cross-sectional height was varied \pm 1.6 mm. The initial cross-sectional widths of the medial meniscus were 9.8, 11.7, and 14.3 mm in the anterior, middle, and posterior regions, respectively. Similarly, the initial cross-sectional heights of the medial meniscus were 7.0, 7.2, and 6.6 mm in the

Table 1
Material parameters for model components

Model component	Constitutive relation	Material parameters
Femoral/tibial cartilage	Linearly elastic, isotropic	$E = 15 \text{ MPa}$, $\nu = 0.475$
Lateral/medial menisci	Linearly elastic, transversely isotropic	$E_{\text{axial/radial}} = 20 \text{ MPa}$, $E_{\text{circum}} = 150 \text{ MPa}$, $\nu_{\text{in-plane}} = 0.2$, $\nu_{\text{out-of-plane}} = 0.3$, shear modulus = 57.7 MPa
Anterior cruciate ligament (ACL)	1-D nonlinear spring	Anterior bundle: Reference strain = 0.06 Nonlinear stiffness = 5000 N Posterior bundle: Reference strain = 0.10 Nonlinear stiffness = 5000 N
Medial collateral ligament (MCL)	1-D nonlinear spring	Anterior bundle: Reference strain = 0.0 Nonlinear stiffness = 4000 N Posterior bundle: Reference strain = 0.0 Nonlinear stiffness = 4000 N
Transverse ligament (TL)	1-D linear spring	Stiffness = 900 N/mm
Horn attachments	1-D linear spring	Stiffness = 2000 N/mm
Posterior cruciate ligament (PCL)	1-D nonlinear spring	Anterior bundle: Reference strain = -0.24 Nonlinear stiffness = 9000 N Posterior bundle: Reference strain = -0.03 Nonlinear stiffness = 9000 N
Lateral collateral ligament (LCL)	1-D nonlinear spring	Anterior bundle: Reference strain = -0.25 Nonlinear stiffness = 2000 N Superior bundle: Reference strain = -0.05 Nonlinear stiffness = 2000 N Posterior bundle: Reference strain = 0.08 Nonlinear stiffness = 2000 N

anterior, middle, and posterior regions, respectively. The initial cross-sectional widths of the lateral meniscus were 11.9, 12.0, and 11.6 mm in the anterior, middle and posterior regions, respectively. The initial cross-sectional heights were 6.5, 7.2, and 6.0 mm in the anterior, middle, and posterior regions, respectively. Each cross-sectional dimension throughout the meniscus was varied linearly with a zero percent change at the inner border and a 100% change at the peripheral border (Figs. 2a and b).

Four different contact parameters were determined for each case: maximum pressure, mean pressure, contact area, and location of the center of pressure (anteroposterior (AP) and medial/lateral (ML)). The location of the center of pressure was determined in an anatomic-based coordinate system. To create this system, first a line was drawn parallel to the posterior osteochondral junction of the proximal tibia to define the ML direction. The AP direction was defined as perpendicular to this line. The origin was placed at half the maximum AP distance and half the maximum ML distance. In this coordinate system, the anterior and medial directions were positive.

Changes in contact behavior were computed as the root-mean-square normalized difference (RMSND):

$$\text{RMSND} = \sqrt{\frac{\sum_{i=1}^m (D_i)^2}{m}} \quad \text{with } m = 4 \quad \text{and}$$

$$D_i^2 = \left(\frac{X_{n,i} - X_{r,i}}{X_{r,i}} \right)_{\text{lateral}}^2 + \left(\frac{X_{n,i} - X_{r,i}}{X_{r,i}} \right)_{\text{medial}}^2$$

where the reference values for $X_{r,i}$ were determined from the original geometry, and the values for $X_{n,i}$ were computed from the solution with the new geometry. The ML and AP locations of the center of pressure were combined by weighting each only 50% to combine them into one quantity, so that the location of the center of pressure was given equal weighting to the maximum pressure, mean pressure, and contact area. In evaluating the results of the geometry analysis, an increase in the RMSND of 10% was considered important.

Results

Both of the pressure-related contact variables (maximum pressure and mean pressure) on the medial tibial plateau were more sensitive to the transverse parameters of the medial meniscus than the cross-sectional parameters of the medial meniscus. The maximum pressure was most sensitive to the transverse width, particularly decreases in width; decreasing the transverse width 1 SD increased the maximum pressure by 0.93 MPa or 27%, whereas increasing the width decreased the maximum pressure by 8.8% at most (Fig. 3). Likewise, the mean

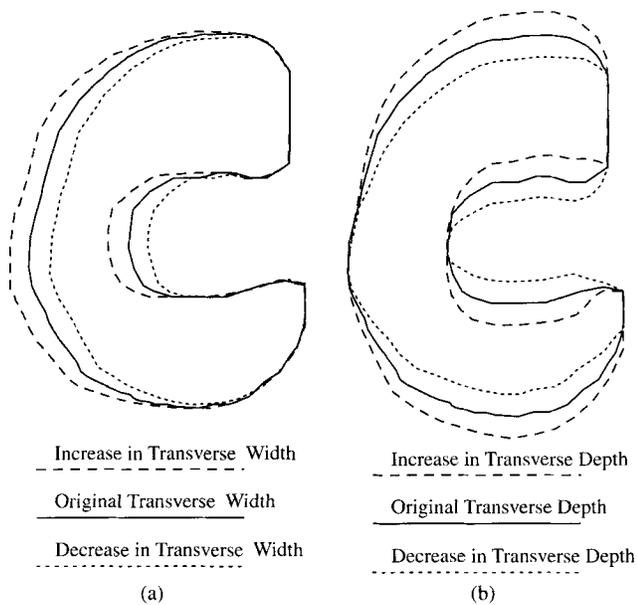


Fig. 1. Schematic illustrating how the transverse widths (a) and transverse depths (b) of the menisci were changed.

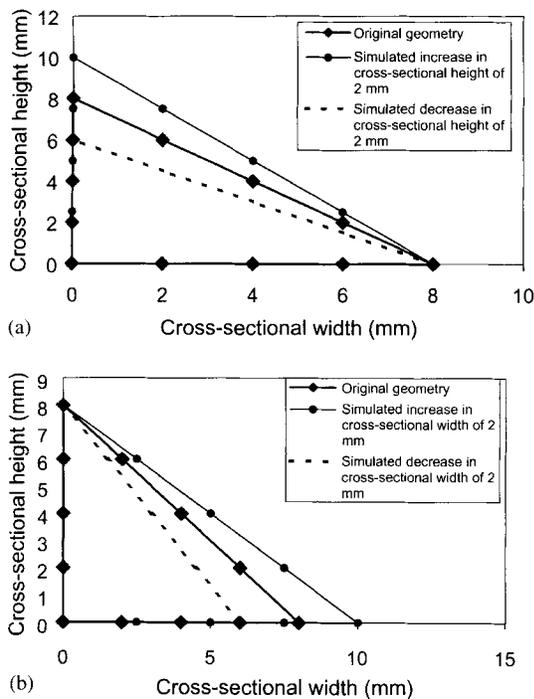


Fig. 2. Schematic illustrating how the cross-sectional widths (a) and cross-sectional heights (b) of the menisci were changed.

pressure was most sensitive to the transverse width; decreasing the width by 1 SD increased the mean pressure by 0.56 MPa or 42%, and increasing the width decreased the mean pressure by 0.36 MPa or 26%.

The area-related variables (contact area and location of the center of pressure) were more sensitive to the cross-sectional parameters than the transverse para-

eters of the medial meniscus. The contact area on the medial tibial plateau was most sensitive to the cross-sectional height of the medial meniscus (Fig. 3). Increasing the cross-sectional height by 1 SD increased the contact area by 104 mm² or 29%, and decreasing the height by 1 SD decreased the area by 135 mm² or 38%. Likewise the AP location of the center of pressure on the medial tibial plateau was most sensitive to the cross-sectional height of the medial meniscus. Decreasing the height shifted the AP location by up to 6 mm anteriorly. Finally, either increasing or decreasing the cross-sectional width of the medial meniscus shifted the ML location of the center of pressure by about 6 mm laterally.

Changing the geometric parameters of the medial meniscus also affected the contact variables on the lateral tibial plateau, and all of the contact variables were more sensitive to the cross-sectional parameters of the medial meniscus than the transverse parameters. For example, increasing the cross-sectional height by 1 SD decreased the maximum pressure by 1.17 MPa or 32% and also decreased the mean pressure by 0.39 MPa or 24%. Neither of the area-related contact variables of the lateral tibial plateau was as sensitive to the cross-sectional parameters of the medial meniscus as the pressure-related variables.

The tolerances on individual parameters describing the geometry of the medial meniscus were determined by considering the changes in the contact variables of both tibial plateaus combined via the RMSND. For increases/decreases in parameters of the medial meniscus, the tolerances for an RMSND of 10% were +0.4 SD/−0.5 SD (+1.6 mm/−2.0 mm) for the transverse depth, +0.2 SD/−0.3 SD (+0.6 mm/−0.9 mm) for the transverse width, +0.25 SD/−0.2 SD (+0.4 mm/−0.3 mm) for the cross-sectional height, and +0.35 SD/−0.30 SD (+1.0 mm/−0.9 mm) for the cross-sectional width (Fig. 4).

In general the contact variables of the lateral tibial plateau were not as sensitive to the geometric parameters of the lateral meniscus (Fig. 5) as the contact variables of the medial tibial plateau were to the geometric parameters of the medial meniscus (Fig. 3). The maximum pressure of the lateral plateau was most sensitive to the transverse depth of the lateral meniscus, with an increase in the maximum pressure of 0.74 MPa or 20%. The mean pressure of the lateral plateau was most sensitive to the cross-sectional height of the lateral meniscus. Decreasing the height by 1 SD increased the mean pressure by 0.37 MPa or 24%. The contact area was most sensitive to the transverse width (followed closely by the cross-sectional width), with a decrease in area of 60 mm² or 16%. Finally, the location of the center of pressure on the lateral plateau was not particularly sensitive to any of the geometric parameters of the lateral meniscus, shifting medially by about 2.5 mm at most for an increase in the cross-sectional height.

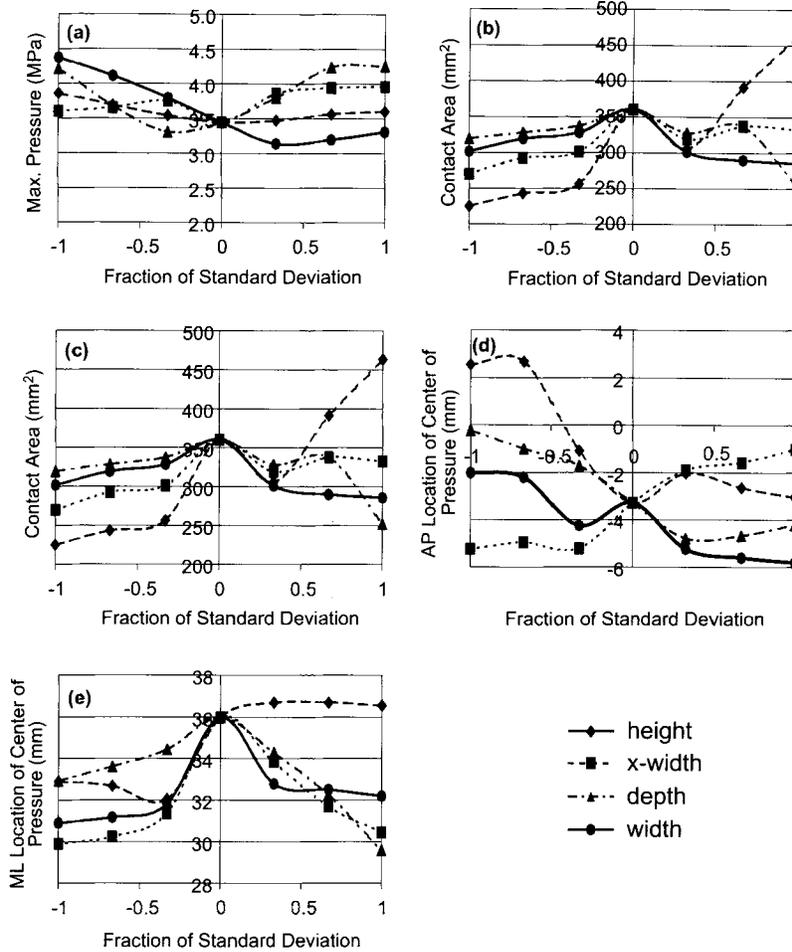


Fig. 3. Changes in contact variables of the medial tibial plateau with changes in parameters describing the geometry of the medial meniscus. Five contact variables are plotted (a–e), and on each plot are the values of one of the contact variables in response to each of the four geometric parameters that were changed. Positive is anterior and medial for the AP and ML locations of the center of pressure, respectively.

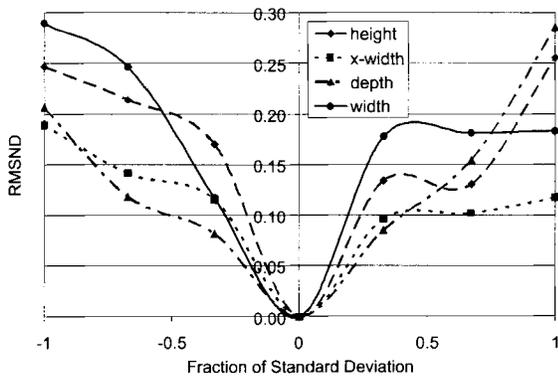


Fig. 4. RMSND values for both the medial and lateral tibial plateaus combined with changes in the parameters describing the geometry of the medial meniscus.

The contact variables of the medial tibial plateau were affected by changes in the geometric parameters of the lateral meniscus. The maximum pressure of the medial plateau was most sensitive to the transverse

depth of the lateral meniscus; decreasing the depth by 1 SD increased the maximum pressure by 0.43 MPa or 12%. The mean pressure of the medial plateau was most sensitive to the cross-sectional height of the lateral meniscus; increasing the cross-sectional height decreased the mean pressure by 0.09 MPa or 7%. The contact area was most sensitive to the transverse depth; increasing the depth by 1 SD decreased the contact area by 56 mm² or 15%. The ML location of the center of pressure on the medial plateau was sensitive to the cross-sectional height of the lateral meniscus with a 1 SD increase in height causing a 4.5 mm lateral shift in the location of the center of pressure.

For increases/decreases in parameters of the lateral meniscus, the tolerances for an RMSND of 10% were +1.0 SD/–0.5 SD (+4.0 mm/–2.0 mm) for the transverse depth, +1.05 SD/–0.55 SD (+3.2 mm/–1.6 mm) for the transverse width, +0.85 SD/–0.8 SD (+1.4 mm/–1.3 mm) for the cross-sectional height, and +1.05 SD/–0.9 SD (+3.2 mm/–2.7 mm) for the cross-sectional width (Fig. 6).

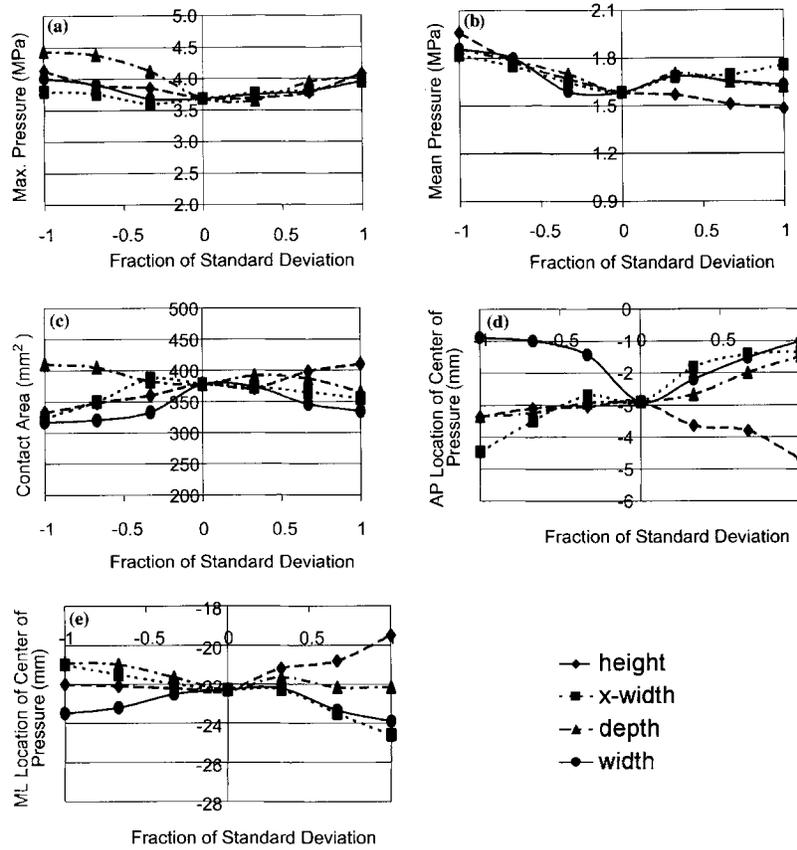


Fig. 5. Changes in contact variables of the lateral tibial plateau with changes in parameters describing the geometry of the lateral meniscus. Five contact variables are plotted (a–e), and on each plot are the values of one of the contact variables in response to each of the four geometric parameters that were changed. Positive is anterior and medial for the AP and ML locations of center of pressure, respectively.

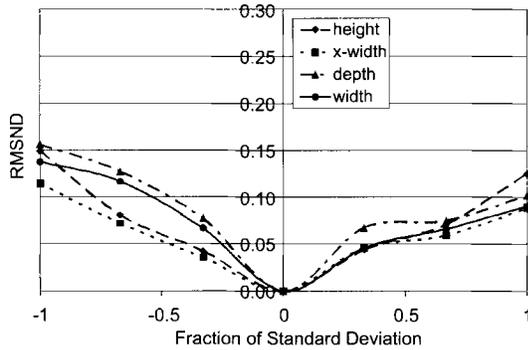


Fig. 6. RMSND values for both the medial and lateral tibial plateaus combined with changes in the parameters describing the geometry of the lateral meniscus.

Discussion

The objectives of the current study were to determine the sensitivity of the contact variables of the tibial plateau to both transverse and cross-sectional parameters describing the geometry of both menisci and to determine the tolerances on these parameters. A previously

developed FEM of the tibiofemoral joint of a human cadaveric knee, including both menisci, was used to compute the contact stress distribution on the tibial plateau. The key findings were that: the contact behavior was equally sensitive to changes in the transverse and cross-sectional parameters; the contact behavior was more sensitive to changes in the parameters describing the geometry of the medial meniscus than parameters describing the geometry of the lateral meniscus; and less than a 0.5 mm change in the medial meniscus height could be tolerated before the RMSND exceeds 10%.

Several methodological issues could influence our results and their interpretation. Increased contact stresses have been hypothesized to lead to osteoarthritis (OA) [2,17], but meniscal displacements and motion have not been directly linked to OA. Thus, the interpretation of changes in these quantities is difficult to assess. Also, while changes might occur in either the motion or displacement of the meniscal replacement compared to the native meniscus, it is ultimately the cartilage that breaks down during osteoarthritis. Hence, the altered loading state that the cartilage may experience due to changes in meniscal geometry is of primary interest.

Some of the meniscal ligaments (coronary, ligament of Humphrey, and ligament of Wrisberg) were not included in the FEM. Previous work showed that cutting of the coronary ligament did not affect the contact pressure distribution of the knee joint under loading conditions similar to those that we applied [22]. Moreover, the ligaments of Humphrey and Wrisberg have been found in only 40% of knee specimens [12]. Due to the variable presence of these ligaments in knees, they were not included in the model.

Even though Haut et al. quantified four transverse parameters and 15 cross-sectional parameters (five parameters over three regions) to describe meniscal geometry [16], only a subset of these parameters was investigated in the current study based on the conclusions of a recent experimental study [18]. An overall transverse width was used, and the ratio of enclosure (a transverse parameter computed as the ratio of the distance between the two horns divided by the depth), was not studied thus limiting the number of transverse parameters to two. Similarly, the cross-sectional geometry was either increased or decreased over the entire meniscus rather than over regions [16]. Noting recent experimental results, which show the cross-sectional width and height to be strong predictors of contact variables [18], the remaining three cross-sectional parameters—bulge, slope and height ratio—were not considered.

Because we are aware of no studies that indicate how these four meniscal parameters vary between menisci, we varied the width linearly with the height, and the height was varied linearly with the width. However, we do not know how these parameters vary from meniscus to meniscus. Each meniscus probably scales differently, so the choice of the scaling method is probably representative of some but not all menisci.

Placing a practical upper bound on the RMSND is difficult since no data exist to establish what relative changes in contact variables accelerate the rate of cartilage wear. Nevertheless, a 10% difference in contact variables represents a significant reduction for the changes in the contact variables seen for the meniscectomized knee [2,4,6,19,28]. Peak contact stresses on the lateral and medial articular surfaces of the tibia increase over 300% in the meniscectomized knee [4,19,25,29], and contact areas decrease by 50% [4,10,19,25]. Thus, it is reasonable to assume that changes of only 10% would reduce the rate of cartilage wear relative to that of the meniscectomized condition.

While each parameter was isolated and investigated individually for its effect on contact behavior, we did not investigate combined effects from variations in more than one parameter. When selecting a meniscal allograft, more than one geometric parameter will probably not match those of the original meniscus, and thus a combined effect will be experienced. The tolerances may

either increase or decrease due to the combined effects. Therefore, it would be useful to investigate the combined effects of the parameters described in this study on the contact variables.

Many experimental studies have hypothesized that poor sizing of meniscal allografts [1,25,31,32] alters contact patterns and leads to unfavorable results such as osteophyte formation, but this is the first study to isolate specific geometrical parameters and determine their role in the contact behavior of the joint. Our data show that large changes in contact variables occurred when each of the four geometric parameters was varied ± 1 SD. Depending on the parameter, the RMSND ranged from a maximum of 29% to a minimum of 19%, which is still substantial (Fig. 4). This suggests that the contact pressure distribution of the joint as a whole and particularly the medial tibial plateau is highly dependent on all of the parameters studied. Thus in either the design of meniscal replacements or selection of meniscal allografts, it is important to define tolerances on all of the variables studied.

Because the sensitivity of the contact variables was greater for parameters describing medial meniscal geometry than parameters describing the lateral meniscal geometry, the tolerances on all parameters are tighter for the medial meniscus. The tighter tolerances are likely due to differences in the ratios of enclosure of the two menisci and the peripheral attachment of the medial meniscus. The ratio of enclosure is a transverse parameter computed as the ratio of the distance between the two horns divided by the depth [16]. The medial meniscus has a larger ratio of enclosure, which may result in a different and smaller degree of conformity with the femoral condyle when compared to the more circularly shaped lateral meniscus. In addition, the lateral meniscus shifts more freely on the tibia than the medial meniscus, which is attached to the deep MCL and also extends to reach the peripheral edges of the plateau more so than the lateral meniscus. The freedom of movement combined with the greater conformity with the femoral condyle may allow the lateral meniscus to tolerate greater variations in the transverse parameters than the medial meniscus.

We demonstrated previously that the contact pressure is also sensitive to the material properties of the menisci and in particular to both the circumferential and the radial/axial moduli [14], raising the question of whether changes in material properties affect tolerances on meniscal geometric parameters. Regardless of the answer, the two general conclusions of the current study (i.e., that contact behavior is equally sensitive to both transverse and cross-sectional parameters and that parameters describing the geometry of the medial meniscus affect contact behavior more than those of the lateral meniscus) would be expected to remain valid for menisci with different material properties.

In summary, using an FEM of the knee to compute tibiofemoral contact, we demonstrated that contact behavior was similarly sensitive to relative changes in both the transverse and cross-sectional parameters. Therefore in the selection of meniscal allografts for a recipient knee, allografts should be selected such that all of these parameters are within defined tolerances. Likewise, in the design of a synthetic replacement, it would be advantageous to match all four geometrical parameters to those of the native meniscus. Changes in parameters describing the geometry of the medial meniscus also had a more significant effect on contact behavior than changes in parameters describing the geometry of the lateral meniscus. Thus, tighter tolerances as indicated should be placed on the medial meniscal parameters.

Acknowledgements

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