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The Combined Effect of Frontal Plane Tibiofemoral Knee Angle and Meniscectomy on the Cartilage Contact Stresses and Strains

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Abstract—Abnormal tibiofemoral alignment can create loading conditions at the knee that may lead to the initiation and progression of knee osteoarthritis (OA). The degenerative changes of the articular cartilage may occur earlier and with greater severity in individuals with abnormal frontal plane tibiofemoral alignment who undergo a partial or total meniscectomy. In this investigation, subject specific 3D finite element knee models were created from magnetic resonance images of two female subjects to study the combined effect of frontal plane tibiofemoral alignment and total and partial meniscectomy on the stress and strain at the knee cartilage. Different amounts of medial and lateral meniscectomies were modeled and subject specific loading conditions were determined from motion analysis and force platform data during single-leg support. The results showed that the maximum stresses and strains occurred on the medial tibial cartilage after medial meniscectomy but a greater percentage change in the contact stresses and strains occurred in the lateral cartilage after lateral meniscectomy for both subjects due to the resultant greater load bearing role of the lateral meniscus. The results indicate that individual's frontal plane knee alignment and their unique local force distribution between the cartilage and meniscus play an important role in the biomechanical effects of total and partial meniscectomy.

Keywords—Finite element analysis, Varus, Valgus, Meniscectomy, Knee.

INTRODUCTION

Osteoarthritis (OA) is a degenerative disease of articular cartilage that affects millions of people.[59] Several reported symptoms of knee OA include limited mobility, pain and joint deformity.[7] Experimental studies have shown that excessive loading on the cartilage can cause damage that could lead to subsequent OA.[1,37,53,55] Local biomechanical factors may severely affect the initiation and progression of OA due to abnormal loading conditions at the knee joint.[7,17,24,26,28,59,60] Frontal plane tibiofemoral angle and meniscectomy have been shown to lead to knee OA initiation and progression.[11,13,24–28,34,38,40,41,47,59] Allen *et al*.[2] found a significant increase in knee OA after meniscectomy with pre-existing abnormal tibiofemoral alignment. Furthermore, meniscectomy may alter the frontal plane tibiofemoral angle after removal of the meniscus and cause further change in the varus moment and corresponding load distribution at the knee.[9] In this study, the finite element method was utilized to compare the stress and strain in the knee cartilage between two subjects with different frontal plane tibiofemoral angle after different amounts of medial and lateral meniscus removal.

The tibiofemoral alignment at the knee is measured by the angle formed by the intersection of the anatomical axes of the femur and the tibia, Fig. 1a. A normal knee has a tibiofemoral angle of approximately 5–7° valgus.[32,35] Deviation from this angle leads to a varus aligned knee, also described as ‘‘bow-legged’’ with an angle less than 5° valgus. A valgus aligned knee is described as ‘‘knocked-kneed’’ with an angle greater than 7° valgus.

During walking, approximately 70–75% of the load passes to the medial compartment of the knee joint.[3,5,30] The varus knee moment, or adduction moment, is the primary factor in the distribution of the force to the medial compartment of the knee joint during normal gait.[3,5,68] A varus aligned knee will have a moment that increases the loading on the medial compartment of the knee and a valgus knee will have a moment that increases the loading on the lateral compartment of the knee during single-leg support. However, only in extremely malaligned valgus knees (approximately 15° valgus) will the lateral knee compartment experience more load than the medial knee compartment.[33,59] Individuals who demonstrate a varus knee alignment have been shown to increase the

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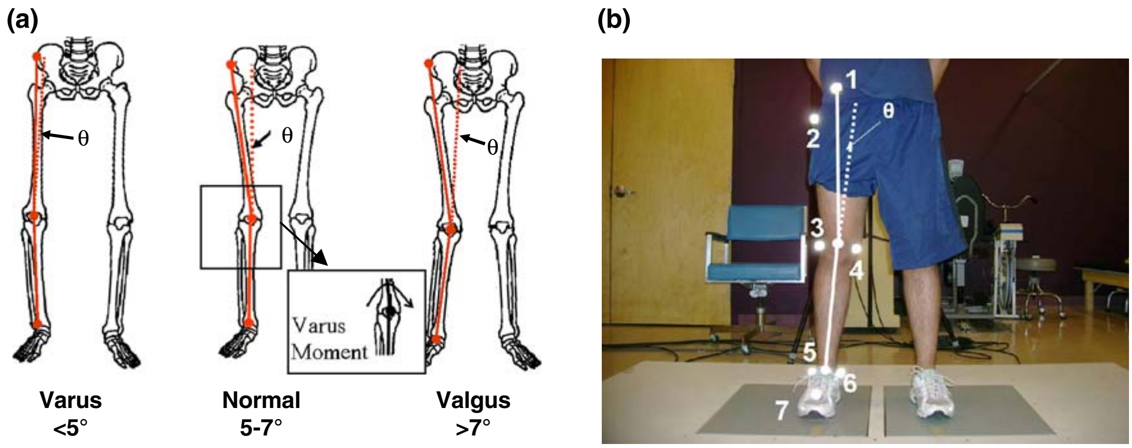

FIGURE 1. (a) Frontal plane tibiofemoral angle measured at the intersection of the axis of the femur and the tibia and the varus moment at the knee. Varus or ''bow-legged'' alignment has an angle less than 5°, normal alignment is 5–7° and valgus or ''knocked-kneed'' is greater than 7°. (b) The marker set used in the motion analysis experiment. Passive reflective markers are placed at anatomical locations at (1) anterior superior iliac spine (ASIS), (2) the greater trochanter, (3) the lateral femoral condyles, (4) the medial femoral condyles, (5) the lateral malleoli, (6) the medial malleoli, (7) the head of the second metatarsal, and (8) the second sacral vertebra (S2) of the spine.

risk of medial compartment OA progression and valgus alignment has been shown to increase the risk of lateral OA progression in as little as 18 months.[59]

The meniscus is an important component of the knee as it provides joint stability, load bearing, joint lubrication and shock absorption.[8,27,38,52,69] Damage and tears to the meniscus may occur due to trauma from activities or due to degenerative changes in the menisci that occur with age.[21,31] Untreated meniscus tears may result in knee locking, pain and knee instability.[16,19] However, while meniscus tears may result in reported symptoms of pain, meniscal tears in middle-aged or older adults may also not be symptomatic.[24] In order to preserve as much of the meniscus function as possible, partial meniscectomies are performed to remove the damaged section. Partial and total meniscectomies alter the kinematics of the knee and load distribution and decrease the contact area at the knee causing greater magnitude of compressive stress and shear stress to the articular cartilage which can damage the knee cartilage and lead to subsequent OA.[18,27,34,38–41,46,47] Long-term follow up studies have shown that the prevalence on knee OA increases with meniscectomy and individuals with meniscectomy show OA in the knee 10–20 years earlier than individuals with primary knee OA with no meniscectomy.[11,21–23,34,54,56,57] The articular cartilage degeneration and subsequent OA is most likely due to the changes in the mechanical loading after meniscectomy.[49,64,67] Studies have shown cartilage deterioration is more frequent after lateral meniscectomy than medial meniscectomy.[2,33,34] This could be due to the convex shape of the lateral tibial plateau which could increase the local articular cartilage stress and also the greater load carrying role of the lateral meniscus and the lateral tibial plateau being more protected by the meniscus than the medial compartment.[2]

Finite element analysis is a valuable, non-invasive tool which has been used to model the knee to obtain the internal forces, stresses and strains in the bones and soft tissues at different joints.[4,29,44,45,66] Previous finite element studies of meniscectomy include both 2D axisymmetric models and 3D models.[50–52,61,62,64,69] However, these finite element studies applied axial loads and did not consider subject specific physiological loading during functional activities which would provide a more realistic representation of the actual forces and moments within the knee joint. Previously developed 3D finite element models provide a good representation of knee joint contact, however subject specific loading conditions determined from motion analysis and force platform experiments were not utilized in these studies.

In this investigation, 3D finite element knee models of subjects specific with different frontal knee alignment were created using magnetic resonance images (MRI). Finite element analyses were performed to understand the effect of the frontal plane tibiofemoral angle during static single-leg stance on the contact stresses and strains at the knee cartilage after partial and total meniscectomy. In order to enhance and develop a more realistic 3D knee model, the loading conditions were determined from an inverse dynamic analysis and muscle force reduction model from subject specific data. Finite element simulations of various medial and lateral meniscectomy were performed separately. There has been no known study that has quantified the effect of the tibiofemoral alignment combined with meniscectomy on the stress and strain at the knee cartilage during functional activity using subject specific finite element models and loading conditions. The finite element models in this investigation applied an axial load, a posterior shear load and a varus moment that models the physiological loading that occurs during single-leg support. This process could be very beneficial in identifying individuals most susceptible to knee OA and prescribing intervention measures to slow the progression or possibly delay the onset of OA.

METHODS

Two healthy females of the same age (21 years) and similar weight with no history of knee OA or prior knee injury were recruited from the Northeastern University community for this study. Institutional Review Board approval was obtained for experiments along with signed consent from the subjects.

Knee Joint Reactions and Tibiofemoral Angle Assessment

Kinematics and kinetic data were collected with a six camera motion analysis system (EVaRT 5.0, Motion Analysis Corporation, Santa Rosa, CA) with a capture rate of 120 Hz. Two force platforms (models OR6-6-2000, OR6-7-2000, Advanced Mechanical Technology, Inc., Waltham, MA) were used and time synchronized with the motion cameras with data sampled at 1200 Hz. Passive reflective markers were placed on bony landmarks of the body to define the different segments of the leg and the locations of the knee joint and ankle joint center. The markers were placed on the leg the subjects would use to kick a ball at the (1) anterior superior iliac spine (ASIS), (2) the greater trochanter (3, 4) the lateral and medial femoral condyles, (5, 6) lateral and medial malleoli, (7) the head of the second metatarsal, and (8) the second sacral vertebra of the spine (S2), Fig. 1b.

The subject performed three trials of static double-leg and single-leg stance for 10 s intervals. During the double leg-stance condition, the subjects were told to stand with their weight evenly distributed with the second metatarsal of the two feet approximately 30 cm apart. The experiments were monitored in real time to ensure even weight distribution over each leg. The axis of the femur in the frontal plane was defined with the marker at the ASIS and the knee joint center and the axis of the tibia was defined with at the ankle joint center to the knee joint center, Fig. 1b, similar to previous studies which used markers and motion analysis to define the frontal plane angle.[28,58] The tibiofemoral angle θ was assessed while the subjects were in this standard double-leg stance to ensure minimal rotation of the tibia.[14,30] For the single-leg stance condition, the subjects were told to balance their weight on the leg covered with the reflective markers. The subjects were allowed to practice to become familiar to the procedure and to align themselves with the global coordinate system.

Subject specific loading at the knee was determined from a motion analysis and force platform data. A classical inverse dynamic analysis was used to calculate the knee joint reactions forces and moments.[65] In this investigation, only single-leg static stance was used to determine the knee joint reactions to define the loading conditions in the finite element simulations but this method has been successfully used to define the loading conditions during the stance phase of the gait cycle[12] and the loading conditions in FEA studies of the healthy knee joints for single-leg stance.[66]

The contribution of the muscle forces was determined using a muscle force reduction method similar to previous studies by Morrison[48] and Schipplein and Andriacchi.[57] In summary, the muscles across the knee are reduced into three groups (hamstrings, quadriceps and gastrocnemius groups) and the line of action and location of the muscles with respect to the center of the knee joint were taken from the literature[36] and balanced to the external extension/flexion moment obtained from the inverse dynamic analysis to obtain the muscle force contributions. The overall muscle force contributions represent a minimum due to the absence of co-contraction in the reduction model.

Three-Dimensional Finite Element Models

Subject specific 3D finite element knee models were created from sagittal view magnetic resonance images (MRI). The MRI were obtained from Weymouth MRI in Weymouth, MA, USA. The MRI were taken of the subject in the supine, non-load bearing position. MRI were taken in the early morning to avoid the day long weight bearing of the knee joint which may compress the cartilage and reduce the overall thickness. The reduced thickness may affect the results on the finite element study as Li *et al*.[44] saw an 8% variation in cartilage thickness led to 10% differences in the contact stresses.

The method used to construct the 3D models was similar to previously published investigations utilizing 3D knee models for contact kinematics analysis[42] and finite element studies.[43–45] The sagittal MRI were 150 mm $\times$ 150 mm and were spaced 2 mm apart with 256 $\times$ 256 pixel resolution (1.7 pixel/mm) with approximately 45 slices for each subject. The boundaries of the different segments in each 2D MRI were digitized using the solid modeling program Rhinoceros (Rhinoceros 3.0, Seattle, Washington). A 3D point

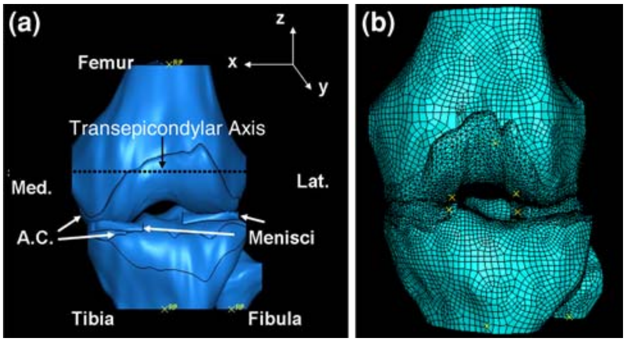

FIGURE 2. (a) Three-dimensional geometry of the left knee, which includes femur, tibia, fibula, articular cartilage and lateral and medial menisci with the dotted line representing the transepicondylar axis and the location where the loading was applied. (b) A typical mesh of the knee geometry.

cloud was constructed by aligning each 2D segment in its respective position. The 3D point cloud was used to define the 3D surface geometry of the individual knee components which included the bones, articular cartilage and meniscus, Fig. 2a.

The 3D knee models were exported into the finite element software package, ABAQUS v 6.5-6 (Simulia, Providence, RI, USA). The material properties were assigned to the different components of the knee based on previously published data. The loading conditions were defined by the results of the motion analysis and force platform data. Forces and moments were applied to the model to simulate static single-leg stance.

The bones were meshed with 4-node bilinear rigid quad elements to represent their surface geometry. For the cartilage and the meniscus, a free meshing technique was used with 4-node linear tetrahedron elements, Fig. 2b. An average element length of 1.25 mm was used after a model convergence analysis was performed. Model convergence was tested by recording four contact variables (normal strain, von Mises stress, contact force and contact area) in the cartilage for various average mesh length size. Parts were meshed with tetrahedron elements with average side length of 1.0, 1.25, 1.5, 2.0, 2.5, 3.0 mm. Adjusting the average mesh size length from 1.25 to 1.0 mm changed maximum strain approximately 0.10%. All other contact variables changed by no greater than 4.8%. For all models, an average mesh length of 1.25 mm was used due to the computational time saved. Contact was defined between surfaces with a friction coefficient of 0.002 which was taken from a previously published study.[15]

Material Properties

The bones in the model included the femur, tibia and fibula. The bones were modeled as rigid because they are much stiffer compared to cartilage and meniscus. Modeling the bones as rigid has been shown to have negligible effect on the contact results ($<2\%$) and drastically saves computational time.[29]

TABLE 1. Material properties used to define the cartilage and meniscus for elastic conditions.

Part	Material properties
Cartilage	Isotropic elastic: $E = 15$ MPa $\nu = 0.45$
Meniscus	Transversely isotropic elastic: $E_1 = E_3 = 20$ MPa, $E_2 = 120$ MPa, $\nu_{13} = 0.2$, $\nu_{12} = \nu_{23} = 0.3$ $G_{12} = G_{23} = 57.7$ MPa, $G_{13} = 8.33$ MPa

For the transversely isotropic meniscus, E_1 and E_2 denote Young's modulus in the axial and radial direction, respectively, and E_3 denotes Young's modulus in the circumferential direction.

Modeling of articular cartilage has varied from study to study. Many studies assume cartilage to behave as a linear elastic material.[29,43–45,49,51,52,67] These assumptions are justified considering that the mechanical response of cartilage does not vary significantly with time at short loading durations.[6,20]

In this investigation the cartilage was modeled as one layer isotropic, elastic and rigidly attached to the bone surface and meniscus was modeled as transversely isotopic elastic with material properties derived from previous published studies,[29,69] Table 1. The menisci were attached to the tibial plateau at the meniscal horns using a set of linear spring elements similar to the methods used by Haut Donahue *et al.*[29] At each horn attachment, ten linear springs with a stiffness of 200 N/mm attached the horn to the tibial plateau. The total stiffness for each horn attachment was 2000 N/mm. A transverse ligament was modeled as a linear spring and attached the anterior horns of the lateral and medial menisci. A single spring element was used with a stiffness of 900 N/mm.

The location of the different ligaments was obtained using MRI and included the anterior cruciate ligament (ACL), posterior cruciate ligament (PCL), medial collateral ligament (MCL) and lateral collateral ligament (LCL). The ligaments were modeled as one-dimensional nonlinear spring elements according to their functional bundles based on actual ligament anatomy. The ACL was modeled as an anteromedial and posterolateral bands. The PCL was modeled as an anterolateral band and a posteromedial band. The MCL was modeled as a superficial portion and a deep (inferior) portion. The superficial portion was subdivided into an anterior portion and a posterior portion, and the LCL was modeled with a similar method. Each functional bundle of the ligaments was represented

TABLE 2. Subject tibiofemoral angle, weight and loading conditions for the FEA model.

Subject	Tibiofemoral angle	Weight (N)	F_z (N)	F_y (N)	M_y (N-m)
1 Varus	0.20°	640	−811	−212	−24.68
2 Valgus	10.34°	704	−854	−164	−22.18

The compressive load, F_z, and shear load, F_y, are a summation of the reaction forces and internal muscle forces determined with the reduction model and the moment, M_y, represents the varus knee moment. Figure 2 shows the orientation of the loading.

with a non-linear spring element with the following piecewise force–displacement relationship[10]

$$f = \begin{cases} \frac{1}{4}k\varepsilon^2/\varepsilon_l & 0 \le \varepsilon \le 2\varepsilon_l \\ k(\varepsilon - \varepsilon_l) & \varepsilon > 2\varepsilon_l \\ 0 & \varepsilon < 0 \end{cases}$$

where f is the applied force, k is the ligament stiffness parameter, ε_l is the non-linear strain parameter and ε is the strain in the ligaments calculated from $\varepsilon = (L - L_0)/L_0$, where L is the ligament length and L_0 is the zero-load length of the ligament. The zero-load length of the element was estimated from $L_0 = L_r/(\varepsilon_r + 1)$, where L_r is the ligament initial length from the MRI and ε_r is the reference strain that was found by Blankevoort *et al.*[10] The same model has been used in the author's previous study to model the healthy knee joint.[66]

Boundary and Loading Conditions

For static single-leg stance simulations, the tibia and the fibula were fixed in all translations and rotations. The femur was held fixed at 0° flexion and all other translations and rotations were unconstrained. The force and moments were applied at the midpoint of the transepicondylar axis, Fig. 2a, and included the axial force, F_z, posterior force, F_y and varus knee moment M_y, Table 2. The compressive reaction force, F_z, and posterior directed shear force, F_y, are a summation of the knee reactions and muscle force contributions. A meniscectomy could cause the thickness of the respective compartment to thin, altering the tibiofemoral angle, which could change the overall varus moment at the knee.[9] However, in this investigation, it was assumed that the varus moment determined from the motion analysis and force platform was held constant for all variations of amount of meniscectomy.

Meniscectomy

Three different types of partial medial meniscectomies were modeled for both subjects as well as a total medial meniscectomy. The three different types of partial medial meniscectomy included a posterior section, a posterior-central section and an extended posterior-central section. Four different lateral meniscectomies were modeled for both subjects as well as a total lateral meniscectomy. Posterior and anterior partial lateral meniscectomies were modeled as well as extended versions of both that removed a greater area of the lateral meniscus from its respective location. Figure 3 shows the different meniscectomies considered in this investigation for both subjects.

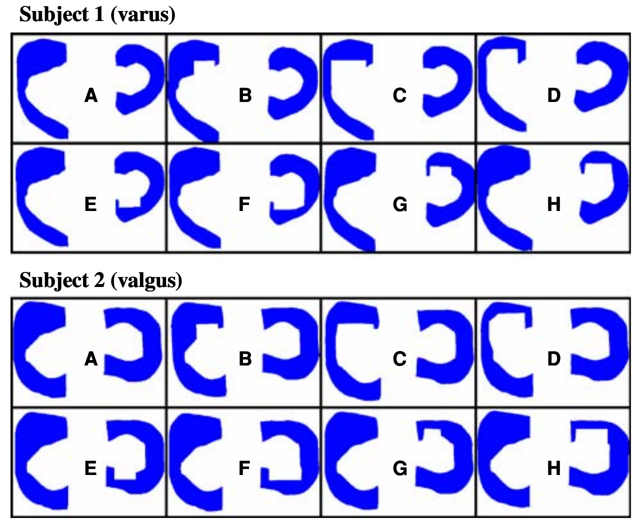

FIGURE 3. The (A) healthy meniscus and the medial meniscectomies, (B) posterior, (C) posterior-central and (D) longitudinal as well as the lateral meniscectomies, (E) anterior, (F) anterior-extended, (G) posterior, and (H) posterior-extended, modeled in FEA simulations for Subject 1 and Subject 2.

Table 3 shows the percentage of the area removed on the superior face of the medial and lateral meniscus for different types of meniscectomy. An attempt was made to remove the same relative amount of area in both subjects for the same type of meniscectomy. However, this proved to be difficult due to the unique geometry of the menisci for each of the subjects. Still, the greatest difference between the two subjects for the same type of meniscectomy (medial longitudinal meniscectomy) was less then 2.81% with an overall average difference of less than 1.80%.

RESULTS

Subject 1 had varus frontal plane tibiofemoral alignment (0.20° valgus, bow-legged) and Subject 2 had valgus alignment (10.34° valgus, knocked-kneed), Table 2. Subject 1 had an overall greater normal force distribution to the medial knee compartment (82% medial knee compartment, 18% lateral knee compartment) compared to Subject 2 (78% medial, 22% lateral) due to the greater varus moment cause by the

TABLE 3. The type of meniscectomy, the area of the meniscus removed and the total contact area for Subject 1 (varus alignment) and Subject 2 (valgus alignment).

Type	% Area removed Medial	% Area removed Lateral	Total contact area (mm^2)	% Medial	% Lateral
Subject 1					
Healthy joint	0	0	758	63.46	36.54
Posterior	6.19	0	751	62.58	37.42
Posterior-central	14.15	0	723	60.72	39.28
Longitudinal	18.07	0	709	59.38	40.62
Total medial	100	0	639	55.40	44.60
Anterior	0	4.01	738	64.23	35.77
Ant-extended	0	7.91	716	66.20	33.80
Posterior	0	7.18	750	63.87	36.13
Post-extended	0	11.31	737	64.99	35.01
Total lateral	0	100.00	607	78.25	21.75
Subject 2					
Healthy joint	0	0	721.00	65.19	34.81
Posterior	8.09	0	696.00	63.22	36.78
Posterior-central	11.67	0	678.00	61.21	38.79
Longitudinal	20.88	0	617.00	55.27	44.73
Total medial	100.00	0	529.00	45.94	54.06
Anterior	0	4.33	719.00	65.65	34.35
Ant-extended	0	7.35	710.00	66.20	33.80
Posterior	0	4.83	717.00	65.41	34.59
Post-extended	0	9.16	707.00	66.76	33.24
Total lateral	0	100.00	640.00	73.75	26.25

"bow-legged" alignment. Observing the varus knee moment during single-leg stance, Table 2, the results agreed with previous published studies that have shown the link between frontal alignment and the magnitude of the varus moment and force distribution at the knee.[54,56,64]

Medial Meniscectomy

The location of the maximum normal stress, shear stress and strain occurred on the cartilage-to-cartilage contact on the medial femoral cartilage for all types of medial meniscectomy, Fig. 4. The stress distribution showed minimal change on the lateral compartment after different amounts of medial meniscectomy, Fig. 4. The total contact area between the femoral condyles and the tibial plateau (superior meniscus surface and tibial cartilage) decreased with increased removal of the meniscus (lateral and medial) for both subjects, Table 3. The magnitude of the (a) maximum normal stress, (b) shear stress and (c) normal strain on the medial femoral cartilage increased with greater removal of the medial meniscus for both Subject 1 (varus) and Subject 2 (valgus), Fig. 5. The magnitude of the maximum normal stress on the medial femoral cartilage was always greater for Subject 1 (varus)

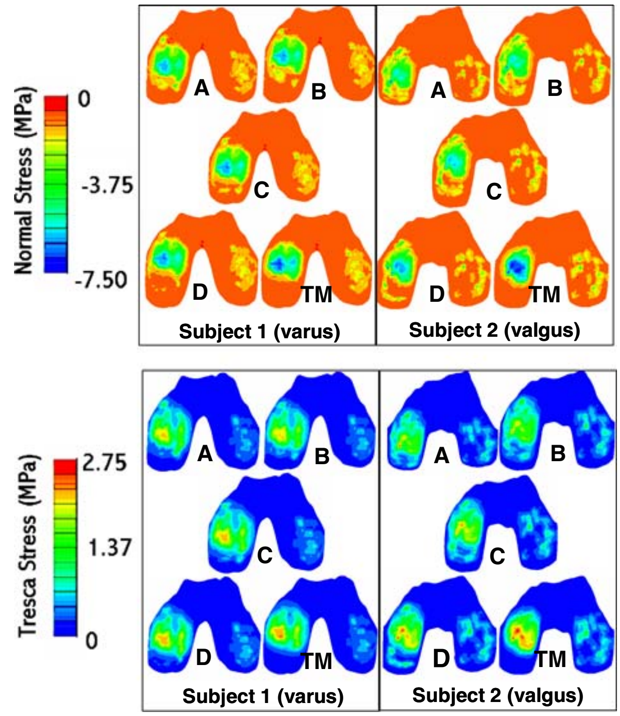

FIGURE 4. Normal stress and Tresca stress distribution on the femoral cartilage of the left knee for Subject 1 (varus alignment) and Subject 2 (valgus alignment) for different amounts of medial meniscectomy. The letters correspond to Fig. 3, where (A) is the healthy joint, (B) is the posterior resection, (C) is the posterior-central resection, (D) is the extended posterior-central resection and (TM) is the total medial meniscectomy.

compared to Subject 2 (valgus), Fig. 5a. With a larger area of resection, the maximum shear stress (Fig. 5b) and normal strain (Fig. 5c) became larger for Subject 2 (valgus). The percentage increase in the normal stress, shear stress, and normal strain also became larger for Subject 2 after greater removal of the meniscus, Table 4.

Lateral Meniscectomy

After different amounts of lateral meniscectomy, inspection of the normal stress and Tresca shear stress distribution showed little change on the medial cartilage, Fig. 6. However, the stress distribution showed the maximum stress on the lateral compartment occurred on the lateral femoral cartilage at the cartilage-to-cartilage contact. The magnitude of the maximum compressive stress, shear stress and normal strain on the lateral cartilage was always greater for Subject 2 (valgus) compared to Subject 1 (varus) on the lateral femoral cartilage after various types of lateral meniscectomy, Fig. 7. The percentage increase in the compressive stress, shear stress, and normal strain are shown in Table 6. The larger values of the contact

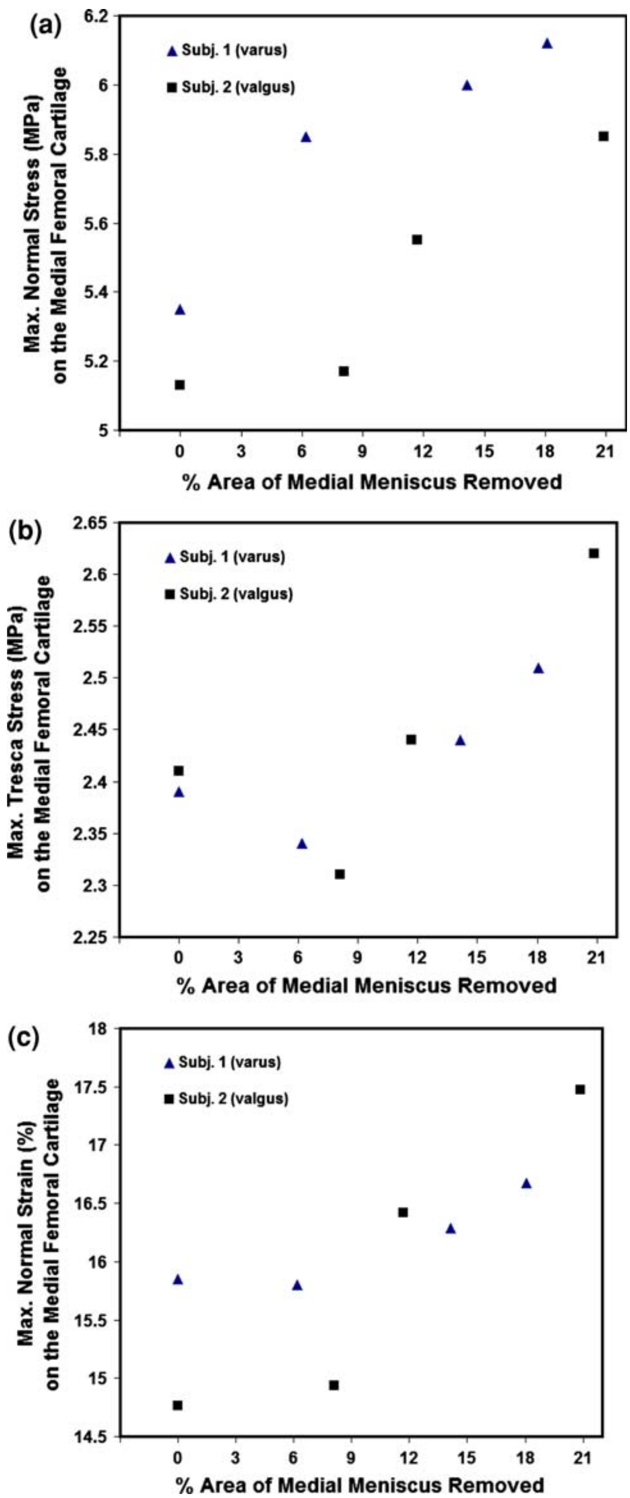

FIGURE 5. The magnitude of the (a) maximum normal stress, (b) Tresca shear strain, and (c) normal strain in the medial femoral cartilage with area of medial meniscus removed from medial meniscectomy for Subject 1 (varus) and Subject 2 (valgus).

variables are attributed to the greater force distribution on the lateral side due to the smaller varus moment for Subject 2.

TABLE 4. The percentage increase in the compressive stress, shear stress, and normal strain on the medial femoral condyle for different types of medial meniscectomy for Subject 1 (varus alignment) and Subject 2 (valgus alignment).

Type	Max. compressive stress	Max. shear stress	Max. normal strain
Subject 1			
Posterior	9.35	−2.09	−0.32
Posterior-central	12.15	2.09	2.78
Longitudinal	14.39	5.02	5.17
Total medial	22.99	15.06	14.32
Subject 2			
Posterior	0.78	−4.15	1.22
Posterior-central	8.19	1.24	11.25
Longitudinal	14.04	8.71	18.36
Total medial	30.99	25.31	35.77

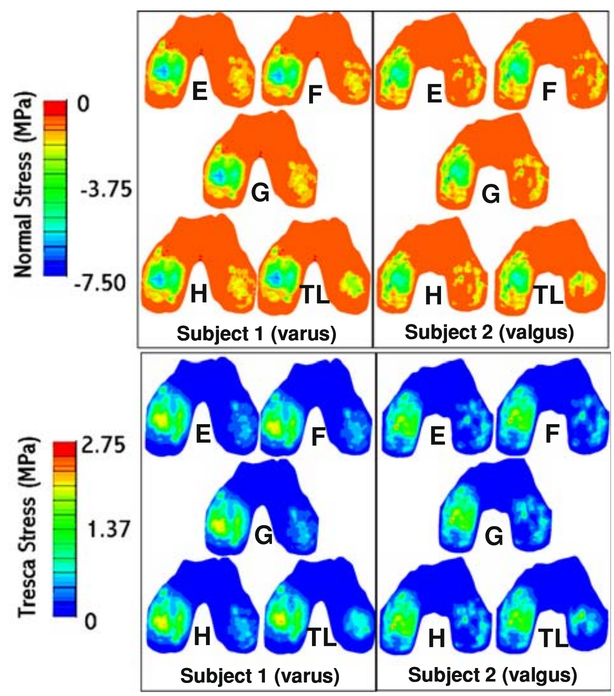

FIGURE 6. Normal stress and Tresca stress distribution on the femoral cartilage of the left knee for Subject 1 (varus alignment) and Subject 2 (valgus alignment) for different amounts of lateral meniscectomy. The letters correspond to Fig. 3, where (E) is the anterior resection, (F) is the extended-anterior resection, (G) is the posterior resection, (H) is the extended-posterior resection, and (TL) is the total lateral meniscectomy.

The results showed the lateral meniscus carried a majority of the load in the anterior horn, similar to other studies.[51,69] This caused larger percentage increases in the contact stresses after an anterior meniscectomy to the lateral meniscus for both subjects, Table 6. Similar results have been found in an FEA study of lateral meniscectomy by Peña *et al.*[51]

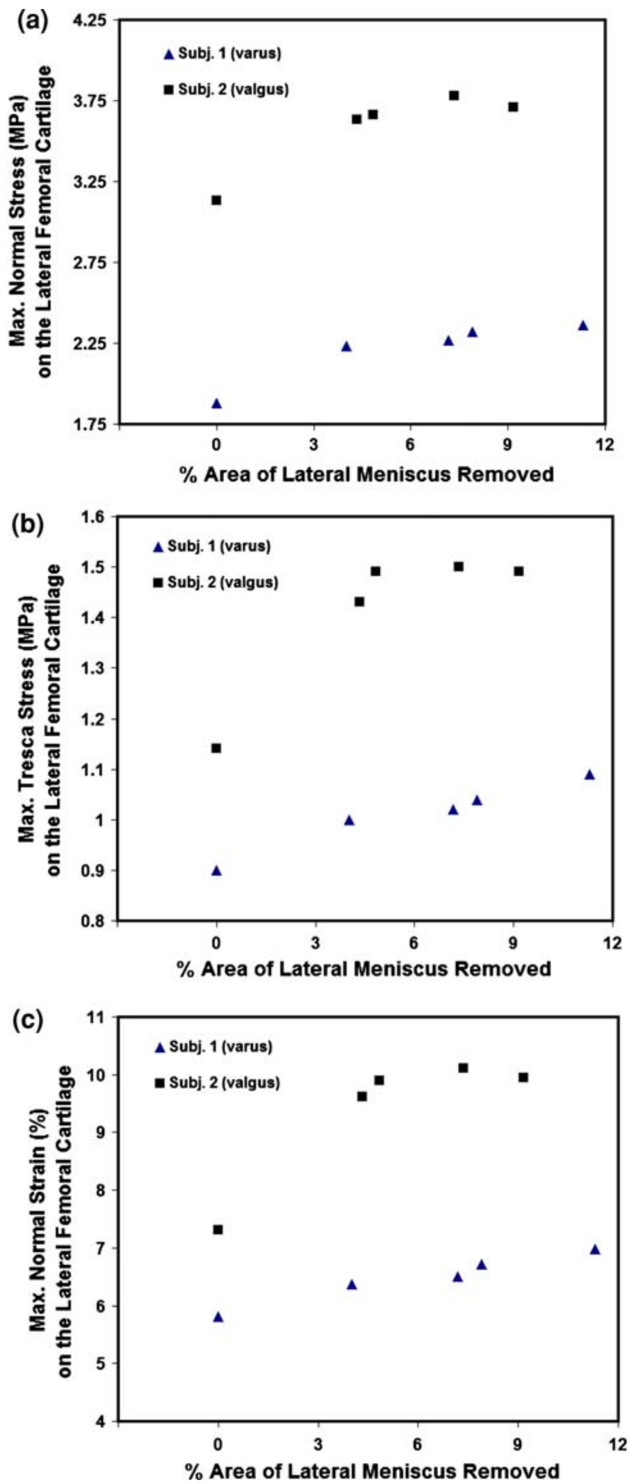

FIGURE 7. The magnitude of the (a) maximum compressive stress, (b) shear strain, and (c) normal strain in the lateral femoral cartilage with area of lateral meniscus removed from lateral meniscectomy for Subject 1 (varus) and Subject 2 (valgus).

However, only axial loading was applied to their model and they did not account for the varus moment that occurs during single-leg support.

DISCUSSION

Subject 1 (varus) had a greater load distributed to the medial knee compartment compared to Subject 2 (valgus) due to the greater varus tibiofemoral alignment, and thus, a larger varus knee moment. As mentioned previously, during single-leg support, approximately 70–75% of the load passes to the medial compartment of the knee joint due to the varus moment.[3,5,30] Each subject exhibited greater than 75% of the load to the medial compartment of the knee which could be attributed to the LCL acting as the only structure to oppose the external varus moment. Including the muscle forces in the frontal plane may decrease the distribution of the total knee force to the medial knee compartment. These results are different from those of Haut Donahue *et al.*,[29] who calculated an even force distribution between the medial and lateral knee compartments when applying an axial compressive load of 800 N to a 3D finite element knee model. This comparison shows the importance of applying the varus knee moment in finite element knee models.

The application of the varus knee moments also led to each subject demonstrating a larger magnitude of stress and strain on the medial cartilage compared to the lateral cartilage. Peña *et al.*[49] applied only an axial compressive load of 1150 N and an anterior tibial load of 134 N to a 3D fully-intact finite element knee model and found maximum normal stress of 3.11 MPa on the lateral femoral cartilage and 2.68 MPa on the medial cartilage. In the current study, for a healthy fully-intact knee joint, with the application of the varus knee moment, Subject 1 (varus) had maximum normal stress of 5.35 and 2.00 MPa on the medial and lateral femoral cartilage, respectively, and Subject 2 (valgus) had maximum normal stress of 5.13 and 3.13 MPa on the medial and lateral femoral cartilage, respectively, Figs. 5a and 7a. The magnitude of the normal stresses on the medial knee cartilage was approximately double with the application of the varus knee moment in the current model compared to previous models. This illustrates the importance of including the varus knee moment, when studying knee biomechanics.

For both medial and lateral meniscectomies, the overall magnitude of the stress and strain in the femoral cartilage increased with increased removal of the meniscus due to a decrease in the contact area at the knee joint. It was expected that for all types of medial meniscectomy, Subject 1 (varus) would have a greater increase in the stresses and strains due to the greater magnitude of the varus moment and force distribution to the medial compartment. This is not the case as Table 5 shows the local distribution of the force on the medial compartment between the articular cartilage

TABLE 5. Local force distribution between the meniscus and cartilage on the medial and lateral compartment for different types of meniscectomies for Subject 1 (varus alignment) and Subject 2 (valgus alignment).

			Local force distribution (%)			
	% Area removed		Medial		Lateral	
Type	Medial	Lateral	Cartilage	Meniscus	Cartilage	Meniscus
Subject 1						
Healthy joint	0	0	86.60	13.40	48.64	51.36
Posterior	6.19	0	87.70	12.30	48.85	51.15
Posterior-central	14.15	0	90.57	9.43	49.06	50.94
Longitudinal	18.07	0	92.45	7.55	49.08	50.92
Total medial	100	0	100.00	0.00	48.94	51.06
Anterior	0	4.01	86.97	13.03	51.16	48.84
Ant-extended	0	7.91	86.98	13.02	54.69	45.31
Posterior	0	7.18	86.54	13.46	52.53	47.47
Post-extended	0	11.31	86.55	13.45	56.30	43.70
Total lateral	0	100	87.52	12.48	100.00	0.00
Subject 2						
Healthy joint	0.00	0.00	62.91	37.09	19.00	81.00
Posterior	8.09	0.00	64.06	35.94	19.50	80.50
Posterior-central	11.67	0.00	71.88	28.13	20.55	79.45
Longitudinal	20.88	0.00	79.53	20.47	23.25	76.75
Total medial	100	0.00	100.00	0.00	26.65	73.35
Anterior	0.00	4.33	62.85	37.15	22.46	77.54
Ant-extended	0.00	7.35	62.72	37.28	26.94	73.06
Posterior	0.00	4.83	62.84	37.16	21.07	78.93
Post-extended	0.00	9.16	62.81	37.19	22.74	77.26
Total lateral	0.00	100	61.71	38.29	100.00	0.00

and the meniscus for Subject 1 (varus) in a fully intact and healthy joint is 87% in the cartilage and 13% in the meniscus. Where as in Subject 2, the cartilage carried 63% of the load and the meniscus carried 37%. Walker and Erkman[63] showed that in a study of fourteen knees on average the load at the knee is distributed approximately evenly between the cartilage and the meniscus on the medial compartment and on the lateral compartment the majority of the load is carried on the meniscus. For Subject 1 in the medial compartment, the medial meniscus carries much less force than the medial cartilage. This could be due to abnormalities in development or due to the alignment. The medial meniscus in Subject 2 carried more load in a healthy joint and is the reason large medial meniscectomy is worse for in Subject 2 compared to Subject 1.

For total medial meniscectomy, Subject 1 (varus) had an increase on the medial cartilage in the overall maximum compressive stress, shear stress and normal strain of 23, 15, and 14%, respectively. Subject 2 (valgus) had increases of 31, 25, and 36%, respectively, Table 4. The larger increases can be explained again by the greater role the medial meniscus played in Subject 2 compared to Subject 1 in carrying the load distributed to the medial compartment of the knee. Because of the larger load distribution to the lateral compartment, it was expected that for all types of lateral meniscectomy, Subject 2 (valgus) would have a greater increase in the stresses and strains after meniscectomy compared to Subject 1 (varus). Unlike for medial meniscectomy, this hypothesis held true.

TABLE 6. The percentage increase in the compressive stress, shear stress, and normal strain on the lateral femoral condyle after lateral meniscectomies for Subject 1 (varus alignment) and Subject 2 (valgus alignment).

Type	Max. compressive stress	Max. shear stress	Max. normal strain
Subject 1			
Anterior	18.62	11.11	9.64
Ant-extended	23.40	15.56	15.49
Posterior	20.74	13.33	12.05
Post-extended	25.53	21.11	20.14
Total lateral	72.87	68.89	76.42
Subject 2			
Anterior	15.97	25.44	31.46
Ant-extended	20.77	31.58	38.30
Posterior	16.93	30.70	35.29
Post-extended	18.53	30.70	35.98
Total lateral	61.02	96.49	107.52

For total lateral meniscectomy, Subject 1 (varus) had an increase on the lateral cartilage in the overall maximum compressive stress, shear stress and normal strain of 73, 69, and 76%, respectively. Subject 2 (valgus) had increases of 61, 96, and 108%, respectively, Table 6. The larger percentage increase for

Subject 2 can be attributed to the greater load bearing role of the lateral meniscus in Subject 2 compared to Subject 1. The local force distribution on the lateral compartment in Subject 1 is 49% in the cartilage and 51% in the meniscus. The force distribution on the lateral compartment for Subject 2 is 19% in the cartilage and 81% in the meniscus, Table 5.

Peña *et al.*[52] showed greater percentage changes in the normal stress in the femoral cartilage after different amounts of total and partial lateral meniscectomy, i.e., 456% increase in maximum normal stress and 687% increase in maximum shear stress after total lateral meniscectomy. However, the meniscus was modeled as isotropic elastic compared to transversely isotropic in the current model which exhibited greater stiffness. Furthermore, the 3D knee model was developed from a volunteer and subject specific loading conditions were not accounted for which include the varus knee moment. The current model accounts for subject specific data with a more realistic material model of the meniscus.

Previous published literature has reported greater degenerative changes after lateral meniscectomy compared to medial meniscectomy.[2,33,34] This could be explained by the greater relative role the lateral meniscus plays in the local force distribution compared to the medial meniscus. The finite element results showed greater overall magnitude of the contact stresses on the medial side, but the overall percentage increase after lateral meniscectomy was much greater, again due to the increased role of local force distribution in the lateral meniscus. The results support the hypothesis of Peña *et al.*[51] which states that the percentage increase in stress is more detrimental than the maximum stress criterion when considering OA progression. Although the medial compartment is exposed to greater load, the medial cartilage carries a majority of the local load and medial meniscectomy will not cause as great a change in the contact stress and strain as a lateral meniscectomy would cause to the lateral cartilage. This agrees with previous research that observed lateral meniscectomy was more detrimental in terms of articular cartilage degradation compared to medial meniscectomy.[2,33,34]

Limitations that exist in this model were that elastic conditions were used to define the material properties of the articular cartilage and the meniscus. This assumption is valid for the short loading duration in this investigation. Future investigations will use poroelastic conditions to observe the time-dependent properties of the soft tissue at the knee joint.

The loading conditions represent a minimum of the total forces that the knee is exposed to during static stance. The co-contraction of the muscles at the knee adds a great deal of force to the knee during dynamic activities. Furthermore, the muscle forces did not add any additional support to oppose the external varus moment. The LCL was the only structure which opposed the external varus moment and including the muscles may decrease the value of the stresses and strain as well as the percentage to the total knee force distributed to the medial knee compartment. Complex modeling of the ligaments such as the 3D hyperelastic modeling similar to Peña *et al.*,[49–52] may provide more resistance to the posterior directed shear force and reduce the overall shear stress in the model.

The most significant limitation to the current study is the small sample size. The construction of the 3D finite element model of the knee is a time consuming process due to computational processing and is a reason only two subjects were used in this comparison study. The small subject population makes it impossible to generalize the results. However, the results show that subject specific knee joint geometry and loading conditions are very important when analyzing the contact behavior of the knee. It is highly recommended that a subject specific approach be taken when using finite element models to determine the effect of meniscectomy on the stresses and strains at the knee which may accelerate degenerative changes within the knee.

This subject specific finite element study is the only known investigation that analyzes the combined effect of the varus moment that occurs during single-leg support and meniscectomy on the contact stresses and strains at the knee cartilage. The varus moment is a key factor in the force distribution at the knee and gives a more physiologically accurate loading condition at the knee than an axial load alone.

Two subjects, one with varus frontal plane tibiofemoral alignment and one with valgus alignment, were used to create 3D finite element knee models to observe the effects the tibiofemoral alignment and partial and total meniscectomy on the contact stresses and strains at the knee cartilage. Subject specific loading conditions and knee joint geometry were used to simulate single-leg support in finite element models. The results showed the fontal plane alignment and local force distribution between the articular cartilage and meniscus must be accounted for when analyzing the effect of meniscus removal on the stresses and strains in the knee cartilage. Furthermore, the lateral articular cartilage was exposed to higher percentage increase of stress and strain after different amounts of meniscectomy. Specifically, the results showed increase varus knee moment at single-leg stance led to greater stress and strain magnitude at the medial articular cartilage. However, the percentage change in the stress and strain was correlated with the local force distribution between the articular cartilage and the meniscus at the respective knee compartment. The results indicate that an individual's frontal plane knee alignment and unique local force distribution between the articular cartilage and meniscus play an important role in the biomechanical effects of total and partial meniscectomy. This model could be possibly utilized to predict the potential for onset of knee OA changes as well as provide information to delay the onset or slow the progression of OA.

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