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Displacements of the Tibial Tuberosity

Effects of the Surgical Parameters

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A three-dimensional computer model is used, based on the finite element method, to investigate the effects of 1-, 1.5-, and 2-cm tibial tubercle elevations and of 0.5- and 1-cm medial displacements of the tuberosity, performed with different bone shingles. Patellar kinematics and patellofemoral interface peak pressure, between 45° and 135° of passive knee flexion, are compared for these different surgical parameters with those of a normal knee not surgically treated. The shingle lengths of 3, 5, 7, and 10 cm have little influence on the results. Augmenting tubercle medializations decrease the lateral peak pressure but result in an overpressure of the medial facet that is 154% of the normal peak value. With knee flexion between 45° and 60°, increasing tubercle elevations decreases lateral and medial peak pressures. With flexion of more than 60°, increasing elevations decrease the lateral peak pressure, but they augment and even cause overpressure on the medial facet. An overpressure on the lateral facet also is seen in midrange knee flexion (75°–90°) for all tubercle elevation values. Increasing tubercle elevations and medializations appear to be the predominant parameters from a biomechanical point of view.

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Common patellar pathologic conditions, such as pain, abnormal tracking, chondromalacia, and arthrosis,[16] are thought to be related to malalignment factors,[16] patellar instabilities,[34] abnormal contact surfaces, or abnormal pressures.[5,6,30] In addition, these different factors could be correlated among themselves.[1,16,34] Previous studies clearly have shown the influence of the realignment of the extensor apparatus on patellofemoral contact pressure, experimentally[3,4,12,20,24,27] and theoretically,[7,8,10] and on patellar kinematics.[8,22,33] Recurrent subluxation of the patella also has been found to be associated with high values of the patellofemoral lateral force.[5,14,15,23] Abnormal tracking also can lead to an alteration of the contact surface and of the contact pressure. These mechanical factors are thought to be instrumental in the development of osteoarthrosis[29] and in the degeneration of the patellar cartilage,[5,23] although it is unclear whether it is too much[30] or too little[6] stress that initiates the arthrotic process.

Because a strong coupling[1] exists between these factors, it is difficult clinically and experimentally to assess the primal cause(s) of these pathologic conditions. Two classic procedures involving the extensor apparatus are used to relieve these pathologic conditions: a tibial tubercle elevation to decrease patellofemoral contact pressure[18] and a medial displacement of the tuberosity to achieve patellar stability.[32] These procedures (Fig 1) often give inconsistent results, and all of their respective consequences have not been investigated.[1] In particular, the amount of tubercle displacement,[3,16,17,18,24,25,28] the osteotomy plane, and the length of the bone shingle[16,17,24] are matters of controversy in the literature. If clinical studies give the rate of success of the procedures, it often is with different criteria of success.[25,28] Experimental work gives mainly the pressure distribution[3,4,12,27] or the bone kinematics,[9,33] but rarely both. This lack of thorough and simultaneous investigation, which is difficult to do experimentally, justifies the development of a mathematical model, as patellar kinematics, contact surfaces, and pressure distribution are strongly correlated.[1]

The goal of this study is to obtain simultaneously the patellar kinematics and the contact pressure distribution at different steps of passive knee flexion. The effects of two surgical procedures on patellar pressure and kinematics are investigated for different anterior or medial displacements of the tibial tuberosity, obtained by the rotation of different bone shingles.

MATERIALS AND METHODS

Finite Element Model

A three-dimensional finite element model of the patellofemoral joint[2,8] is used. This model previously has been validated through the simulations of experimental knee kinematics.[9] Meshes of the patella and of the femoral articular surface are generated by using series of 4-mm spaced sagittal slices obtained by computed tomography of a normal knee. The mesh characteristics are summarized in Table 1. Three basic regions are defined in the patella: cortical, cancellous bone, and cartilage, which are assumed to be elastic isotropic and homogenous media (in Fig 2, cortical regions are white and cartilage is light gray). The patellar tendon is modeled by three elastic fibers connecting the patella with the tibial inser-

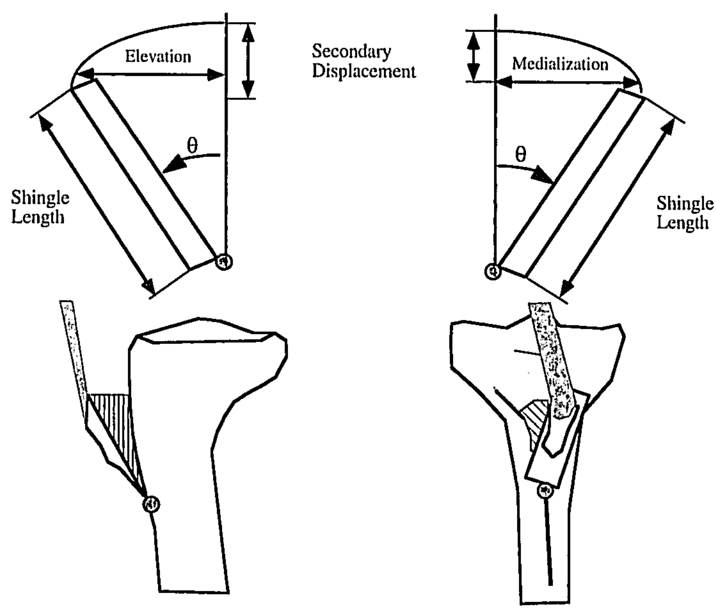

Fig 1. Elevation and medial displacement of the tibial tuberosity. Secondary displacement is induced by the shingle rotation of angle Θ during a tibial tubercle elevation or a medial displacement of the tuberosity.

tions of the patellar tendon. The fiber insertion points on the tuberosity are obtained from preliminary experiments. They correspond to the insertions of three bundles of fibers in the patellar tendon. They are marked with radiopaque markers on the specimens. The patellofemoral interface is modeled by frictionless large slip contact elements.[8]

The chosen geometries were those of a normal knee. The geometric parameters are given in Table 2 and appear to be in the normal range. The anteroposterior thickness of the patellar regions varies: cartilage, 1.7 to 3.3 mm; cortex, 0.7 to 3.4 mm; and cancellous bone, 3.3 to 16.1 mm. A sensitivity analysis of the model also has been performed.[8] Material constants were taken from the literature[21,26,31] (Table 3).

Passive knee flexion is simulated with a constant 40 N force applied by the quadriceps and the tibia in neutral position. This validated model permits simultaneous insight of patellar kinematics and stress distribution.[8]

Procedure Simulation

Three tibial tubercle elevations of 1, 1.5, and 2 cm[3,4,18] are simulated using various shingle lengths (5, 7, and 10 cm). A certain consensus gives 1 cm as the upper limit for medial displacement of the tuberosity.[5,16] Consequently, two medial displacements of the tuberosity of 0.5 and 1 cm also are simulated (for 3, 5, and 7 cm shingle lengths). The shingle lengths have been chosen[5,16] to permit comparison with recent experimental work.[24] No combination of medial and anterior displacement has been investigated, as was done in some recent work.[20]

TABLE 1. Characteristics of the Finite Element Mesh

Mesh Description	Knee Simulated
Number of nodes	1445
Flexion increment (°)	7.5
Cortex (8 node isoparametric elements)	280
Cancellous bone (8 node isoparametric elements)	320
Cartilage (8 node isoparametric elements)	180
Femur (16 node Hermite facets)	148
Tendon (2 node line elements)	3
Contact (node-facet large slip elements)	99

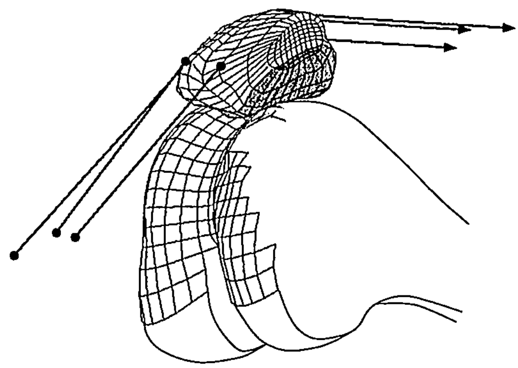

Fig 2. Finite element model. The black arrows represent the orientation of the quadriceps extension force. Distal black lines represent the three fibers that model the patellar tendon.

The bone cut was made along the anterior tibial crest and parallel to the tibial coronal plane. Tubercle elevation or medial displacement are simulated by rotating a bone shingle around a mediolateral axis (in a tubercle elevation procedure) and around an anteroposterior axis (in a medialization procedure), respectively. All of these axes pass through the distal end of the bone shingles. With rotation of the shingle, the tibial attachment of the patellar tendon is moved in the requested direction (either anteriorly or medially) and more distally with shorter shingles (Fig 1). The tibial trajectory is not modified significantly by these procedures because it has been confirmed previously.[33]

Patellar kinematics and contact peak pressures are investigated between 45° and 135° of passive knee flexion. Kinematics are expressed by three rotations and translations of a moving coordinate system attached to the patella relative to a fixed coordinate frame attached to the femur (Fig 3). Rotations are accomplished first around the x_f axis (patellar flexion) then around the y_f axis (patellar tilt) and finally around the z_f axis (patellar rotation). The peak values of the contact pressure on the lateral and medial facets of the patellar cartilage also are considered. Hereafter, the values of the variables for a knee not surgically treated will be called normal. Patellar peak pressures and kinematics are compared for different shingle lengths, varying tubercle elevations, and medial displacements. Although this study is

TABLE 2. Characteristics of the Knee Geometry

Knee Characteristic	Knee Simulated	Normal Knee	Chondromalacia/ Recurrent Subluxation
Gender	Female		
Age (years)	63		
Wiberg type	II	65%	
Odd facet	Yes	Yes	
Valgus angle (°)	5	5–10	
Q angle (°)	14	15	20/15
Sulcus angle (°)	143	137 ± 6	139/147
Ratio patellar tendon length: patellar length	1.10	1.06 ± 0.12	1.08/1.23
Patellar cartilage thickness (mm)	1.7–3.3	≤ 4–5	
Patellar cortex thickness (mm)	0.7–3.4		
Femoral (AP) depth (mm)	70	69.9 ± 2.6	
Femoral (LM) width (mm)	68	69.7 ± 2.7	
Patellar (AP) depth (mm)	25	25	
Patellar (LM) width (mm)	40	40.2 ± 3.1	
Patellar (PD) height (mm)	34	29.9 ± 1.3	

AP = anteroposterior; LM = lateromedial; PD = proximodistal. The normal and abnormal values were taken from the literature.[5,16,19]

not a statistical situation of repeated measurements, the pressure output of the model can be modeled to some extent as a black box model, in which some parameters can be varied. In this context, a multivariate analysis of variance (MANOVA) of the data was done, taking into account the dependence between lateral and medial peak pressures. The relative weights of the tubercle displacement, the shingle length, and the angle of knee flexion on the peak pressures are investigated with F tests at a 5% significant level.

RESULTS

Effects of Medial Displacement of the Tuberosity

With knee flexion, normal patellar shift monotonously decreases from a +9-mm medial position to a –2-mm lateral position of the patella relative to the femur origin. After medialization of the tuberosity, the patellar shift is slightly more medial (positive x_f translation; mean difference at 105°, 0.5 mm) than the shift of the patella not surgically treated. The patella also reaches more proximal (positive y_f translation; mean difference at 105°, 0.8 mm) and more anterior (positive z_f translation; mean difference at 135°, 2.4 mm) than do the normal values. The rotation of the patella is systematically more external than is the normal rotation over all of the flexion range (negative values in Fig 4; mean difference at 105°, –2.6°). The mean differences are maximum at the chosen flexion angles given between brackets.

Normal peak pressures vary with knee flexion (Table 4). Results have been normalized to permit percentage comparisons with the normal values. Relative peak pressures for the different medializations, shingle lengths, and angles of knee flexion are provided in Table 5 on the lateral and the medial facet. A 0.5-cm medial displacement causes a global decrease of the relative lateral peak pressure over all of the flexion range (mean, 0.93) and an augmentation on the medial peak pressure (mean, 1.19). This tendency is enhanced by a more pronounced 1-cm medial displacement (lateral facet mean, 0.85; medial facet mean, 1.35). Respective mean values for increasing medializations are shown in Figure 5 for the

TABLE 3. Material Constants

Model Region	λ (MPa)	μ (MPa)	E (MPa)	ν
Cortical bone	8.65×10^3	5.76×10^3	1.5×10^4	0.30
Cancellous bone	1.73×10^2	1.15×10^4	3.0×10^2	0.30
Cartilage	1.065×10^1	6.8×10^{-3}	2.0	0.47
Patellar tendon		5.0×10^1	10^2	

The values were taken from the literature.[21,26,31] λ and μ are the Lamé coefficients, E is the Young modulus, and ν is the Poisson ratio.

medial (solid line) and for the lateral facet (dashed line). The mean peak pressures clearly increase on the medial facet and decrease on the lateral facet for increasing medializations. The MANOVA has shown that the amount of medialization and the knee flexion angle are significant parameters of the model peak pressures on the lateral and medial facet of the patella.

Effects of Tibial Tubercle Elevation

With increasing knee flexion, the patella reaches more lateral (mean difference at 105°, –0.7 mm), more distal (mean difference at 90°, –1 mm), and more posterior (mean difference at 135°, –4 mm) after a tubercle elevation procedure. The patellar rotation also is more external (mean difference at 105°, –4.8°) than are the normal values. The mean difference is maximum at the chosen flexion angles given between brackets.

Relative peak pressures for the different elevations, shingle lengths, and angles of knee flexion are shown in Table 6.

Lateral Facet

On the lateral facet, a 1-cm elevation of the tuberosity causes a small decrease of the peak pressure over all the flexion range (mean, 0.94). For a 1.5-cm elevation, a decrease is noticeable (mean, 0.92), but the difference between the 1.5-cm and the 1-cm values is small (mean, –0.03). For a 2-cm elevation, the peak pressure decrease is more pronounced (mean, 0.83). Mean relative values for the different elevations are shown in Figure 6. This lateral decrease particularly is observed between 45° and 60° knee flexion for the three elevations and can reach 45% of the normal value for a 2-cm elevation.

A general surpassing of the normal lateral peak pressure, to 133% of its normal value, is seen between 75° and 90° for all tubercle elevations. It also is found that even if the peak pressures at 135° are below the peak pressures of the knee not surgically treated, an increase of the peak pressure occurs between the 1-cm elevation and the 1.5-cm and 2-cm tubercle elevations.

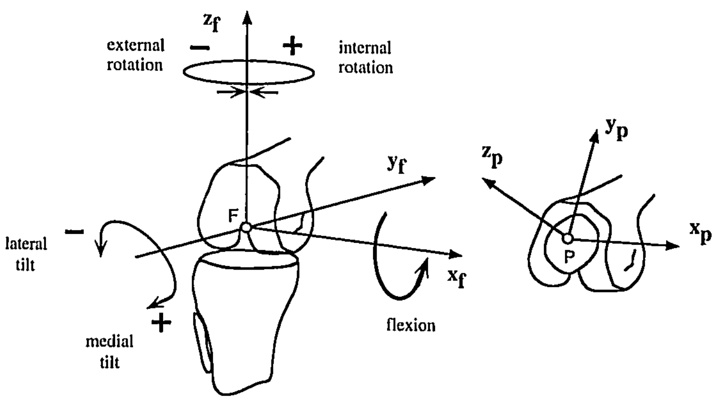

Fig 3. Definitions of reference frames and of patellar flexion, tilt, and rotation. F = the femoral intercondylar notch; P = patellar centroid. $Fx_fy_fz_f$ and $Px_py_pz_p$ are the femoral and patellar reference frames, respectively.

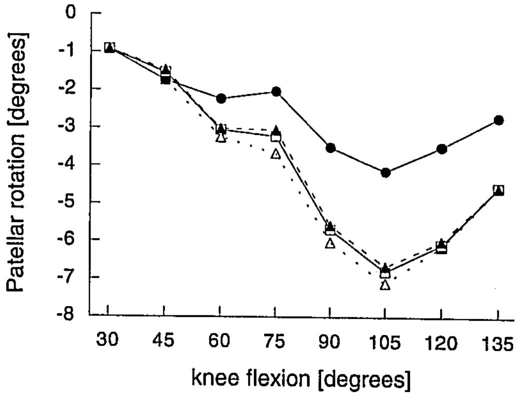

Fig 4. Patellar rotation angle for the 3- (open triangle, dotted line), 5- (square, solid line) and 7-cm (filled triangle, dashed line) shingles and the values from the knee not surgically treated (filled circle) for the entire knee flexion range for a 1-cm medial displacement of the tuberosity.

Medial Facet

Over the entire flexion range considered, the medial facet peak pressures are smaller for a 1-cm elevation than for the 1.5-cm procedure, in opposition to the lateral facet (1 cm mean, 0.82; 1.5 cm mean, 0.88). A 2-cm elevation brings no clear pressure variation over the entire flexion range (mean, 1.04). Respective values are shown in Figure 6. On the contrary, between 45° and 60° knee flexion, increasing the tubercle elevation decreases more clearly the medial peak pressures: the mean difference between the 2-cm value and the 1.5-cm and 1-cm values are –0.12 and –0.18, respectively. However, from 75° to 135° flexion, these medial peak pressures systematically are smaller for lower tubercle elevations. There is also a surpassing of the normal medial peak pressure, to 138%, which is achieved for a 2-cm elevation for all shingles and flexion angles greater than 75°.

The MANOVA has shown that the amount of elevation and the knee flexion angle are significant parameters of the model peak pressures on the lateral and medial facet of the patella.

Effects of the Shingle Length

Shingle length has almost no influence on contact peak pressure and patellar tracking after both surgical procedures. The MANOVA has detected only a significant factor for the shingle length in the analysis of the medial peak pressure after a medialization procedure. However, this shingle factor is approximately two orders of magnitude smaller than the medialization factor. This can be seen by comparing the mean difference between the medial peak pressures for a 3-cm and a 7-cm shingle (mean, –0.04), with the mean difference between 0.5 and 1 cm medialization peak pressures (mean, –0.16). Shorter shingles cause a small decrease of the medial peak pressure in a medialization procedure.

DISCUSSION

Passive knee flexion has been simulated under a constant 40 N quadriceps force, which is lower than physiologic loads. One could ask if the contact areas could radically change with the application of a more physiologic load that is three times body weight, for example. Forty Newtons is a physiologic load for an isometric contraction of the quadriceps that has been used in recent work on the contact areas after similar surgical procedures.[20] In addition, the effects of quadriceps tension force on the patellar trajectory have been shown to be small,[22,33] especially in the flexion range used in this study. The validation of the model[8] has been

TABLE 4. Normal Peak Pressures at Different Steps of Knee Flexion

Patellar Facet	45°	60°	75°	90°	105°	120°	135°	Mean
Lateral (MPa)	0.71	0.40	0.28	0.37	0.54	0.60	0.67	0.51
Medial (MPa)	0.46	0.48	0.31	0.37	0.27	0.34	0.59	0.40

TABLE 5. Relative Lateral and Medial Peak Pressure After a Medialization of the Tuberosity at Different Steps of Knee Flexion

Medialization	Shingle Length	45°	60°	75°	90°	105°	120°	135°	Mean
Lateral facet									
0.5	3	0.94	0.89	0.92	0.89	0.95	0.97	0.94	0.93
	5	0.93	0.90	0.94	0.91	0.95	0.96	0.94	0.93
	7	0.92	0.90	0.94	0.92	0.95	0.96	0.93	0.93
1.0	3	0.85	0.80	0.88	0.79	0.87	0.92	0.88	0.86
	5	0.83	0.79	0.84	0.82	0.89	0.91	0.87	0.85
	7	0.82	0.79	0.85	0.83	0.89	0.91	0.87	0.85
Medial facet									
0.5	3	1.18	1.17	1.25	1.13	1.13	1.12	1.22	1.17
	5	1.22	1.17	1.28	1.17	1.16	1.13	1.22	1.19
	7	1.23	1.18	1.28	1.18	1.17	1.13	1.22	1.20
1.0	3	1.30	1.32	1.43	1.25	1.24	1.24	1.41	1.31
	5	1.38	1.35	1.52	1.33	1.31	1.24	1.40	1.36
	7	1.39	1.36	1.54	1.36	1.33	1.24	1.39	1.37

done by the comparison of predicted numeric results with equivalent experimental results during passive knee flexion. A 40-N constant force was used in this experimental work.[9] The authors' finite element model permits more important quadriceps loads, but a 40-N force (approximately 5% body weight) was chosen to compare the new results with the available experimental and numerical data[8,9] and with part of the literature that has used approximately 5%,[20,33] 20%,[24] and 150%[27] body weight loadings.[13] However, passive knee flexion has been simulated in the current study, and care should be taken in extrapolating these results to daily activities when other forces and torques are applied.

A sensitivity analysis of the model also has been performed.[8] When the geometries were measured, the reconstruction errors were small (approximately 0.1 mm) when compared with the uncertainty on localization of the patellar tendon and rectus femoris insertion points on the patella (approximately 1–2 mm). Thus, the authors have displaced the insertion points of the patellar tendon fibers and the quadriceps force application points and have checked to see if the model results were modified significantly. For example, a 1- or 2-mm anterior or poste-

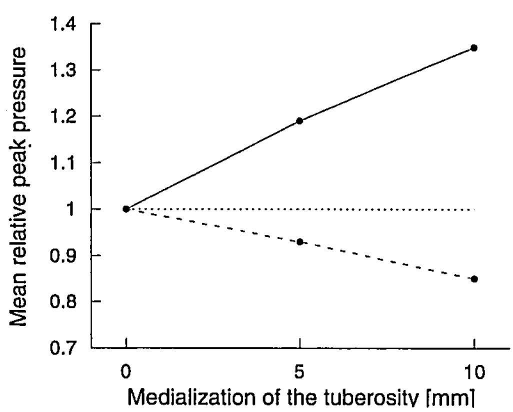

Fig 5. Medial displacement of the tuberosity: average relative peak pressures on the medial (solid line) and lateral (dashed line) facets. Values are in percentage of the normal (not surgically treated) peak pressure.

TABLE 6. Relative Lateral and Medial Peak Pressure After Elevation of the Tuberosity at Different Steps of Knee Flexion

Elevation	Shingle Length	45°	60°	75°	90°	105°	120°	135°	Mean
Lateral facet									
1.0	5	0.75	1.02	1.32	1.28	0.81	0.79	0.65	0.95
	7	0.72	0.97	1.33	1.30	0.83	0.79	0.65	0.94
	10	0.70	0.93	1.33	1.31	0.84	0.79	0.65	0.94
1.5	5	0.60	0.92	1.20	1.20	0.72	0.87	1.01	0.93
	7	0.56	0.84	1.21	1.19	0.72	0.87	1.01	0.92
	10	0.54	0.79	1.22	1.18	0.71	0.86	1.01	0.90
2.0	5	0.45	0.85	1.15	1.08	0.63	0.57	0.97	0.81
	7	0.45	0.86	1.15	1.11	0.63	0.62	0.97	0.83
	10	0.45	0.86	1.16	1.13	0.63	0.65	0.98	0.84
Medial facet									
1.0	5	0.99	0.90	0.71	0.62	0.74	0.83	0.83	0.80
	7	1.01	0.89	0.83	0.66	0.69	0.83	0.83	0.82
	10	1.01	0.88	0.92	0.73	0.65	0.83	0.83	0.84
1.5	5	0.96	0.84	0.97	0.72	0.79	0.92	0.88	0.87
	7	0.96	0.82	0.99	0.73	0.82	0.93	0.88	0.88
	10	0.96	0.81	1.08	0.77	0.83	0.94	0.88	0.89
2.0	5	0.90	0.66	1.01	1.11	1.38	1.08	1.19	1.05
	7	0.89	0.65	0.97	1.12	1.35	1.12	1.19	1.04
	10	0.88	0.65	0.95	1.12	1.33	1.13	1.19	1.04

rior shift has been performed on the point triplet of quadriceps force application. In all cases, between the altered and original tendon insertions used in the sensitivity simulations, the root mean square deviations are less than 0.2° for rotations and less than 0.03 mm for the translations.[8] A sensitivity analysis to material coefficients was done: different Young moduli of 0.2 MPa, 2 MPa (standard value), and 20 MPa were used to simulate the cartilage with Poisson ratios of 0.47 MPa and 0.3 MPa. The patellar kinematics were not significantly different from those computed with the chosen values (Table 3). However, a systematic augmentation of the peak pressures was noticed with increasing cartilage moduli on both facets. This is consistent with the Hertz contact theory. With low cartilage modulus, the peak pressures became more uniform throughout flexion on both facets.

The main argument about the value of the current study lies in a previous experimental work.[27] Diminution of the pressures after an elevation procedure has been observed to be more pronounced for knees with a bipolar patellofemoral arthrosis than for normal knees. The current results are encouraging for additional development of the finite element model. The simulations of pathologic geometries certainly would be rich in results.

Medial Displacement of the Tuberosity

The goal of medial displacement of the tuberosity is to prevent lateral abnormal tracking of the patella, which is common in the presence of an important leg valgus. Slight medialization of patellar tracking is seen in the simulations after medialization of the tuberosity. This effect is small because with flexion of more than 45°, the patella already is inserted firmly in the femoral groove. The patellar kinematics are altered only slightly by medial displacement of the tuberosity. However, these perturbations of the normal kinematics have had considerable influence on patellar peak pressure. There seems to be a compensation between the lat-

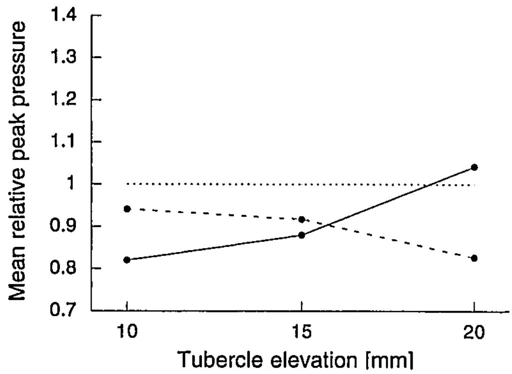

Fig 6. Elevation of the tuberosity: average relative peak pressures on the medial (solid line) and lateral (dashed line) facets. Values are in percentage of the normal (not surgically treated) peak pressure.

eral and medial facet. The peak loads acting on the lateral facet are decreased over the entire flexion range. However, the medial facet is at the same time more heavily stressed, to 154% of the normal peak pressure for a 1-cm displacement. This feature is more pronounced with increasing medial displacement. This relationship could be related to the well known arthrosis of the medial facet caused by excessive medial displacement.[17]

Tibial Tubercle Elevation

The goal of a tubercle elevation procedure is to diminish the patellar contact pressure. In the simulations, increasing tubercle elevation procedures diminish lateral and medial peak pressures between 45° and 60° knee flexion (to 45% of lateral value and to 65% of the medial) and significantly modify the patellar kinematics. This peak pressure decrease is more important for increasing elevations, which is consistent with reports in the literature.[4,27] However, over the entire flexion range, a compensation between lateral and medial peak pressure could occur for the tubercle elevation procedures. Indeed, increasing tubercle elevations decrease the lateral peak pressure and increase the medial peak pressure at high angles of flexion. This procedure does not appear wholly beneficial with regard to peak pressure decrease. It is highly sensitive to the facet considered and to the flexion angle considered. An important increase of the lateral peak pressure was seen between 75° and 90° (to 133% of its normal value). This effect was seen on the proximal zone of the lateral facet in the experimental work of others.[4] It has been concluded that the surgeon should search for the best compromise, depending on the location of the lesions, between the global pressure decrease and this local pressure surpassing of the normal values on the lateral facet.[4] The 1.5-cm elevation could appear as a good compromise between these two conflicting features, showing no increase above the normal medial peak pressure, as occurs with a 2-cm elevation. However, this result cannot be confirmed statistically.

Long term results have been reported in terms of effectiveness of the tubercle elevation procedure.[25,28] A spatial inspection of the degenerative joint is needed to correlate the disrupted regions with abnormal values of the biomechanical variables. During followup of patients, the clinician should check for a systematic arthrosis of the lateral contact region because the model has computed an increase above that of the normal lateral peak pressure between 75° and 90° knee flexion.

Shingle Length

Shingle length (3–10 cm) has little influence on contact peak pressure and patellar tracking after both surgical procedures in the specified range of knee flexion. A previous experimental study[24] on tubercle elevation procedures has shown a statistical pressure decrease for longer shingles (less than 10% at 15° and 30° knee flexion). This result could be observed until midflexion on the lateral facet, where an additional decrease was observed for longer shingles at small flexion angles (mean 5% additional decrease of the peak pressure between 45° and 60°). The authors in that study concluded that this length was of primary importance at less than 30° knee flexion, a result that the current study cannot confirm with flexion of more than 45°. Consequently, other factors such as fracture potential and bone vascularity should be predominant in the right choice of shingle length.

Passive knee motion greater than 45° knee flexion has been simulated for a medial or an anterior displacement of the tuberosity. At less than 45° flexion, knee kinematics could not be simulated because of the lack of soft structures in the model. In a recent experimental study,[9] the soft structures had no influence on patellar kinematics when the knee was flexed more than 45°. However, they have an important stabilizing effect near full extension. In the future, a combination of medial and anterior displacements of the tubercle should be studied and compared with recent experimental data.[12,20] Daily activities should be computed to understand better the knee pathologic conditions, such as in an experimental study simulating level walking and stair climbing with an anatomic specimen knee.[11]

Clinical Relevance

Shingle length has no global influence on contact peak pressure and patellar tracking after both surgical procedures in the specified range of knee flexion. The major parameter appears to be the amount of elevation or medialization of the tubercle. Increasing tubercle elevations decreases lateral and medial peak pressures between 45° and 60° knee flexion. At more than 60° flexion, increasing elevations decrease the lateral peak pressure, but they increase and even surpass the medial normal pressures for a 2-cm elevation. A medial displacement of the tuberosity leads globally to surpassing the normal (not surgically treated) medial peak pressure and to diminishing its lateral counterpart.

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