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REVIEW ARTICLE

Basic kinematics and biomechanics of the patellofemoral joint
Part 2 : The patella in total knee arthroplasty

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Patellar and femoral component in total knee arthroplasty are inextricably linked as a functional unit. The configuration of this unit has been a matter of ongoing debate, and the myriad of different patellar and femoral components currently available reflect the lack of consensus with respect to the ideal design. One of the major challenges is to overcome the biomechanical disadvantages of a small contact area through which high contact pressures are transferred, making this mechanical construct the weakest part of the prosthetic knee. Contact areas are highly dependent on the congruency of the patellofemoral joint articulation, and are significantly smaller for dome shaped patellar components compared to those of more anatomic designs. However, when exposed to 3-dimensional movements, the contact areas of the dome shaped patella are significantly greater, indicating enhanced forgiveness regarding patellar malpositioning. Although contact stresses, a function of implant design and surface conformity, can reach levels far beyond the yield strength of UHMWPE, catastrophic failure of resurfaced patellar components, commonly seen in metal backed patellae, fashionable in the 1980s, has rarely been observed since. Although plastic deformation and wear of UHMWPE continue to represent a problem, in the absence of suitable alternatives polyethylene remains the bearing surface of choice. The appreciation of the consequences of the mechanical environment on the behaviour of the patellofemoral joint is of particular importance in the endeavour to develop knee replacement systems which provide satisfactory function together with clinical long-term success.

Keywords : TKA ; patella ; patellofemoral joint ; kinematics ; biomechanics ; contact area ; UHMWPE.

INTRODUCTION

The articulation between patella and femur is relatively complex and displays intricate biomechanical behaviour. Concerns exist with regard to the treatment of the patella in total knee arthroplasty, as surgically imposed changes through resurfacing may have significant effects on kinematic behaviour and clinical performance of the patello-femoral joint (*28*). Although symptoms originating from the patello-femoral joint, following knee arthroplasty, are rarely severe enough to justify revision, they may be sufficient to spoil an otherwise satisfactory result.

Matthews and associates, who investigated the load bearing characteristics of the patello-femoral joint, remarked that *'High patello-femoral load values, small patello-femoral contact areas, and resultant high stress magnitudes indicate the need for caution in the design and development of a patello-femoral component for total joint replacement prostheses'* (*61*). Analysis of retrieved patellar components and the significant failure rate of metal

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backed patellar designs continue to underscore the extreme mechanical environment in which these implants are expected to perform (*20*). Complications arising from patellar resurfacing are still considerable and include patellar component deformation, wear, fracture and loosening (*49,64,67,76*). It is therefore not surprising that patellar resurfacing remains controversial (*49,52,87,88*).

Patellar tracking, contact area, and pressure distribution differ quite significantly between native and prosthetic knee (*47,100*). Mechanical features to be considered in the creation of a prosthetic patello-femoral joint should include functional range of motion in multi-axial planes, stability, fixation, dimensions, load transfer areas, and materials. A successful patello-femoral articulation must be designed to function under high stress conditions, and over a long period of time, as ground reaction, gravitational, ligamentous, and muscular forces all act to produce significant compressive, shear, and torsional loads (*38,69,92*). Hence both design and materials used must be at least compatible with the mechanical forces of up to 5 × body weight (BW), as encountered during activities of daily living (*87*).

PATELLAR COMPONENT DESIGN

The multitude of patellar components currently available reflects the lack of consensus with respect to the ideal design (Fig. 1). Articular surface geometries of patellar components vary greatly but can be classified into five basic shapes : convex or dome shaped ; modified dome shaped, also known as sombrero hat ; anatomically shaped ; cylindrical or saddle shaped ; mobile bearing (Fig. 2) (*51*). Every implant design bears particular advantages regarding conformity, stability, forgiveness and wear pattern, with none being ultimately superior. Advantages attributed to a particular design should however not be generalized to all designs of similar shape as the behaviour of a particular patellar component is directly dependent on a number of variables with the surface geometry of the mating femoral component probably being the most important (*18,25,63,102,104,116*). Apart from component positioning and alignment other factors such as patient's demographics (e.g. body mass index, mechanical leg-alignment, range of motion) will also influence the performance of the patello-femoral joint (*41*).

Dome shaped patella

The majority of currently available patellar components belongs to the all-polyethylene dome shaped type ; its prevalence today is itself an offshoot of its practicality. The mating geometries between patella and femur are simple spherical shapes, which usually provide congruency only in the early flexion range up to 70° (Fig. 2 & 3). At higher flexion angles the convex patellar surface contacts the convex inner surfaces of the femoral condyles, exposing the patella to high stresses and point contact. Some of these problems have been addressed successfully through design adaptation of femoral condylar and trochlear geometries. Extension of the trochlear groove concavity onto the inner portion of the femoral condyles has provided for an increase of patello-femoral congruency in flexion. The principal advantage of dome shaped components compared to all other designs is their ability to allow for flexion to occur in various planes, hence avoiding edge loading, a problem associated with modified-dome and anatomic patellar devices (Fig. 4). Because of the spherical shape, rotatory alignment of the implant is less critical, highlighting the relative forgiveness of dome shaped patellae regarding minor degrees of malpositioning and making it easy to implant. Despite their excellent clinical results however, failure of cemented all-polyethylene dome shaped patellar components is not uncommon and is attributed to the exposure to high contact stresses (*4,24,32,33,45,89, 113*).

Modified dome shaped patella or sombrero hat

In an attempt to increase the contact area in flexion, the standard dome patella was modified to include a concave surface near the circumference, allowing it to more closely match the curve of the femoral condyles in the axial plane (*9*). The modified dome shaped patella, also known as 'sombrero hat', improves the articulation with the convexities

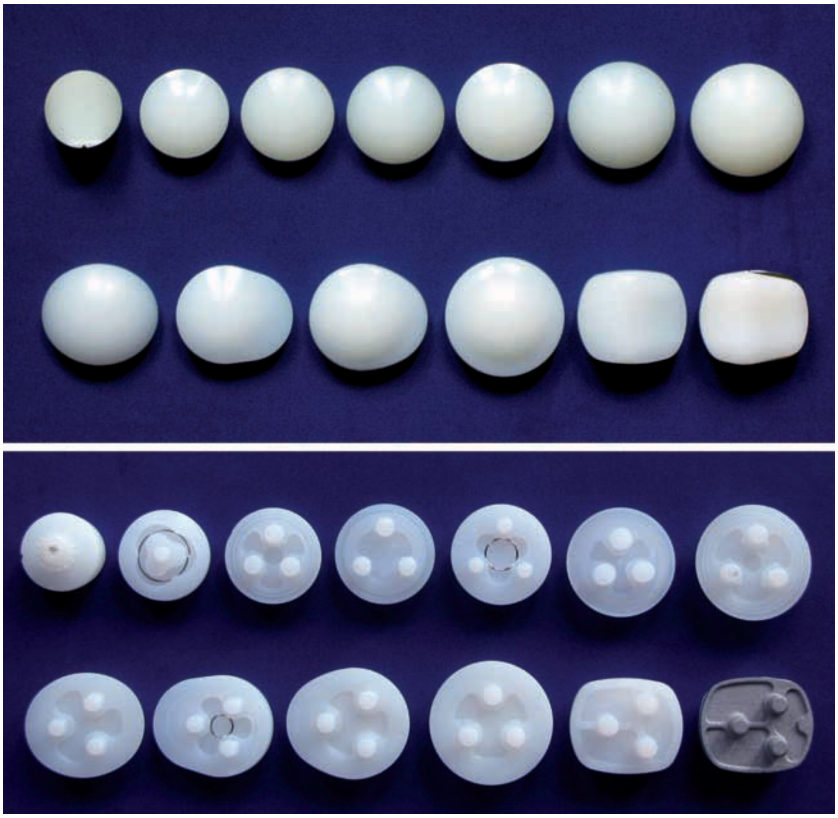

Fig. 1. — Commercially available patellar components. Articulating surface (top), retro-patellar surface (bottom). Top row from left to right : MEDIAL ROTATION KNEE®, Finsbury ; GENESIS® II (biconvex), Smith&Nephew ; VANGUARD®, Biomet ; OPTETRAK®, Exactech ; GENESIS® II dome, Smith&Nephew ; ADVANCE® Medial-Pivot, Wright Inc. ; AGC®, Biomet Inc. Bottom row : PFC-Sigma®, DePuy ; JOURNEY® (off-set dome) Smith&Nephew ; TRIATHLON® (off-set dome), Stryker Inc. ; TRIATHLON® (sombrero), Stryker ; LCS® (all poly), DePuy ; LCS® (rotating platform), DePuy.

of the femoral condyles especially at higher flexion angles (Fig. 2). Wear simulator studies confirmed that the increased conformity enhances the life of the component by more than 20 times when compared to a standard dome component (*44*). Concerns however remain, since the amount of conformity that is acceptable must be considered in relation to patellar motion.

Anatomically shaped patella

Prostheses with anatomical surface profile (Fig. 2) have distinct lateral and medial facets. They provide a more conforming articulation with an increased contact area and reduced contact stresses between patellar and femoral component, thereby decreasing the risk of subluxation (*10,12,46*). A variety of anatomically shaped patellar implants have been available over the years, including a mobile bearing variant. Although anatomic patellar implants make the most sense theoretically, they have introduced a number of complexities into the instrumentation and surgical technique. Due to their high level congruency with the femoral component they are more sensitive to mal-positioning and hence more difficult to implant (*56*).

Cylindrical or saddle shaped patella

The cylindrical or saddle shaped patellar component (Fig. 1 & 2) occupies a fringe position in total knee arthroplasty. The initial idea was developed by

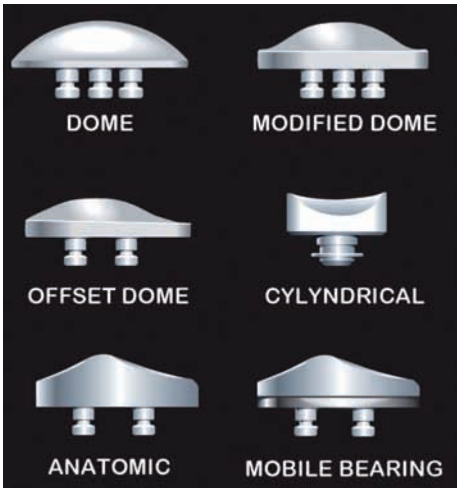

Fig. 2. — Common types of patellar component surface configuration (Copyrights of illustration remain with author).

Freeman and Swanson in the late 1970s, who attempted to combine a high level of congruency with relatively large contact areas throughout flexion of up to 110° (*34,35*). Due to design specifics, the patella becomes highly dependent on a close matching geometry of the femoral component in the sagittal plane, requiring a femoral trochlea with a single radius (*35,53*). The diameter of the patellar component is reduced to 25-30 mm, allowing the implant to be recessed into the patella, similar to an inlay technique. Subsequently the remaining patellar rim participates in articulating with the femoral component and in further assisting stress dissipation. The patellar implant possesses a central peg with a collar and can be used with or without cement. If left uncemented the implant retains the ability to self-centre and to rotate as has been observed in revision situations for reasons unrelated to the patella (*2,53*). Although fibrous in-growth may eventually halt this process, the implant is likely to have already adopted favourable alignment (*107*). Despite concerns of being rotationally constrained, the design concept has provided for satisfactory function and pain relief, with 10 year survival rates of 96 to 98.4% (*35,53,57,108*).

Mobile bearing patella

A different biomechanical concept has been conceived with the anatomically shaped mobile bearing metal backed patella (Fig. 1 & 2) (*11,12,13*). The design is based on the same principle as rotating platform tibial components. The clinical performance record of mobile bearing patellae has been surprisingly good and has not been characterized by the complications generally associated with metal backed prostheses. Reported survival rates of up to 99.5% at 12 years have been attributed to the high conformity and low stresses permitted by the mobile bearing articulation (*48*). The absence of significant back-side wear in mobile patellar bearings has led some clinicians to believe that these devices may not actually rotate in service. It has therefore

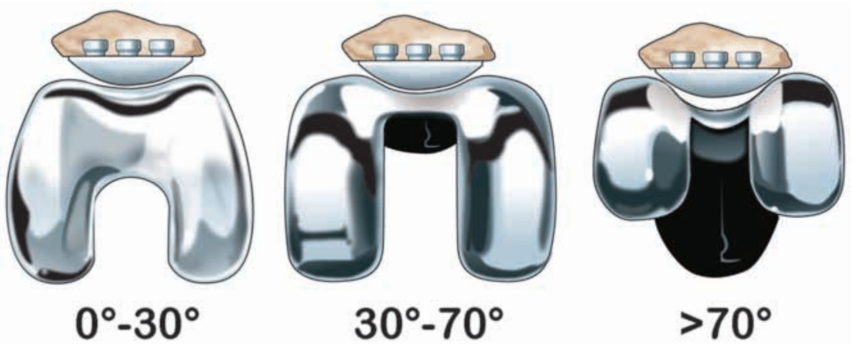

Fig. 3. — Contact position between patellar implant and femoral component at various degrees of knee flexion. Point contact in extension and early flexion, due to limited conformity between patellar implant and femoral flange ; line contact in mid-flexion (30°-70°), due to increasing conformity between patellar implant and trochlea ; bifurcation of contact area beyond 70° (Copyrights of illustration remain with author).

Fig. 4. — Dome-shaped patellar components when articulating with a designated femoral component may compensate for limited degrees of patellar tilt and rotation by maintaining acceptable contact congruency, especially in the mid-flexion range.

been speculated that the advantages of mobile bearing patellae may in fact be their ability to compensate for variations in surgical alignment by rotating into a preferential position after engagement with the femoral component and simply to stay there (*14,63*).

PATELLAR CONTACT AREA AND KINEMATICS

The contact area of the prosthetic patello-femoral joint measures, at best, no more than 40% of the contact area established for the native knee (*40,51,59,*

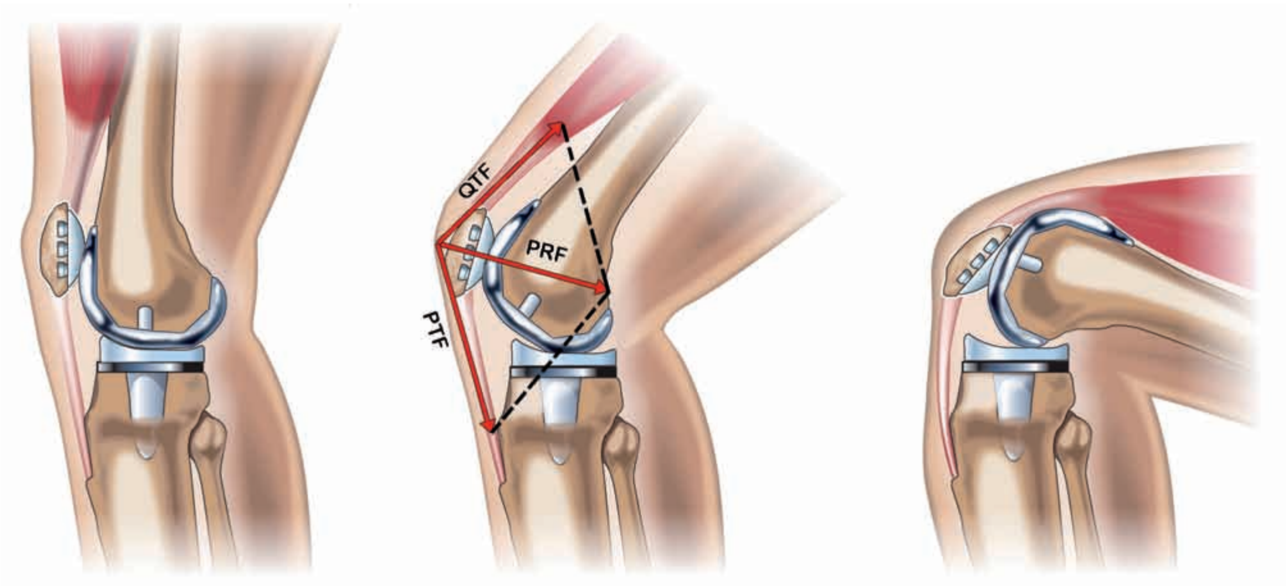

Fig. 5. — Changes of patellar position in relation to the femoral component during extension, mid-flexion and full flexion in the sagittal plane. With increasing flexion the patella declines and moves in a posterior direction with the femur, whilst the patellar contact area moves proximally. The centre image also shows the force vectors acting on the patellofemoral joint (Copyrights of illustration remain with author).

68,103,115). Measurements obtained experimentally vary widely and depend on the technical set-up and the level of compression force applied during testing. For dome shaped designs contact areas range from 13 to 162 mm^2, with highest values usually observed between 30° and 90° degrees of knee flexion (*68*). Values for modified dome shaped, anatomic, and cylindrical patellar components may vary, but due to the increased level of conformity are generally higher with contact areas of up to 270 mm^2 (*59,68,94,97*).

Up to 75° of flexion the contact area between prosthetic patella and femur is relatively large and contact pressures are generally low. As with the native patella the area of contact on the patellar component moves proximally with increasing flexion reaching the superior patellar pole between 60° to 90° depending on trochlea design (Fig. 5). Beyond this point the patella leaves the trochlea in most arthroplasty designs, leading to bifurcation of the patello-femoral contact area (Fig. 3). The transition from a one-area to a two-area contact is associated with a significant decrease in contact surface, whilst patello-femoral compressive force continues to rise. The direct influence of this transition in contact area on wear pattern can be observed in retrieved patellar components, which demonstrate deformation and development of characteristic facets at the margin of the polyethylene surface (Fig. 6) (*29,44, 62,63,114*).

As with the native patella the motion path of the resurfaced patella is complex and influenced by extrinsic stability, provided by muscle and soft tissue support, and intrinsic stability, provided by implant design. Intrinsic design stability is defined as the capacity of the implant alone, with or without patellar resurfacing, to resist interaction between implant and muscular, capsular, or ligamentous structures.

The geometry of the prosthetic components, as well as surgically imposed changes, will bring with them a plethora of variables, which all have the potential to influence patellar tracking. However, even in a well aligned and balanced total knee prosthesis the resurfaced patella will present a complex three-dimensional movement pattern broadly similar to the native knee, and predominantly consisting of rotation in both sagittal (flexion-extension) and axial (medio-lateral tilt) planes, as well as rotation and translation in the coronal plane (Fig. 7) (*82*). Studies have shown that the patella may rotate as much as 15° with respect to the femur, with most of the rotation occurring beyond 50° of knee flexion (*82*). The patella may translate medially in the coronal plane during the initial 30° of knee flexion, returning to neutral at about 60° after which it moves laterally by as much as 8° (*70*). Finally, for every 30° of knee flexion, 20° of patello-femoral flexion occur, defined as rotation in the sagittal plane (Fig. 5) (*55*).

Fig. 6. — Focal area of polyethylene deformation (cold-flow) at the periphery of a retrieved patellar component, created through bifurcation of contact at flexion > 70°. Changes considered to accommodate non-conformity between patellar implant and femoral condyle by creeping in ways to reduce contact stress . Component courtesy of Prof. W. Plitz, Ludwig-Maximilian-University, Munich.

Stiehl *et al* (*99*) assessed patellar kinematic patterns and were able to demonstrate that patellar axis rotation, which compares the angle between the patellar tendon and the sagittal axis of the patella, increases with flexion in TKA (Fig. 5 & 7) beyond the levels observed in normal knees. Contact position of dome-shaped and anatomically shaped patellar components showed greater variability compared to the normal knee, with the average contact position for the resurfaced patellae lying more superior, and tilt angles being significantly increased. However, the kinematic behaviour of an anatomically shaped or an unresurfaced patella

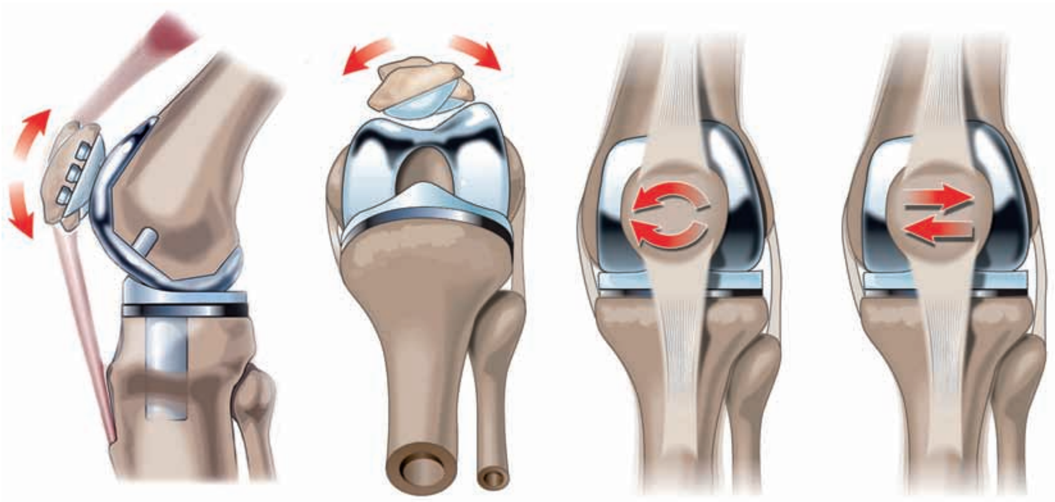

Fig. 7. — Various types of patellar movement known to occur in both resurfaced and un-resurfaced patellae when articulating with a prosthetic femoral component. From left to right : rotation in the sagittal plane (flexion-extension), in the axial plane (medio-lateral tilt), and in the coronal plane ; translation in the coronal plane (Copyrights of illustration remain with author).

more closely resembled normal knee kinematics, compared to those observed with dome shaped designs.

The complexities of the patello-femoral movement pattern highlight the difficulties in reproducing natural patellar kinematics when resurfacing the patello-femoral joint. Although an unconstrained patello-femoral articulation would allow the patella to move relatively unrestrictedly, it requires a low level of conformity between the mating surfaces, which in turn would lead to an increase in contact stresses. In contrast, a highly conforming articulation will constrain patellar movement, imparting unwanted shear forces which may increase the risk of patellar subluxation and component loosening.

In a cadaver study Kim *et al* (*51*) assessed the effect of patellar kinematics on the contact area of dome, modified dome, anatomic and rotating patellar designs. Under optimal tracking conditions the contact areas of the dome shaped patella were significantly smaller compared to the modified dome and anatomic designs. When exposed to 3-dimensional movements however, the contact area of the dome shaped patella was significantly greater, indicating enhanced forgiveness regarding patellar malpositioning, whilst modified dome and anatomic components appeared more sensitive to patellar mal-alignment.

PATELLO-FEMORAL FORCES

The mechanical environment of the replaced patello-femoral joint differs quite significantly from the natural knee and is biomechanically disadvantaged by having smaller contact areas through which high contact stresses are transferred. Anterior patellar strain, a measure of the effect of external forces on the geometric configuration of the patella, has shown a threefold increase following TKA (*62*). Contact stress, measured in megapascal (1 MPa = 1 N/mm^2), is defined as force divided by the area over which the force is applied. It will increase with a rise in reaction force, but decrease with an increase in contact area. As we know from the native knee, the increase in patello-femoral contact area with flexion up to 90° together with the 'turn-round' phenomenon of the quadriceps tendon beyond 90°, help to dissipate the patello-femoral reaction force (PRF) over a larger area (*37,38*). Despite these compensatory mechanisms however, we observe a net increase in contact stress during flexion in TKA as reaction forces increase disproportionately compared to the contact area.

The forces transmitted by the patella originate from the pull of the quadriceps, resulting in a tension force in the patellar tendon and a contact pressure force between the patella and the trochlea. In a practical simplified model, these forces act coplanar (in the sagittal plane) and even concurrent, in such a way that it is permissible to consider them as a single resultant force (Fig. 5). Experimental *in vitro* studies have been able to show that these forces can be quite considerable, with PFR values of 1.2 × body weight (BW) for simple activities such as walking on level ground, 5.7 × BW for descending stairs or rising from a chair and 7.7 × BW for jogging (*87*). *In vivo* studies have so far only looked at peak forces generated within the replaced tibio-femoral joint, which confirmed approximate values of 1.3 × BW for biking, 2.7 × BW for walking, 3.8 × BW for tennis and up to 4.5 × BW for golf (*21*). The level of contact stresses is directly influenced by the magnitude of the contact force (PRF). The magnitude of PRF is a function of implant design. Certain patient demographics like younger age, above average BMI, and increased post-operative flexion, especially in those patients of high demand, are likely to further increase the level of compressive and shear forces on the patellar component during knee flexion (*29,64,72*).

The fixation surface of all-polyethylene onlay patellar components has also been subject to biomechanical investigations. Large single central fixation lugs, which were popular in the 1970s and 80s, required significant bone removal leaving only a relatively shallow bone bridge below the lug. This created focal stress raisers leading to an increased risk of patellar fracture close to the fixation site (*9,19*). Single lug patellar components have hence been largely abandoned in favour of three smaller fixation lugs placed more peripherally. Such an arrangement is subject to less stress compared to lugs placed centrally, especially if lugs are oriented in a transverse direction (*17*). The construct of three smaller and peripherally placed lugs has been shown to avoid precarious bone weakening and provides better resistance against tilt and rotational forces (*54,58*). Inlay patellar components of convex, biconvex or cylindrical configuration are inset into the retropatellar surface and continue to use single peg fixation. These pegs are usually quite small in size and the low rate of complication with this technique may be due to the additional strength gained through peripheral bone preservation. In addition the particular geometry of the patellar component and the moving centre of loading produced by knee mechanics and interaction with the femoral trochlea impose peculiar stresses on the patellar fixation site. The resulting strains are compressive, shear, and tensile in character and presumed to be relatively small. They are often referred to as ‘micromotion’ and implicated as a mechanical contribution to loosening (*8*).

MATERIAL SCIENCE AND PERFORMANCE OF PATELLAR IMPLANTS

Owing to the great disparity between moduli and strength of cobalt-chrome alloys on the one hand and ultra high molecular weight polyethylene (UHMWPE) on the other, *wear* is primarily observed on the polymeric side of the prosthetic patello-femoral articulation. Notwithstanding its limitations, UHMWPE has evolved as the material of choice for the patellar component based on the low friction principle (*16*). Mechanical properties of UHMWPE are far from being ideal, with *yield* strength affected by the level of molecular weight, degree of cross linking and sterilisation method.

Uniaxial yield strength of UHMWPE, which equals the lowest stress at which the material undergoes plastic deformation, is estimated at around 23 MPa (*3,20,39,83,96*). Concerns have been raised if such stresses are applied continually. For industrial applications repeated maximum contact stresses of 10 MPa are hence recommended, a value which incidentally is identical to the yield strength estimated for articular cartilage (*80*). Buechel *et al* (*12*) have even suggested that for long-term human use maximum contact stresses of 5 MPa may be more appropriate, as body temperature further reduces the strength of UHMWPE by almost 25%. *In vitro* contact stress analysis has confirmed that all-polyethylene dome shaped patellar components produced contact pressures between 20 to 30 MPa in extension, rising to between 36 and 100 MPa at 90° to 120° of knee flexion, therefore exceeding the yield strength of UHWMPE (23 MPa) by up to 400% (*20,47,51,63,115*). Anatomically shaped rotating platform patellar components produced significantly lower values, mostly staying below the yield strength of UHMWPE (*20,63*). *Wear* simulator studies further confirmed that congruent patellar components (modified dome and anatomic) exhibited significantly lower rates of creep and wear than dome shaped designs, again indicating that conformity is critical to wear resistance and protection against post-*yield* deformation (*20,43,44*).

Viscoelastic properties of surface cartilage allow for its deformation under load and subsequent increase in pressure transmitting area. Due to differences in elastic modulus between cartilage and UHMWPE, the prosthetic patella however has limited ability to change its surface contact area through variations in patello-femoral load (*6,42,44,63*).

Xu *et al* (*115*) and associates were able to demonstrate the effect of patellar resurfacing on contact area and pressure in cadaveric knees. The mean contact area between 30° to 120° of flexion in the non-resurfaced patello-femoral joint ranged from 70 to 150 mm^2, whilst peak patellar contact pressures did not exceed 12 MPa (*115*). Once resurfaced the mean contact area decreased almost 10 fold to 10 to 15 mm^2, creating a dramatic increase in patellar contact pressure values of 50 to 100 MPa. Greenwald *et al* (*15,68,98*) performed biomechanical studies assessing patellar surface contact area, compression force, and contact pressure using a variety of different prosthetic models. These authors (*15*) found that patello-femoral contact pressure values at knee flexion angles beyond 45° exceeded polyethylene yield strength in all tested components with peak measurement of up to 75 N/mm^2 (= 75MPa) (Fig. 8). The authors postulate that contemporary component designs should provide for congruent patellar contact throughout flexion and extension, which seeks to minimise surface and sub-surface stresses.

Steubben *et al* (*97,98*) measured the distribution of patello-femoral surface stresses by mapping areas above and below the tensile *yield* strength of polyethylene. All implants whether of dome, modified-dome, or anatomic shape demonstrated material yielding over their range of flexion. Their results indicated the importance in appreciating the location of the yield areas within a given patellar component, as rim loaded contact areas above yield are more likely to deform and *wear*. It has hence been suggested that polymer integrity does not rest primarily with the size of the contact area, but rather with the extent of the surface within this region which exceeds yield condition.

Subsequently, contact stresses above the *yield* strength do not necessarily lead to catastrophic failure as demonstrated by the large number of relatively undamaged retrievals. As highest values of contact stress are experienced during flexion, variations in patient’s activity may not expose the patellar component to large cyclic loads frequently enough to accumulate damage. McNamara (*63*) considered the constraining effect of surrounding polyethylene responsible for this phenomenon. *Yield* in polyethylene is characterized by plastic deformation (creep) rather than brittle failure, which explains why non-conforming patellar components are capable of ‘wearing-in’ (*6*). Retrieval studies (*32,114*) have shown that creep of polyethylene occurs independent of *wear,* which permits adaptation to the tracking position. Such surface adaptation produces characteristic facets at the margin of spherical patellar components, increasing the contact area particularly in flexion where there is least congruency between patella and femur (Fig. 6) (*19,44,63*). Although reduction in contact stresses of 23% to 58% through increased conformity have been reported, contact stress values remain above the UHMWP yield strength (*20,29,40*). Elbert *et al* (*29*) were surprised that, despite artificial “wearing in” of a polyethylene patellar surface into a concave shape, the von Mises stress (a criterion used in predicting the onset of *yield* in ductile materials) was at or near the polyethylene yield stress in most of the contact areas, which suggested that the deformation might continue anyway. Williams (*112*) found, via analysis of von Mises stress, that most stresses above yield strength occurred 1-2 mm below the articulating surface area in the newly manufactured component, whilst in retrieved com-

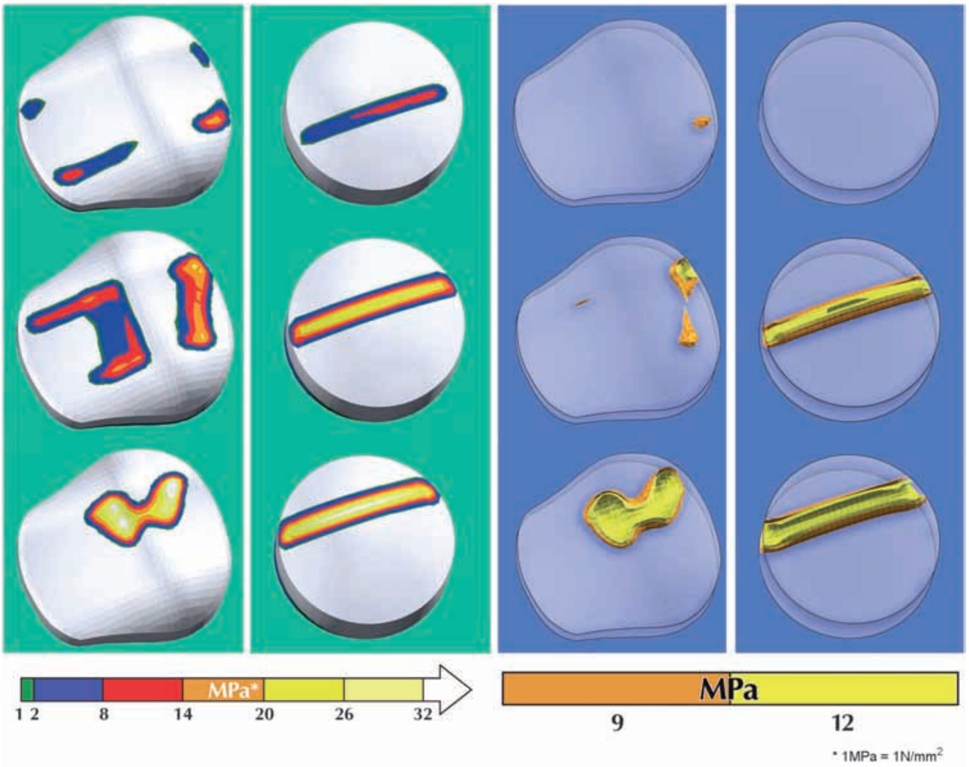

Fig. 8. — Green background : contact areas and contact stress images, simulating walking gait (15°, PRF = 420N, top row), stair ascend (45°, PRF = 1760N, middle row), and chair raise (90°, PRF = 1950N, bottom row) for anatomic and dome shaped patellar components. Purple background : von Mises subsurface stress images for the same activities, illustrating the volume of polymer stressed above 9 MPa (*68*). Illustration courtesy of Seth Greenwald of the Orthopaedic Research Laboratories, Cleveland/OH, USA.

ponents von Mises stress remained near yielding through the depth of the implant (*112*) (Fig. 8). Due to sub-surface stresses, permanent deformation may henceforth be expected to continue even when the component has ‘worn-in’ (*29*). Although Collier *et al* (*20*) conceded that “*all-polyethylene patellar components are not the answer as an ideal bearing surface*”, in the absence of a suitable alternative, UHMWPE is likely to remain the material of choice at least in the foreseeable future (*20*).

FEMORAL COMPONENT DESIGN

The patella, whether native or prosthetic, cannot be considered in separation as it works in direct partnership with the femoral component. Contact areas are highly dependent on the congruency of the patello-femoral joint articulation at all angles of knee flexion, whilst motion constraints of the patella are determinated by the surface geometry of the femoral component (intrinsic stability) and by the balance of soft-tissue forces (extrinsic stability). Following on from the disappointing results of early arthroplasty designs which frankly ignored the patello-femoral joint, Seedhom suggested an array of design changes to the femoral component in order to improve patello-femoral kinematics and function (Fig. 9) (*91*). It is generally believed that a more congruent patello-femoral articulation with a deepened trochlear groove that extends both proximally and distally together with a build-up lateral trochlear wall is likely to provide for improved patellar tracking and enhance patellar stability during flexion and extension (Fig. 10) (*18,91,104, 116*).

Bartel *et al* (*3*) demonstrated the importance of conformity in prosthetic design to increase contact area and to decrease contact stress. Current femoral prosthesis designs display a wide variation in length, depth and orientation of the trochlear groove, sagittal radius, and axial geometry (*25,104*). Anatomically shaped femoral component designs appear to be particularly suitable when articulating against the non-resurfaced patella, and hence referred to as 'patella-friendly (Fig. 11). They provide increased conformity between native patella and femoral component and require minimal biological patellar remodelling (*12,49*). Non-anatomical designs are those where the trochlear groove is concave spherical and designed to accommodate a non-anatomical patella usually of dome-shaped design.

The group of Freeman (*30,36,53*) believes that the design of the trochlea is the key feature in provid-

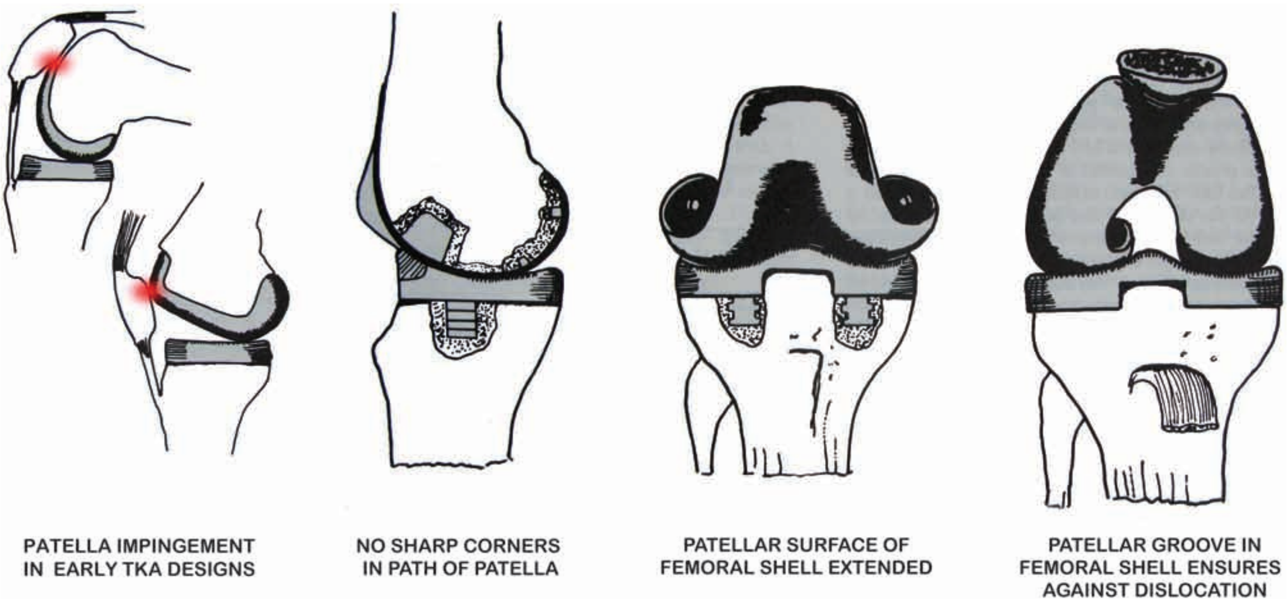

Fig. 9. — Failure to accommodate the patella in early arthroplasty designs resulted in patello-femoral impingement (areas denoted in red) and anterior knee pain. The 3 images on the right denote design alterations suggested by Seedhom in 1974 to overcome problems of impingement and to improve patello-femoral kinematics and function (*91*).

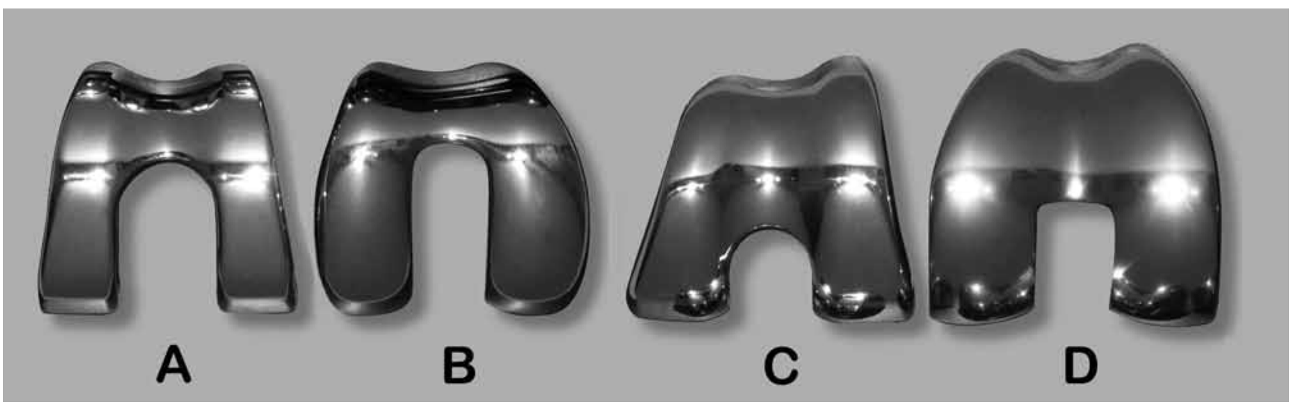

Fig. 10. — Characteristic design features of various femoral components which have shown to exert significant effects on patellar kinematics and biomechanics. Femoral components with relatively 'patella-unfriendly' design features usually provide a symmetric, shallow and short trochlear groove (A = unmodified Ortholoc®, Dow Corning Wright ; B = Townley®, Biopro). Femoral components with relatively 'patella-friendly' design features usually provide an asymmetric, deepened central femoral groove, elevated lateral trochlear flange, and distal extension of trochlear groove (C = modified Ortholoc®, Wright Medical ; D = Buechel-Pappas®, Endotec). Ortholoc implants courtesy of Leo Whiteside of the Missouri Bone & Joint Research Foundation, St. Louis, MO.

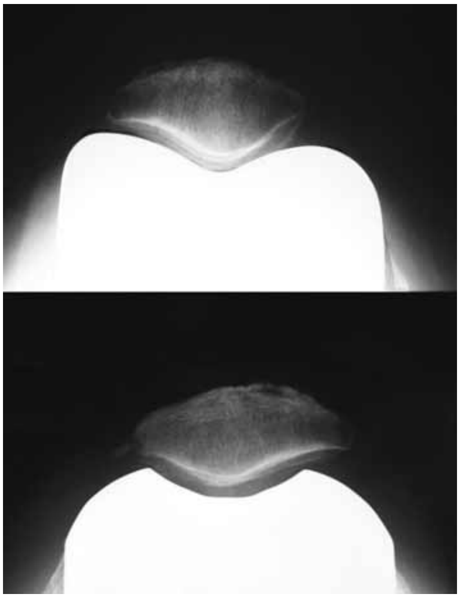

Fig. 11. — Postoperative radiographs (Merchant's view) showing a 'patellar-friendly' anatomic femoral component (top) and a patellar unfriendly femoral component (bottom), both articulating against the native patella.

ing satisfactory clinical results. They postulate that the floor of the prosthetic trochlea, viewed from the side, should be circular (single radius), similar to the native knee, extending from 0° to 110° of flexion. Furthermore it should be recessed to an anatomical extent in order to restore the patello-femoral joint line, a feature not to be confused with the height of the patella in relation to the tibio-femoral joint. In a large cohort study using such a design the same authors found no clinical differences between resurfaced and native patellae at a mean follow-up of 10 years. Based on this observation it has generally been accepted that increasing the radius of curvature and deepening of the trochlear groove reduces patello-femoral shear and compressive forces *(15,18,74,104)*. Some experimental evidence also exists that the depth of the trochlear groove may be a more important variable in the prevention against patellar subluxation than the shape of the articulating surface itself *(15)*. Excessive deepening of the trochlea groove however will decrease the moment arm of the quadriceps muscle force as the patella is brought closer to the centre of rotation of the knee.

The importance of femoral component design and its influence on patello-femoral performance has been highlighted by Theiss *et al (104)*, based on clinical results of two arthroplasty designs with distinct differences in trochlear geometry. A 14-fold decrease in patellar related complications was observed when using a patellar friendly design. Similar results have been reported by Yoshii *et al (116)* in an experimental study. These authors were able to demonstrate that specific femoral design changes (e.g. 1 mm deepening of the trochlear groove, elevation of the lateral trochlear flange) improved patellar tracking compared to an unmodified femoral component (Fig. 10).

Proximal extension of the femoral flange will help to capture the patella during early flexion whilst extension of the concave shape of the trochlear groove onto the intercondylar surface will allow for increased metal-to-plastic contact at higher flexion angles *(85,90,116)*.

The effect of valgus alignment of the trochlear groove on shear stresses, compared to symmetrical designs, has been investigated with mixed results. Asymmetric trochlear groove designs are thought to provide for earlier patellar capture through prominence of the lateral flange and to decrease the predominant valgus force vector thus reducing patellar shear *(25)*. In some reports reduction in lateral shear forces of up to 10% was observed, whilst others saw either no effect or even a shift toward the generation of medial shear forces *(18,25,74,104)*. The exact clinical advantages of asymmetric designs have remained largely theoretical, lacking compelling clinical proof of their effectiveness *(18,111)*.

Lateral subluxation. Compressive and lateral forces acting at the patello-femoral articulation increase with knee flexion *(110)*. The magnitude of the lateral forces, which are depending on valgus alignment, Q-angle and soft tissue balance may, if excessive, cause patellar subluxation and contribute to component failure. Steubben, Postak and Greenwald *(98)* investigated the resistance offered to lateral subluxation of the resurfaced patella by defining the intrinsic lateral stability of various patello-femoral designs. They disregarded surgical variables such as component placement, alignment and correction of varus and valgus deformity, but recognized their importance in assisting this process. They found that the medio-lateral component of force was highly dependent on the interaction of condylar and patellar surface geometry. All tested implants presented force values required to produce lateral subluxation at or above those measured for the native knee. Force values of up to 2250 N at 90° were generated by some designs representing a 6 fold increase compared to the native patella, highlighting that appropriate design changes, e.g. deepening of the trochlear groove, can significantly increase resistance to patellar subluxation (*97,98*).

Conformity or non-conformity between femoral and patellar components (*82,99,101,109*) ? The level of conformity between femoral and patellar components influences the joint's ability to tolerate natural variations in motion. Conformity increases contact area and stability, whilst non-conformity allows the patella to establish an 'equilibrium of forces', and avoids excessive shear forces from arising. Potential advantages of conforming designs may hence be offset by an increase in constraint, potentially resulting in deleterious effects on patellar tracking and fixation. This typically leads to a compromise whereby conformity and subsequently contact areas are reduced to avoid over-constraining the joint. The question however, of how much contact area to sacrifice and how to best achieve this compromise remains unanswered. As a general trend most clinicians favour spherical patellar implants over anatomic patellar designs for ease of application, since they are less prone to surgical malalignment (*79,84*). Especially in combination with a mobile bearing TKA, dome patellar components have provided improved tracking and reduction in patello-femoral contact stress (*86*).

EFFECT OF CRUCIATE RETENTION OR SUBSTITUTION

Moment arms affecting the patella are dependent on the distance between the patello-femoral joint to the axis of rotation (flexion and extension) of the femoral component. They are increased if the axis is deviated posteriorly from its physiologic position. Femoral rollback facilitates this process and represents a characteristic feature of normal knee kinematics. Increased rollback effectively lengthens the patellar moment arm, thus increasing the efficacy of the extensor mechanism. D'Lima *et al* (*25*) investigated the influence of various degrees of posterior femoral rollback on patellofemoral compressive force. Femoral rollback resulting from PCL preservation produced reductions in patellofemoral compressive force of up to 7% throughout knee flexion, whereas the effect in PCL-substituting devices only became noticeable after cam-post engagement, with maximum effect recorded at 85 degrees of knee flexion. Miller *et al* (*66*), in an earlier study comparing PCL-retaining with PCL-substituting arthroplasties, failed to note femoral rollback when the PCL was retained. They stipulated that the absence of the anterior cruciate ligament may render the PCL ineffective, which may explain the appearance of paradoxical movements (reverse rollback) observed on fluoroscopic investigation (*22,23,25*). Although PCL substitution kept patellofemoral forces close to the level of the native knee, a lateral release became necessary in 50% of knees, raising potential concerns about an increase in patellofemoral stress through ligamentous tension. This notion has also been expressed by Ranawat and Sculco (*75,76*), who raised concern that femoral rollback either through a cam and post mechanism, as in posterior-stabilizing designs, or through a functional posterior cruciate ligament (PCL) may increase tensile forces across the patella in flexion. Overall, patellar thickness following resurfacing should therefore not exceed preoperative values, particularly in posterior-stabilized designs, as this will tighten the extensor mechanism, create loss of flexion, and increase both anterior patellar strain and PRF (*27,65,81,95,106*).

THE UNRESURFACED (NATIVE) PATELLA

Due to differences in the modulus of elasticity, the articular surface of the patella, if left un-resurfaced, must adapt to the geometry of the opposing surface by bedding-in. This process of remodelling, also known as 'stress contouring', produces gradual adaptation of the retro-patellar surface and subchondral bone plate to the trochlear shape (*93*).

Keblish *et al* (*50*) noted that minimal remodelling was required if the patella was exposed to an anatomical design with constant radius of curvature and uniform femoral geometry, whilst excessive remodelling was observed in non-anatomical designs. The remodelling process is time dependent and not displayed on axial radiographs much before two years after implantation.

Matsuda *et al* (*59*) assessed patellofemoral contact stress and contact area following TKA by comparing a non-conforming dome patella, a conforming anatomic patella and an unresurfaced patella with those values obtained in the native knee. In the un-resurfaced patella, peak contact stress and contact area remained almost at the level of the native knee. Following patellar resurfacing patello-femoral contact stress rose beyond yield strength for UHMWPE, with an average increase of 200%, whilst patello-femoral contact area decreased on average by 60%. The authors concluded that although the effect of metal action on cartilage was uncertain, the option of leaving the patella without a prosthetic component remains an attractive one. This is thought to apply especially to those cases where the patella is not severely worn, as peak stresses are known to be closer to normal if the patella is left unresurfaced.

Tanzer *et al* (*102*) looked at the effect of femoral component designs on contact and tracking characteristics of the unresurfaced patella in total knee arthroplasty. The authors noted substantial alterations in patellofemoral contact areas, contact pressures and tracking at higher flexion angles when the native patella was articulating with a prosthetic femoral component. The percentage of patellofemoral contact area compared to the native knee reduced markedly with increasing knee flexion, with measured values of 79%, 69% and 65% at 60°, 90° and 105° respectively.

The surface geometries of some prosthetic femoral components, particularly those of posterior-stabilized design, appear incompatible with the native patella, as the apex of the retropatellar ridge may impinge on the prosthetic intercondylar notch beyond 90° of knee flexion. Patellar deformation and wear are likely consequences and in the case of significant patellar tilt, displacement of the patella into the notch becomes possible (*62*). Whiteside's group (*116*) was able to show that distal extension of the trochlea and shortening of the intercondylar notch safeguard patellar support beyond 90° of knee flexion. Such design modifications are hence important if one considers leaving the patella unresurfaced. Most current femoral components present a surface geometry designed to articulate with a designated patellar component but are ill equipped to accommodate the native patella (Fig. 11) (*60*). Specific efforts are required to improve patellar kinematics and biomechanics by creating a femoral component which not only conforms to the normal trochlea and intercondylar notch topography, but which also takes the movement pattern of the native patella into account. Only then would we be in a position to offer prostheses dedicated to be used with the native patella, compared to the mostly inadequate femoral designs currently available.

SURGICAL EFFECT ON PATELLAR TRACKING

Despite major contributions through geometrical specifics of the replacement, performance and function, the resurfaced patella remains highly dependent on the surgical technique safeguarding correct placement of the components (*52,77,78,88*). Decisions made by the surgeon can compensate for implant design limitations, but conversely may also exacerbate such limitations (*26*). Any digression from the ideal position may affect the proper function and lead to deviation from the ideal tracking pattern. If mal-tracking is not corrected it may increase shear stresses at the fixation site which are likely to increase wear and affect the long-term survival of the patellar component (*49,105,107*). Intra-operative assessment of component positioning is unable to account for the effect of muscles and tendons on the kinematic behaviour of the replacement. Despite careful surgical technique patellar tilt often occurs, as intraoperative tests are static while postoperative function is dynamic (*7,73*).

Mistakes that are known to detrimentally affect patellar tracking are manifold and relate to component mismatch and sizing errors (e.g. undersized

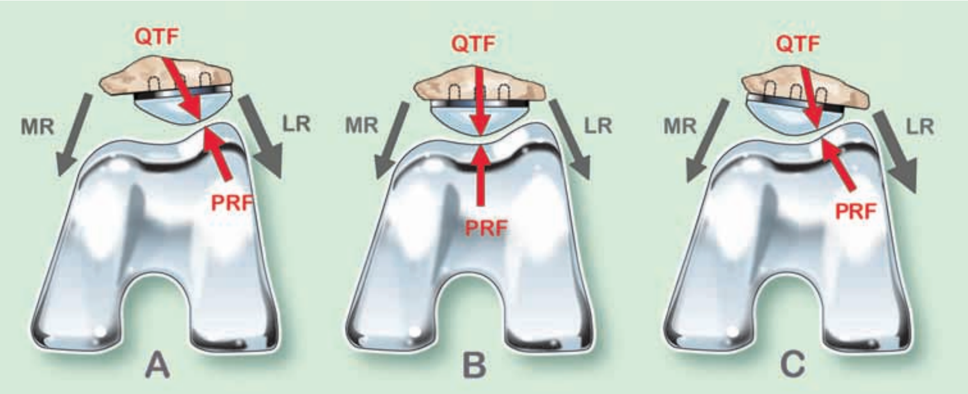

Fig. 12. — Left knee : lateral positioning of the patellar implant on the retro-patellar surface (A) will tighten the lateral retinacular structures (LR) and may provoke lateral subluxation. Overzealous medial positioning of the patellar implant (C) may create lateral patellar tilt through off centre positioning of the quadriceps tendon force (QTF). Moderate medialisation (B) will re-create the asymmetrical contour of the native retropatellar high point, centralise both, QTF and patello-femoral reaction force (PRF), and improve patellar tracking (Copyrights of illustration remain with author).

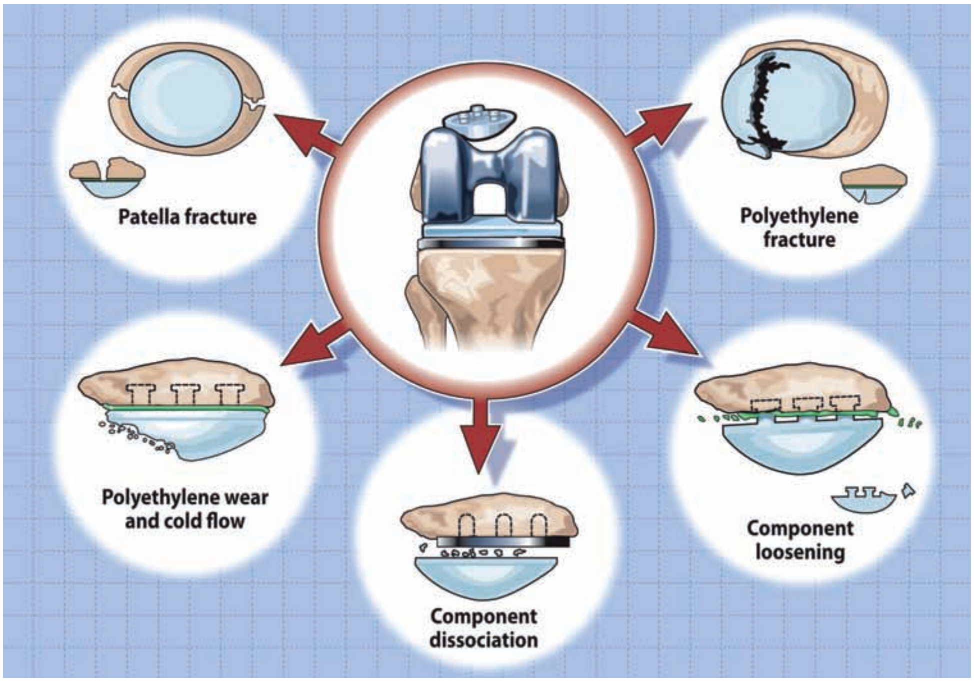

Fig. 13. — Potential complications, commonly associated with patellar mal-tracking, following patellar resurfacing (Copyrights of illustration remain with author).

patella, overstuffing), component malpositioning (e.g. lateralisation of patellar component, internal rotation and medialisation of femoral component, internal rotation of tibial component, excessive joint line elevation > 8 mm), overall leg mal-alignment (e.g. excessive valgus or varus), and ligamentous imbalance (*1,5,31,66,71,79,81,85*). Even minor alterations, such as patellar component placement on the retropatellar surface, have been shown to influence intra-articular force distribution (Fig. 12) (*1*). Any of the aforementioned surgical improprieties may exert a cumulative effect on patellar tracking and stability, potentially leading to disastrous results (Fig. 13). The author would like to refer the reader to other publications on this subject matter as this aspect remains outside the realm of this article (*73,78,79,88*).

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