Patellofemoral Mechanics: a Review of Pathomechanics and Research Approaches

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Abstract

Purpose of Review The patellofemoral joint is a complicated articulation of the patella and femur that is prone to pathologies. The purpose of this review is to report on the current methods of investigating patellofemoral mechanics, factors that affect joint function, and future directions in patellofemoral joint research with emerging technologies and techniques.

Recent Findings While previous hypotheses have suggested that the patella is only a moment arm extender, recent literature has suggested that the patella influences the control of knee moments and forces acting on the tibia as well as contributes to various aspects of patellar function with minimal neural input. With advancements in simulating a six-degrees-of-freedom patellofemoral joint, we have gained a better understanding of patella motion and have shown that geometry and muscle activations impact patella mechanics. Research into influences on patella mechanics from other joints such as the hip and foot has become more prevalent.

Summary In this review, we report current in vivo, in vitro, and in silico approaches to studying the patellofemoral joint. Kinematic and anatomical factors that affect patellofemoral joint function such as patella alta and tilt or bone morphology and ligaments are discussed. Moving forward, we suggest that advanced in vivo dynamic imaging methods coupled to musculoskeletal simulation will provide further understanding of patellofemoral pathomechanics and allow engineers and clinicians to design interventions to mitigate or prevent patellofemoral pathologies.

Keywords Patellofemoral joint · Computational modeling · Biomechanics · Morphology · Imaging · Mechanical function · Statistical shape modeling

Introduction

The patellofemoral joint (PFJ) is the complicated articulation between the patella and the femoral trochlear groove. The patella is the largest sesamoid bone in the human body, possesses the thickest cartilage, and has seven articulating surfaces with varying lengths and curvatures. The patellar ridge separates the medial and lateral facets on the articulating

surface and can be located centrally or offset to the medial or lateral directions (Fig. 1). The patella connects the knee extensor muscles (vastus medialis, vastus intermedius, vastus lateralis, and rectus femoris) to the tibia and contributes to knee extension by transferring the quadriceps force to the tibia via the patellar tendon.

At full knee extension, the patella is situated above the trochlear groove with some articular surface contact in the distal lateral facets. As the knee is flexed, the patellar tendon pulls the patella into the trochlear groove with initial lateral facet contact causing slight medial translation. Around 30° of knee flexion, the medial facets engage in the trochlear groove [1]. Patella flexion increases with knee flexion, with patella flexion reaching the equivalent knee flexion angle 20% later and both the patella and knee reaching maximum flexion at the same time [1, 2]. With increasing knee flexion, there is increasing lateral translation and lateral tilt of the patella after initial movement in the medial direction [2, 3]. After 45° of

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Fig. 1 Anterior and posterior views of the patella with the anterior view depicting regions of the patella and the posterior view showing the facets and patellar ridge of the articular surface

knee flexion, the patella slowly increases in medial rotation. In deep flexion (greater than 90°), contact occurs on the proximal half of the medial and lateral facets and the majority of people have the odd facet engaged at maximum knee flexion [4]. Passive flexion results in contact on the lateral facets while active flexion causes a larger contact area that is similar throughout motion. Passive structures, such as ligaments and bone morphology, and active muscles control patella kinematics, which subsequently influence contact area and mechanics. Current hypotheses suggest kinematic differences in the pathologic population include patella tilt, patella translation, and patella alta, which put the patella in an abnormal location and affect its influence on knee extension.

It has been suggested that the patella's sole function is to provide mechanical advantage to the quadriceps muscles by extending their moment arms. A larger moment arm reduces the muscle forces required to generate torque at the knee. A recent study argued that the patella is more than just a moment arm extender because the proximal-to-distal delivery of forces to the femur and tibia is aligned with the control strategy during jumping maneuvers. This suggests the PFJ provides some *mechanical intelligence* to reduce the burden on the controller [5].

This mechanical intelligence is important because the PFJ is a source of common pathologies affecting a large proportion of both sedentary and active populations from adolescents to older adults [6–8]. Without understanding the mechanical function of the patella, it is not possible to understand why people with PFJ pathologies developed issues in the first place, or determine which mechanical factors are important to correct. A common patellofemoral issue is patellofemoral pain (PFP). PFP affects up to 25% of the general population [9] and tends to persist up to 8 years, but in some cases can exist for up to 20 years [7, 10]. The negative effects of patellofemoral pain are high—individuals with patellofemoral pain often report a lower quality of life compared with

controls and patellofemoral osteoarthritis is most commonly diagnosed to individuals with a history of patellofemoral pain [11, 12]. A common hypothesis to the cause of patellofemoral pain is elevated joint stress [13]. Elevated joint stress may be instigated by reduced contact areas, patellar malalignment, and patella maltracking [13, 14]. Abnormal frontal plane knee and sagittal plane knee and hip mechanics, such as knee adduction and internal hip rotation, have been associated with patellofemoral pain and knee injury [15]. Cross-sectional studies have demonstrated that individuals with patellofemoral pain have greater patellofemoral joint stress compared with asymptomatic controls during walking, running, squatting, knee flexion-extension, and stair ascent and descent [16–20]. Femur internal rotation and adduction have been shown to increase patella cartilage stress, emphasizing the importance of analyzing hip kinematics and other joints when rationalizing changes in patellofemoral mechanics [21]. This suggests the importance of investigating influences on patella mechanics that extend beyond the knee joint.

There is a large variance in physical size and articular surface shape of the patella, and its morphology has an influence on patella kinematics [22]. The average patella has a width-to-height ratio of 1.3 (1.04–1.63) with an average articular surface height of 35.7 (28–47) mm [22]. The average maximum thickness of the patella is 22.6 mm [23]. The cross section where this occurs varies depending on the patella. Wiberg [24] classified the patella into three types based on the posterior surface shape in the transverse plane. The three types describe the differences in symmetry, concavity, and location of the patella ridge. There is an association between type 3 Wiberg patella shape (smaller convex medial facet versus concave lateral facet) and patella dysplasia [25]. The vast difference in patella shape and resultant kinematics emphasizes the importance of considering subject-specific geometry during model simulations. Group mean results may be misleading if

the conclusion for kinematics spans across a spectrum of knee geometries.

Current approaches to investigating patellofemoral mechanics are limited due to the difficulties of directly measuring patellofemoral alignment and kinematics during functional tasks due to the movement of soft tissues surrounding the joint. Technological advancements have improved the accuracy of measuring bone position and orientation (pose) within the patellofemoral joint. This is critical as minor inaccuracies of 0.5 mm in pose can influence maximum joint contact force estimates by up to 578% [22]. In this chapter, we will review current methods for measuring patella mechanics in vivo, in vitro, and in silico and discuss the factors affecting the mechanical function of the patellofemoral joint. We will report on current discoveries related to present methodologies and suggest future directions of the research.

Methods for Studying the Patellofemoral Joint

In Vitro

Experiments using cadaver models provide several advantages. First, they allow tissues to be sequentially dissected out to provide insight into their role in PFJ mechanics. The structural and material properties of the tissues can be quantified, and the muscle forces can be specified and measured. Two major disadvantages of cadaver models include (1) the extreme challenge of replicating in vivo muscle activations and external loads, and (2) most cadavers are sourced from older individuals who likely have different structural and material soft tissue properties compared with younger populations affected by PFJ pathology. Despite these limitations, cadaver models have provided tremendous insight into PFJ function and dysfunction.

A popular experimental paradigm is to secure a cadaver knee specimen to a rig that allows for the knee to be fixed at set tibiofemoral flexion angles [26, 27] or continuously flexed [28, 29]. Forces are applied to relevant muscles via pulleys and weights [30, 31] or hydraulic actuators [29, 32]. The increased access to the joint in this approach greatly simplifies measurements of patellar kinematics using optical [32, 33] or other [34, 35] motion tracking techniques. Contact forces and pressure can be measured in vitro through the use of pressure-sensitive films [36–38]. Mechanical stability of the patellofemoral joint can also be quantified by applying a force to the patella and measuring the resulting displacement [35, 39, 40].

In Vivo

Studying patellofemoral joint biomechanics in vivo is attractive as this allows a better understanding of the true joint

function and variability that exists in healthy and pathological populations. A limitation of in vivo methods is that measurements typically are superficial and measurement of patellar movement beneath the skin and variables such as joint loading is challenging. Additionally, there is limited opportunity to manipulate the variable of interest or control confounding factors such as the stiffness or line of actions of ligaments.

Optical motion capture is frequently used to measure tibiofemoral joint kinematics; however, the large amount of patella motion underneath the skin limits its usefulness in measuring the small translations and rotations of the patella. There have been attempts to use “clamps” that fit over the patella with attached markers to enable the use of optical motion capture [41, 42], but these have rarely been used. The use of intracortical pins can accurately track bone motion [43]; however, these are highly invasive and likely alter natural joint function.

Magnetic resonance imaging (MRI) has been used to accurately measure the patella position, orientation, and geometry non-invasively. Patella tracking has been assessed statically at fixed knee flexion angles in both unloaded [44] and loaded [45] conditions. It has been questioned whether static measures of patellar tracking are reflective of patellar tracking during dynamic activities, however [46, 47]. Techniques have been developed that enable measurement of patellar kinematics for dynamic motions using MRI. In dynamic MRI, images in selected planes are compiled over a number of cycles of repeated motion for a series of time points during the cycle [48–50]. The assumption is that the motion is repeatable for each cycle. Most MR equipment requires participants to lie supine within the machine (Fig. 2a). Vertical open-bore systems exist that allow weight-bearing images to be obtained [51], but these tend to produce lower-quality images than standard closed-bore systems. In either case, the constraint of the system limits the motions that can be examined.

Biplanar fluoroscopy and biplanar videoradiography record x-ray images in two planes during a motion in a calibrated volume (Fig. 2b). The tracking of bone motion in the images can be performed based on implanted metal beads or by matching 3D geometry obtained using CT or MRI with both images. Bead-based tracking is more accurate but requires invasive surgery to implant the beads [52, 53]. These methods have been used to measure quasi-static patellar kinematics in weight-bearing lunges [54–56] and dynamically during tasks such as stair climbing [57]. Higher capture rates have allowed highly dynamic movements such as jump cuts to be investigated as well [58]. The participant is exposed to ionizing radiation, however, which limits the number of trials that can be recorded. The capture volume created by the overlapping area of the two x-ray beams is limited to the size of a basketball, although mobile biplanar videoradiography systems that follow the participant can expand the capture volume [59].

Fig. 2 **a** Example of a knee inside the MR imager for dynamic MRI performing knee flexion-extension cycles. **b** Equipment setup for biplanar videoradiography of a participant performing a level-ground walking trial

Electromyography (EMG) can provide insight into the role of muscle activation patterns in patellar function and pathologies. Both indwelling [60] and surface electrodes [61–63] have been used to compare the timing and intensity of muscle activations between those with patellofemoral pain and asymptomatic controls. Wireless sensors have been used to measure muscle activity during highly dynamic activities. However, issues of motion artifact and selecting appropriate normalization methods can create challenges with EMG. Normalizing to the maximum voluntary isometric contraction of individual muscles is time and energy consuming [64].

In Silico

Computational methods such as finite element (FE) analysis and joint-segment modeling are examples of musculoskeletal modeling that can estimate *in vivo* joint loading and mechanics [17, 65, 66]. Recent FE studies have been used to estimate measures such as bone strain [18] and have examined the effect of factors such as ligament reconstruction [67, 68] and malpositioning of the patella [69, 70]. These models allow for the simulation of six-degrees-of-freedom (three rotations and three translations) patella motion and are validated through comparison with cadaveric results. Methods have been developed over time to improve estimates by incorporating cartilage, ligaments, menisci, and bone, as well as improved physical properties of each. Probabilistic models have demonstrated sensitivity to variations in muscle forces [71] and soft tissue properties [72] when modeling the patellofemoral joint. Representations of soft tissues differ depending on software and interpretation from the researchers. FE models are often computationally intensive and are limited to the joint in question, thus requiring the definition of boundary conditions. With increasing evidence of kinematic influences from other lower-limb joints, full-body musculoskeletal modeling builds on single-joint analysis to incorporate other possible mechanical factors.

Joint-segment musculoskeletal models can be used to simulate joint function and muscle activations at the PFJ. These models represent the musculoskeletal system as rigid segments connected by joints and driven by muscles and can be used to simulate specific joints or full-body motion. These

models can be used to investigate joint mechanics with an inverse dynamics approach which uses prescribed kinematics and external forces and an optimization routine to estimate muscle forces that generate the motion [73, 74]. Alternately, a forward dynamics approach can be used where the resulting motion is calculated based on input muscle activations that are optimized for a given criterion and constraints [75, 76] or may incorporate measured muscle activations [77, 78]. Earlier versions of these models represented the knee as a hinge joint, and the patellofemoral and remaining tibiofemoral degrees of freedom were defined to be functions of the knee flexion angle [79]. These highly constrained knee models were unsuitable for studying the patellofemoral joint because they did not consider the degrees of freedom in which the patella naturally moves; however, there have been improvements to incorporate knee joints that allow for the full six degrees of freedom in both the tibiofemoral and patellofemoral joints [80–82]. These incorporate mechanical factors that alter joint motion such as cartilage, menisci, and muscle line of actions (Fig. 3). Musculoskeletal models require simplifying assumptions to be made, such as modeling cartilage contact using an elastic foundation formulation or representing ligaments as a discrete set of fibers. However, measures of patellofemoral function are easily accessible and dynamic activities can be simulated. Furthermore, they enable investigation into the effect of parameters not easily modifiable *in vivo* or *in vitro*. For example, ligament insertion sites can be modified to simulate surgery or muscle activation and force output can be changed to mimic the results of a strength training intervention. Musculoskeletal models provide the potential for asking “what if” questions and manipulating modifiable parameters.

Factors Affecting Patellofemoral Joint Function

Kinematics and Kinetics

Patella Alta

Patella alta refers to the superior position of the patella on the femur, or a “high-riding” patella (Fig. 4a). This is the result of

Fig. 3 A lower-extremity musculoskeletal model that represents the patellofemoral and tibiofemoral joints as six-degrees-of-freedom joints constrained by cartilage contact and ligament and muscle forces

a long patellar tendon, as opposed to superior insertion of the patellar tendon on the tibia [83]. Patella alta has been associated with patellofemoral pain [84, 85], patellar instability [86, 87], patellofemoral cartilage lesions [88], and patellofemoral osteoarthritis [89–91]. The incidence of patella alta in patellar instability is high, and it is thought to be a major determinant of this condition [87, 92–96]. It is theorized that a patella located too far superiorly does not engage in the trochlear

groove until greater flexion angles, thus reducing stability due to lack of bony restraint [97, 98].

The effect of patella alta on joint kinematics and cartilage contact depends on the task. A superiorly positioned patella has been associated with increased lateral translation of the patella in full extension with quadriceps muscles relaxed [90] and contracted [98]. However, while alta was correlated with lateral patellar tilt when the quadriceps were contracted [98], the correlation was weak for the relaxed case [90] and no relationship was observed for larger flexion angles [98]. Cartilage contact area decreases with patella alta in both walking and static knee postures [14, 70, 98]. While there has been no difference found in cartilage contact stress or patellofemoral joint reaction force for normal walking [14, 99], patella alta has been associated with increased cartilage contact stress for fast walking [14]. Additionally, contact stress and joint reaction force have been found to increase with alta in static knee postures for knee flexion angles greater than 80° [100, 101]. Increased joint reaction force is theorized to be a result of wrapping of the quadriceps tendon on the femur, which causes load sharing being delayed until deeper flexion angles. The quadriceps force was found to be higher than the patellar tendon force with patella alta [102, 103]; however, the patellar tendon moment arm decreased with alta during walking [102] but increased with alta for static knee postures [103].

Patella Translation and Patellar Tilt

Patella translation refers to the horizontal motion of the patella along the medial-lateral axis of the femur. With increasing knee flexion, the patella translates laterally within the trochlear groove. The lateral surface of the trochlear groove along with the medial patellofemoral ligament limits the translation of the patella. In cases of trochlear dysplasia, the sulcus angle of the trochlear groove is decreased and the shallow groove fails to provide lateral constraint, resulting in increased translation and patellar instability [87, 104]. Increased lateral patella translation is often found in individuals with patellofemoral pain [55]; however, it has also been reported that there are large overlaps in the amount of translation

Fig. 4 **a** Patella alta refers to a “high-riding” patella. **b** Trochlear groove depth is commonly quantified using the sulcus angle or lateral trochlear inclination (LTI). **c** The tibial tuberosity-trochlear groove distance is used to quantify misalignment of the extensor mechanism due to a lateralized patellar tendon insertion on the tibia

observed across the groups, which brings into question the mechanism of pain. A reduction of the contact area and an increase in joint contact stress are often related to patella translation [105]; however, the variation in translation required to achieve this reduction in contact may be associated with individual bone morphology. The vastus medialis attachment to the patella has been shown to affect patella translation by providing a medial force [85, 106] or affect patella rotation [107].

Patellar tilt is the rotation along the axis following the patella ridge with respect to the long axis of the femur. Patellar tilt has also been associated with a decrease in the joint contact area and patellofemoral pain. Similarly, passive and active structures influence the tilt of the patella. The activation ratio between the vastus lateralis and the vastus medialis is correlated with patellar tilt in a maltracking pain population [108]. Patellar tilt has also been associated with a tight lateral retinaculum [109] and decreased sulcus angle [110].

Anatomical

Geometry

Articular geometry varies widely, even within a healthy population [111], and can influence the function and pathology of the patellofemoral joint. The most commonly studied geometric feature is the trochlear groove depth (Fig. 4b), which can range from being deep to shallow or flat, with an extremely shallow trochlea being referred to as dysplastic. The trochlear groove is often shallower in those who experience patellofemoral pain [112], patellar instability [86, 87, 92, 113, 114], patellofemoral cartilage lesions [88], and patellofemoral osteoarthritis [115–117]. Other geometrical features, such as the patellar-to-trochlear width ratio or patella volume, have been associated with patellofemoral pathology [9, 118], but have not been as widely studied. The association between trochlear groove depth and pathology may be a result of its effect on joint function. A shallower trochlear groove has been associated with increased lateral translation and the tilt of the patella in vivo [113, 119–122]. The correlations in these observational studies provide valuable insight into the importance of morphology to the patellofemoral joint. However, they are unable to confirm causal relationships.

In an in vitro experiment, Van Haver et al. [31] used 3D printing to model the four Dejour classes of trochlear dysplasia [123] and directly assess the effect on joint biomechanics. They confirmed that trochlear dysplasia results in increased lateral tilt and translation of the patella. They were also able to show that dysplasia can increase patellofemoral cartilage contact pressure, decrease contact area, and reduce lateral mechanical joint stability. Using a finite element model, Fitzpatrick et al. [124] created models with sulcus angles ranging from 124 to 181° and simulated five static knee flexion

angles. Again, a shallower trochlear groove caused increased tilt and translation of the patella.

In our work, we have combined statistical shape modeling with musculoskeletal simulation to explore the shape-function relationship in the patellofemoral joint [125]. Statistical shape modeling provides objective quantification of major sources of geometric variability in three dimensions and overcomes reliability issues of 2D measures [126, 127]. We found that the trochlear groove depth was the second largest source of variation in a group of 14 asymptomatic knees. Musculoskeletal simulation allows the shape-function relationship to be investigated during dynamic activities. Using this approach, we found that during overground walking, a shallower trochlear groove caused increased lateral tilt and translation of the patella. Additionally, force in the medial patellofemoral ligament increased and cartilage contact pressure increased and shifted laterally on both the patella and trochlea.

There has been good agreement among these different approaches in terms of the effect of trochlear groove depth on patellar kinematics (Fig. 5). While each approach has limitations, the consensus among them provides strong evidence of the causal relationship between shape and function. This is important when comparing patients with patellofemoral pain with controls because it implies that patients should be shape matched to the controls to account for the wide range of subject-specific optimal function that is dependent on shape.

Ligaments

Generalized joint laxity and patella hypermobility occur more frequently in those who have experienced a patella dislocation than in the general population [129, 130]. This indicates the importance of ligament laxity to patellofemoral stability. Soft tissue plays an important role in stabilization of the patella, especially at full knee extension when the patella tends to be superior to the bony restraint of the trochlear groove [131].

The medial patellofemoral ligament (MPFL) is thought to be the most important provider of passive lateral stability for the patella. This tissue originates on the superior medial border of the patella and inserts on the medial femoral epicondyle. The length and width of the MPFL vary greatly ranging from 45 to 65 mm and from 3 to 30 mm, respectively [132]. In in vitro studies, the MPFL has been found to provide 50–60% of the restraint to lateral displacement of the patella for small flexion angles [26, 131, 133, 134]. Additionally, the MPFL is often ruptured during patella dislocation [135] and repair has been successful in minimizing redislocation [136].

An overly lateral insertion of the patellar tendon on the tibia is also thought to be one of the major determinants of patellar instability [87]. This is frequently quantified as the medial-lateral distance between the tibial tuberosity and trochlear groove (TT-TG; Fig. 4c) and tends to be larger in those with unstable patellae [94, 95, 137]. In vitro and multibody

Fig. 5 Effect of trochlear groove depth on patella tilt and translation, measured by the bisect offset. Biyani et al. [113] and Powers [119] measured patellar kinematics and geometry in vivo using dynamic MRI for knee flexion-extension. Elias et al. [128] simulated a squat using participant-specific multibody models. Fitzpatrick et al. [124] directly modified the sulcus angle, as well as three other parameters, in a finite element model. Clouthier et al. [125] manipulated the trochlear groove in a musculoskeletal model using statistical shape modeling and simulated overground walking

simulation studies have shown that medialization of the patellar tendon tibial insertion shifts the patella medially, decreases cartilage contact pressure, and increases lateral stability [35, 128, 138, 139] and that lateralization has the opposite effect [40]. The location of the patellar tendon tibial insertion has been shown to provide different levels of patellar stability depending on the trochlear groove shape [140].

Muscles

The patella transfers muscle force from the quadriceps muscles to the tibia via the quadriceps tendon and the patellar tendon. The base of the patella serves as the insertion location of the quadriceps tendon. Additionally, the vastus lateralis and vastus medialis have attachments to the patella on its lateral and medial borders, respectively. Muscle weakness in the vastus medialis has been previously hypothesized as a cause for variations in patella mechanics related to PFP [141]. Delayed activity between the vastus medialis and the vastus lateralis has been proposed as another influence on patella stability [61, 142]; however, there remains controversy as to whether all comparison groups are affected [143]. Rathleff et al. [144] showed that individuals with PFP have larger impairments in neuromuscular knee control. Current theories suggest that hip muscles cause femur rotation, which alters patella pressure by rotating the femur beneath the patella and altering the line of action of quadriceps muscles originating on the femur [21, 145, 146].

Lower-limb musculoskeletal models may require more accurate depictions of patient-specific muscle lines-of-action and muscular wrapping to better represent muscle force directions. The relationship between bone morphology and muscle activation is important. Increased patellofemoral joint stress may occur from decreased or increased muscle activation altering patella mechanics. Further investigation is required to determine whether patellofemoral pain can be associated with

muscle weakness preventing the patella from following its ideal kinematic path or increased muscle force and activation in attempts to correct unideal patella kinematics.

Future Directions

Future research in patellofemoral mechanics will include improving modeling techniques with experimental validations and expanding areas that have previously been limited as a result of inadequate technologies and experimental design. One area for further research is to examine patella biomechanics for a larger range of activities. Patellar tracking differs between static and dynamic assessments [46] and between weight-bearing and non-weight-bearing knee flexion/extension [51]. It is likely that patellar function varies with task, yet constraints in imaging modalities have limited the movements that could be measured previously. Altered tibiofemoral joint kinematics in all three planes influences the patellofemoral contact area [147, 148], emphasizing the importance of 3D experiments and modeling. Improvements in biplanar videoradiography technology and tracking algorithms may provide an opportunity for investigation of a greater variety of tasks.

It is also important to understand the interaction between factors affecting patellofemoral joint function. Patient-specific factors do not vary in isolation, and therefore, examining how factors can change complementarily to each other or affect other factors is important to expand our knowledge of patella function. Fitzpatrick et al. [124] performed 100 finite element simulations where trochlear groove depth, patella alta, tibial tuberosity-trochlear groove distance, and femoral anteversion were simultaneously varied for static knee postures. They found that varying these factors simultaneously resulted in the greatest changes to patella tilt, translation, and contact force, highlighting the complex interactions among these factors. Furthermore, studies of

how factors interact in terms of treatment outcomes can help provide the knowledge necessary to treat patellofemoral pathologies on a patient-specific basis.

Validation of musculoskeletal models is a significant barrier to clinical acceptance of simulation results. Patellar kinematics can be compared with in vivo measurements and muscle activations can be compared with EMG. These can provide reassurance that a model can make realistic predictions. However, the strength of simulation is that it can be used to answer questions that cannot be addressed through in vivo or in vitro approaches. Therefore, we should not limit their use to situations that can be directly validated. Results should be interpreted in light of the assumptions of the model, but musculoskeletal simulation provides a powerful platform that can be used to examine a large range of scenarios to contribute to our understanding of how patient-specific factors affect patellofemoral function and contribute to pathology and high-light potential treatment options and preventative measures.

With new findings from modeling results and advanced imaging, previous hypotheses about patella function and pain causes can be formally tested. We are finding that specific bone geometry results in an optimized patella location based on joint congruence corresponding to joint morphology. An example of one case had the optimized patella location laterally situated with a shallow trochlear groove, which was previously thought to be a negative outcome since laterally displaced patellae tend to have higher joint contact pressure. Animal and in silico models have shown the both underloading and overloading the cartilage can be a cause of degradation [149, 150]. Increased pressure is often associated with increased congruence which may add greater stability within the joint. Further research will provide information on joint congruence and its relationship with patellofemoral pressure, patellofemoral pain, and whether these are mutually exclusive.

Moving forward, we believe combining in vivo experiments with simulation provides a means to draw strong and relevant conclusions about patellofemoral function and pathology. This could be done by determining factors that differ between healthy and pathological populations in vivo and then examining this variability in a controlled manner through simulation. By directly manipulating the relevant factors according to the measured variability, it would be possible to establish causality and understand how the differences can contribute to the pathology and its treatment. More prospective studies that control for shape variations are necessary to isolate possible causations of malfunction in the patellofemoral joint.

Conclusions

The existing literature provides insight into how elements of the patellofemoral joint function. Comparisons of people who

experience patellofemoral pain or patellar instability with asymptomatic populations point to variables of interest that may contribute to dysfunction. A better understanding of the way in which these factors affect patellar kinematics, cartilage contact, and joint stability has been gained through in vitro experiments and computer simulation. However, there remains more to know and patellofemoral pathologies are still considered challenging to treat [151, 152]. Future work should aim to integrate in vivo, in vitro, and simulation approaches to expand our knowledge of how patient-specific factors affect patellofemoral joint function and contribute to pathology and treatment.

Compliance with Ethical Standards

Conflict of Interest Mitchell Wheatley, Michael J. Rainbow, and Allison Clouthier declare that they have no conflict of interest.

Human and Animal Rights and Informed Consent This article does not contain any studies with human or animal subjects performed by any of the authors.

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