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Predicting Three-dimensional Patellofemoral Kinematics from Static Imaging-Based Alignment Measures

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Abstract

Patellofemoral pain syndrome causes significant discomfort and disability among much of the general population. Despite recent breakthroughs in dynamic three-dimensional imaging technologies to assess pathological patellofemoral motion, such tools remain costly for clinical diagnostics applications. Thus, this study investigated whether three-dimensional patellofemoral kinematics could be predicted from routine two-dimensional static measures of patellofemoral joint alignment quantified from magnetic resonance imaging (MRI) data acquired in full knee extension. Twenty-six volunteers clinically diagnosed with patellofemoral pain (19F/7M, 25.9±11.1 years) and twenty-six control subjects (19F/7M, 25.3±7.7 years) were included in this IRB-approved study. Static three-dimensional sagittal T1-weighted gradient recall echo and dynamic MRI scans were acquired. For the dynamic image acquisition, subjects cyclically flexed and extended their knee (at 30 cycles/minute) while a full cine-phase contrast MRI set (24 times frames of anatomic images and x, y, and z-velocity images) was acquired. From these data, static measures of patellofemoral alignment and three-dimensional patellofemoral kinematics were derived. Single and multiple regressions between static and kinematic variables were evaluated. Although shown reliable, the static MRI measures could only partially predict patellofemoral kinematics, with r^2-values ranging from 16%-77%. This makes it imperitave that the current precise, accurate, 3D, dynamic imaging techniques be translated into clinical tools.

INTRODUCTION

Patellofemoral pain syndrome (PFPS), defined as long-term idiopathic anterior knee pain, affects approximately 14-17% of the young active population[1]. It is theorized to be caused by a force imbalance at the knee[2] and/or abnormal bone shape[3,4] that results in pathological patellofemoral (PF) kinematics (pathomechanics)[5,6]. It is believed that such pathomechanics ultimately lead to increased PF joint stress and pain[7,8]. Although PFPS is typically exacerbated by dynamic activities requiring active quadriceps loading, clinical evaluation of the knee relies primarily on 2D static imaging-based alignment measures and other clinical markers, because *in vivo* 3D dynamic PF kinematic measures, quantified during activities requiring active muscle control, are not yet clinically available.

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Imaging technologies have continuously evolved in order to more accurately and quantitatively evaluate the PF joint (as well as other joints) under loaded dynamic conditions. The first imaging studies evaluating PF alignment[9,10] were limited to 2D static imaging of the anterior aspect of the PF joint. The required flexed knee posture of roentgenographic imaging resulted in high false negative rates[9,11]. In the late 1980s, computed tomography (CT) and magnetic resonance imaging (MRI) became available to study PF alignment in a fully-extended position[12-14]. These technologies confirmed that the femoral influence on the patella is lessened as the knee enters terminal extension[11],,[chk] making malalignment most evident in this range[15]. Thus, for interventional decisions, particularly invasive ones, CT and MRI are often favored over x-ray[11,16,17]. Later, cine-MRI was used to quantify 2D PF kinematics during volitional leg extension exercises[6]. The ability to image the PF joint during an activity requiring muscle control[7] allowed PF pathomechanics to be diagnosed that would have otherwise "been overlooked with static examination techniques"[6]. Advances in imaging technologies have enabled the 3D PF kinematics contractions to be quantified during either dynamic exercise requiring quadriceps activity[18,19] or statically with active quadriceps[20]. These 3D studies added to previous findings by documenting kinematic differences beyond the classic axial-plane measures. For example, the results from the Sheehan et al.[18] study supported the long standing association between patella alta and PFPS[21-23], as well as introduced other kinematic variables that may be key components in the development of PFPS, such as pathological PF valgus and flexion[19]. A recent 3D study[24] demonstrated that PF flexion (measured during active extension) could discriminate with 100% accuracy the presence of idiopathic long-term anterior knee pain in a cohort of individuals diagnosed with cerebral palsy.

Many of these novel dynamic imaging methodologies have remained research tools, leaving the simpler static 2D imaging-based measures as the primary methods for diagnosing PF joint pathology[9,10,25]; even though PF pain is typically experienced during dynamic events requiring high quadriceps loading. The validity of predicting 3D PF kinematics from static, 2D, imaging-based measures remains unestablished. Quantifying the relationship between the static and dynamic states of the PF joint may offer improved diagnostics and eventual treatment of suspected patellar maltracking[26]. Thus, the purpose of this study was to determine if 3D PF kinematics, acquired *in vivo* during a volitional leg extension exercise, could be predicted from 2D, static, image-based measures of PF alignment. A secondary aim was to evaluate the reliability of these static measures.

METHODS

For this retrospective, IRB-approved study two cohorts were established from an existing database (controls [n=90] and subjects diagnosed with PFPS [n=54]). Each subject signed informed consent at the time of enrollment. To be included within the PFPS cohort the subject had to be diagnosed with idiopathic anterior knee pain (> 6 months duration), have no history of lower limb surgery, and have positive clinical signs of patellar maltracking[5]. Subjects included within the control cohort had no current or past history of lower leg pain, injury, surgery, or pathology. All included subjects had a previously acquired dynamic cine phase contrast (CPC) MRI dataset and a 3D static sagittal-plane MRI acquisition with the knee in full extension. Static full extension was defined using visual inspection of the

clinical knee angle (created by lines connecting the hip, knee, and ankle joint centers in the sagittal plane). As a check on this visual alignment, the knee was only considered to be in static full extension if the knee angle, as measured in the sagittal GRE images (Fig 1: the 2D acute angle between t_y and f_y), was within two standard deviations a previous control cohort. For this previous cohort, subjects were placed in full extension, using the same visual alignment procedure, prior to acquiring a sagittal plane 3D GRE image of the knee. If both knees qualified for the study, a single knee was randomly selected (control) or the more impaired knee (PFPS cohort) was selected for inclusion. The final study cohort consisted of 26 controls (19F/7M, 25.3±7.7 years, 166.1±9.3cm, 61.1±11.6kg) and 26 subjects diagnosed with PFPS (19F/7M, 25.9±11.1 years, 165.4±9.1cm, 62.2±10.6kg).

Subjects lay supine in an MRI scanner (1.5 T; GE Medical Systems, Milwaukee, WI; or 3.0 T; Philips Electronics, Eindhoven, The Netherlands) during static and dynamic image acquisition. For static scanning, the knee and hip were fully extended with the mid-patella at the center of a knee coil. The ankle was held in the anatomical neutral position using cushion blocks. Subjects were instructed to keep their muscles relaxed while static 3D sagittal T1-weighted high resolution ($1mm^3$) gradient recall echo (GRE) images were acquired (TR=11, TE=5.1, 70-84 slices, Flip Angle = 15°). Image acquisition ranged from just distal to the tibial tuberosity to just superior to the quadriceps' tendon insertion into the rectus femoris (RF). During dynamic scanning (Fig 1) the subject's knee was placed over a cushioned wedge, slightly flexing the hip and knee. Coupled-phased array coils were supported medial and lateral to the knee using a custom-built coil holder. Subjects cyclically flexed and extended their knee (30 cycles/minute, to the beat of an auditory metronome) while a full CPC image set (anatomic and x-, y-, and z-velocity images over 24 times frames) was acquired in a single sagittal plane. The CPC scanning parameters varied slightly between the 3.0T[27] and 1.5T[28] scanners, but the overall temporal resolution (61.2 msec on the 3T and 73.6msec on the 1.5T) was kept similar. Although the images were acquired in a single plane, the 3D velocity enabled tracking of musculoskeletal points of interest in all three dimensions. A 3-plane axial cine image set (anatomical images only, 24 time frames) was acquired in order to establish anatomic coordinate systems[28].

Seven static measures of interest (Fig 2: patellar tilt angle (PTA), lateral patellar displacement (LPD), anterior/posterior displacement (AP_S), inferior/superior displacement (SI_S), bisect offset (BO), patellophyseal index (PPI), and the RF-Q-angle) were quantified from the 3D GRE images. The subscript "S" denotes static. To quantify these measures, the 3D sagittal GRE images were reconstructed into axial images. Femoral vectors and points of interest were located in axial image at the level of the femoral epicondyle. Likewise, patellar vectors and points of interest were located in mid-patellar axial image. The PTA was quantified as the acute angle between the vectors defining the patellar posterior-lateral border and the femoral posterior edge (Fig 2C). Next, the lateral, anterior, and superior distance from the patellar origin (Po, the most posterior patellar point) to the femoral origin (Fo, the deepest point in the femoral sulcus) was defined as the LPD, AP_S, and SI_S displacement (Fig 2C,F). For the purpose of this study medial, anterior, and superior were considered positive directions. The BO[6] and the PPI[22] were also measured (Fig 2B, E). Lastly, the RF-Q-angle was quantified as the acute angle between two vectors defined by three points (the tibial tuberosity, the anterior-mid-patellar, and the center of RF just

superior to the myo-tendinous junction, Fig 2C,D,F). This measure was created previously to be a more precise measure of the clinical Q-angle. The 2D PF alignment measures were redefined relative to the static femoral coordinate system (Fig 2). This effectively rotated all images such that the posterior edge of the femoral condyles (at the level of the epicondylar width) and the vector bisecting the femoral shaft (in the sagittal plane) were aligned with the image right-left and inferior-superior directions, respectively. This was done to reduce errors from slight variations in subject alignment relative to the magnet[25]. To account for size variations across subjects, LPD, AP_S, and SI_S were scaled by the ratio of the average epicondylar width from a previous control cohort, 76.9 mm[28], to the subject-specific epicondylar width.

PF translation (ML_K, AP_K, SI_K) and rotation (tilt, flexion, and varus rotation) during active extension, was analytically tracked through integration of the CPC data (Fig 1). The accuracy of tracking skeletal kinematics using CPC data is less than 0.5 mm[27,29] (average absolute error). These six kinematic variables were expressed relative to the dynamic femoral anatomical coordinate system (Fig 1)[28], which was defined similiary to its static counterpart. PF orientation (Fig 1) was calculated using a body-fixed xyz-Cardan rotation sequence (flexion, tilt, and varus)[30]. All translations were scaled in an identical manner to the static measures.

Multiple regression analyses were used to establish predictive models for the PF kinematics. To determine which static variables were appropriate as input to the multiple regression analyses, individual linear regressions between each static and each dynamic variable were evaluated. This was followed by a multiple regression analysis (Enter method, SPSS Inc, Chicago, IL, v19), if multiple static variables demonstrated significant linear regressions to a single dynamic variable. The multiple regression model used only those variables. Based on guidelines for correlations[31], a model with an r^2 value of greater than 0.60 was defined as being strongly predictive. The numerous regression models were susceptible to type I error. Therefore, regressions were reported only if they remained significant after their p-values were adjusted using a Bonferroni-type false discovery rate procedure[32]. Although the PF kinematics were determined over a large range of motion, the regressions were focused on the PF kinematics from two knee angles (10°, matching the static position, and 20°), both of which could capture the more unstable PF kinematics near full extension. An *a priori* power analysis determined that 26 subjects per cohort were required to determine significant regression between the RF-Q-angle and ML_K (α= 0.05 and β=0.80), based on a previous study[5]. Intraclass correlation coefficients (ICCs), using a two-way mixed effects model, were computed to evaluate intra- and inter-rater repeatability of the static PF alignment measures across 25 knees (randomly selected from both cohorts). The two raters were blinded to each others measures, as well as their original measures. A p-value < 0.05 was considered as significant.

RESULTS

The majority of 3D PF kinematics could be predicted from static MRI measures, but static lateral patellar tilt best predicted its dynamic counterpart (r^2-value ranged from 0.59 to 0.69, Fig 3 and Table 2). Multiple regression analyses tended to increase the r^2-values (Table 3).

This was particularly true for SI_K in the PFPS cohort, where r^2 increased from 0.38 to 0.77 by adding LPD and RF-Q-angle into the regression. LPD alone could explain 47% ($p < 0.001$) of the variability (PFPS cohort) in ML_K (dynamic lateral shift). This improved to 62% ($p<0.001$) upon advancing to a multiple regression analysis using both LPD and RF-Q-angle. Varus rotation (PFPS and control cohorts) along with ML_K and AP_K (control cohort) could not be predicted by a single or multiple static variables.

The regressions at 20° were typically weaker than at 10° and, thus were not reported. This leads to the likley conclusion that the predictive values worsen with increasing variance between the static and dynamic knee angles.

All static MRI measures showed excellent intra- and inter-rater repeatability, with ICCs ranging from 0.95-0.99 (Table 1). The two cohorts were, on average, very well matched with no significant differences in demographics found between cohorts. The MR based knee angle (10.6°± 3.3° and 10.8° ± 4.2° in the control and PFPS cohorts, respectively) tended to overestimate the clinical knee angle.

DISCUSSION

Recent breakthroughs in 3D dynamic imaging technologies[19,29] have enabled the evaluation of more subtle cases of maltracking[6] and the testing of numerous underlying assumptions governing modern PFPS diagnostics and interventions[19; 28]. As these 3D dynamic imaging technologies have not been clinically available, an often unstated assumption has been that static imaging can be used as a surrogate for quantifying the dynamic state of the joint under active quadriceps control. The current findings demonstrate that static measures of PF alignment, quantified in 2D images acquired with the knee in full extension, can only predict a portion of the 3D PF kinematics during a loaded volitional task.

Although significant linear regressions existed for most kinematic variables, the majority of their variability could not be explained by a single static surrogate. This low predictability is likely due to the fact that the forces on the patella from femoral sulcus begin to lessen as the knee moves into terminal extension. This allows the overall dynamic state of the joint to be more strongly influenced by passive soft tissue and active quadriceps forces in terminal extension[6,14,33]. The interaction between the contact, passive soft-tissue, and active quadriceps forces varies across subjects, further reducing the predictability of the dynamic state. For example, a large portion of the unexplained variance in SI_K may be attributed to its dependence on the nominal lengths and material properties of the patellar and quadriceps tendons, which can vary greatly across subjects, particularly in PFPS[18,34].

The static PTA was the only variable that strongly predicted its dynamic counterpart. This supports a previous study that found strong agreement between excessive lateral tilt measured using a clinical exam and measured using 2D static MRI.[13] The strong relationship between the static PTA and PF dynamic lateral tilt may indicate that quadriceps activity has minimal effect on patellar tilt during terminal extension. Thus, even though the femoral sulcus influence diminishes in terminal extension, it may remain a controlling force for lateral patellar tilt[4]. The regression was likely further strengthened by the large range of

values for lateral tilt across both cohorts. Yet, this range in value is the likely reason why none of the recent 3D PF kinematics/alignment studies with active quadriceps reported significant differences in tilt between the PFPS and control cohorts[18-20].

In a previous study McWalter and colleagues[35] found that the value of a single measure of PF joint alignment, acquired statically at 30° flexion with active isometric quadriceps contraction, could not predict the slope (rate of change relative to knee angle) of 3D PF alignment measures acquired statically at various knee flexion angles in a cohort of healthy volunteers. This inability to predict the rate of change PF alignment may be due to the low reliability (0.58-0.59) in measuring slope using the static methodology reported by McWalter and colleagues[20]. Although two previous studies[18,20] have demonstrated significant differences in the slope of PF alignment/kinematic variables when comparing cohorts of subjects with PF pain to control cohorts, the clinical utility of evaluating slope has not been established. Thus, clinical diagnosis of maltracking remains focused on the value (not slope) of PF aligment/kinematics in terminal extension, as this is where the restraints of the femoral sulcus on the patella are weakest and pathology is typically most evident[6,18,36].

Several studies have investigated static joint alignment measures, yet few studies[17,37-41] have reported the reliability of these measures. The reliability measures from the current study (0.94 – 0.99) were better than previous studies and satisfied the guidline for reasonable clinical validity (ICC>0.90)[38]. Only one previous study evaluating patella alta[37,40,41] reported “good” inter-rater reliabilities (ICC > 0.75[38]). The studies measuring patellar tilt and BO reported good reliabilities for measures acquired during active leg extensions (ICC = 0.90[17]) and passively (ICC=0.76 for BO and ICC=0.86 for PTA)[39].

The primary limitation of this study was that specific static counterparts for dynamic PF flexion and varus rotation were not established, thus the predictability of these dynamic variables remained low. Another possible limitation was that the exercise evaluated was open-chain, whereas PF pain is most often thought to be exacerbated by weight-bearing (closed-chain) activities. Yet, PF pain can be induced during free extension[19], as well as during long term stationary sitting with flexed knees[42]. Also, the exercise used in the current study emphasized the conditions of joint instability by requiring a quadriceps force in terminal extension, where the femoral constraints on the patella are at a minimum and pathomechanics are most evident. Based on the work of Hungeford[43], this required quadriceps force in terminal extension was likely greater than that required at or near full extension in previous studies evaluating PF kinematics during free squats[19,44]. Regardless, this study clearly demonstrated the robustness of using specific static variables, such as the patellar tilt angle, to predict PF joint motion. Future studies should examine additional static surrogates for improved prediction.

In conclusion, static PF joint assessment cannot fully represent the dynamic state of this joint and using such static measures for diagnosis would likely produce false negative diagnoses[7,15,17]. This makes it imperitave that the current precise, accurate, 3D, dynamic imaging techniques be translated into clinical tools in order to improve the diagnosis of PF pathomechanics [7,17,33,45] by providing accurate and precise quantification of PF kinematics,

which expands beyond the axial plane, during a volitional activity requiring quadriceps loading.

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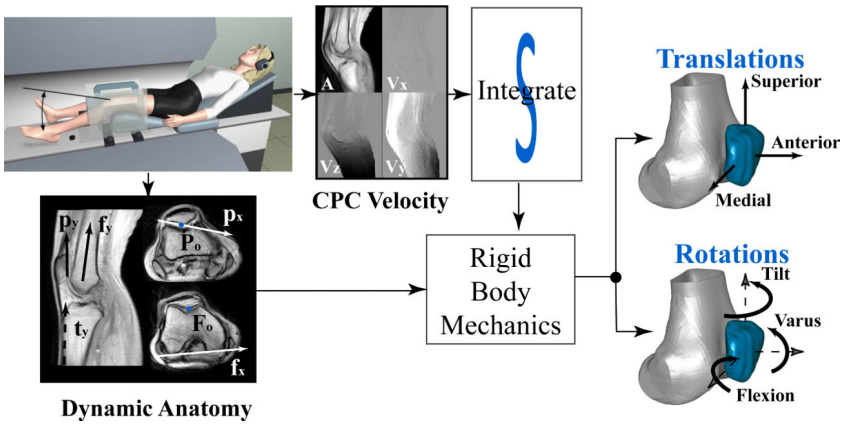

Figure 1. Flow chart for analysis of the dynamic data

The patellar and femoral medial directions (**p**x, **f**x) were parallel to the patellar lateral-posterior edge and the femoral posterior edge. In the mid-patellar sagittal image, the patellar posterior edge and the vector bisecting the distal femoral shaft defined the patellar and femoral superior direction (~**p**y, ~**f**y). The 3D coordinate system was defined such that **pz** = **px** X ~**py** and **py** = **pz** X **px**. The knee angle was the acute angle between **fy** and **ty** (parallel to the tibial anterior edge, in a sagittal image just medial to the tibial tuberosity). The static coordinate system and knee angle used identical definitions. **Abbreviations:** A: anatomic image; Vx, Vy, and Vz: velocity in the x, y, and z directions, respectively; CPC: Cine Phase Contrast; Po and Fo: Origin of the patellar and femoral coordinate systems, respectively.

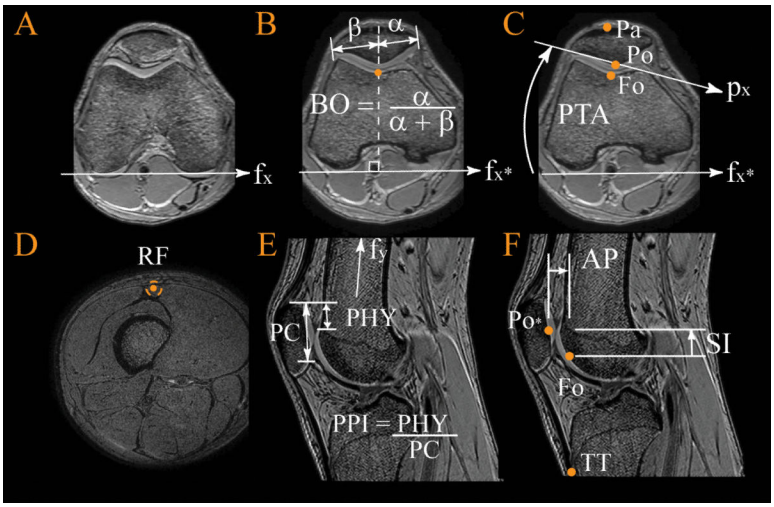

Figure 2. Static Magnetic Resonance Imaging-based Measures of Patellofemoral Alignment

A) fx = the femoral medial direction, which was defined by the most posterior points on the medial and lateral condyles at the level of the epicondylar width. B) BO = bisect offset; fx* a representation of the femoral medial axis as defined in A; α and β = the shortest distance from the most medial and lateral patellar points, respectively, to a line perpendicular to **fx** that passes through Fo, propagated to the mid-patellar plane. C) PTA = patellar tilt angle; Po and Fo = origin of the patellar and femoral coordinate systems; px = the patellar medial direction; Pa = anterior patella. D) RF = rectus femoris. E) PPI = patellophyseal index; PC = length of patellar articular cartilage; PHY = height of patella above anterior physeal line. F) AP = anterior/posterior distance; SI = superior/inferior distance; Po* - a representation of Po, which is defined in C; TT= tibial tuberosity.

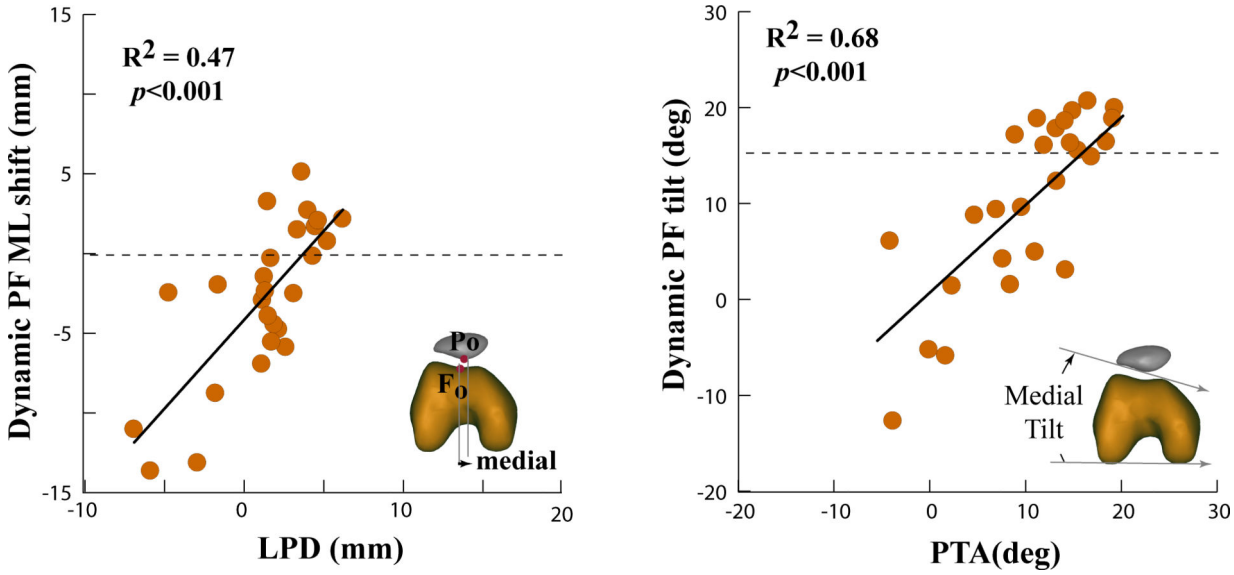

Figure 3. Example Linear Regression Plots for the Patellofemoral Pain Syndrome Cohort

Control averages for the dynamic variables are provided with a dotted line. **Abbreviations**: PTA: patellar tilt angle; ML= the medial-lateral location of the patellar origin relative to the femoral origin; LPD= lateral patellar displacement; deg= degree; mm= millimeter; r^2= coefficients of determination.

Table 1

Intraclass Correlation Coefficients for static Magnetic Resonance Imaging-based Surrogates.

Parameter	PTA	BO	LPD	SI_S	AP_S	RF Q-angle
Inter	0.995	0.985	0.979	0.990	0.948	0.985
Intra	0.993	0.980	0.968	0.992	0.953	0.989

Abbreviations: PTA= patellar tilt angle; BO= Bisect Offset; LPD, AP_S and SI_S= The medial-lateral, anterior-posterior, and superior-inferior location of the patellar origin relative to the femoral origin (static); RF-Q angle= The rectus femoris quadriceps angle.

Table 2

Coefficients of determination (r^2) for the control (C), patellofemoral pain syndrome (PFPS), and combined (All) cohorts

	All	C	PFPS	All	C	PFPS	All	C	PFPS
Translations	**ML_K**			**SI_K**			**AP_K**		
BO	0.27**		0.32						
LPD	0.32**		0.47**	0.22**		0.43**			
AP_S							0.18		
SI_S				0.43**	0.48**	0.38**			
PPI				0.28**	0.36**	0.26**			
PTA	0.23**		0.31						
RF-Q	0.19		0.35			0.30			
Rotations	**Flexion**			**Tilt**			**Varus**		
BO				0.49**	0.27	0.54**			
LPD				0.23**		0.29			
AP_S									
SI_S	0.26**	0.42**							
PPI	0.23**	0.41**							
PTA				0.69**	0.59**	0.68**	0.17		
RF-Q				0.16		0.37			

Nonsignificant regressions are not shown. For the full regression equations, see supplemental material.

Abbreviations: BO = Bisect Offset; LPD, AP_S, and SI_S = The static medial-lateral, anterior-posterior, and superior-inferior location of the patellar origin relative to the femoral origin (static); PPI = patellophyseal index; PTA = patellar tilt angle; RF-Q angle = The rectus femoris quadriceps angle; ML_K, SI_K, AP_K = The medial-lateral, superior-inferior, and anterior posterior location of the patellar origin relative to the femoral origin, measured during active extension at 10° knee angle.

** Indicates $p < 0.001$, otherwise $p < 0.05$.

Table 3

Multiple Regression Equations.

Dynamic Variable	Multiple Regressions	R^2	*p*- value
All			
ML_K	= −4.61 + 0.61*LPD −0.15*PTA	0.38	<0.001
SI_K	= 13.48 −0.62*SI_S −0.38*LPD-0.14*RF_Q	**0.53**	<0.001
Flexion	= 8.435 −0.40*SI_S −0.57*PA_S	0.36	<0.001
PFPS Cohort			
ML_K	= −4.14 + 1.52*LPD −0.13*RF_Q	**0.62**	<0.000
AP_K	= 2.34 −0.50*AP_S −0.001*RF_Q	0.38	0.001
SI_K	= 16.80 + 0.51*SI_S −1.45*LPD +0.15*RF_Q	**0.77**	<0.001
Flexion	= 7.553 −0.307*SI_S −0.540*AP_S	0.32	0.003
Control Cohort			
Flexion	= 20.816 −0.514*SI_S −0.383*PTA	**0.65**	<0.001

Multiple Regressions listed if it increased the predictability more than 5 percentage points from the linear regression.

Abbreviations: ML_K, SI_K, AP_K = The medial-lateral, superior-inferior, and anterior-posterior location of the patellar origin relative to the femoral origin measured during active extension at 10° knee angle; LPD, AP_S, and SI_S = the static medial-lateral, anterior-posterior, and superior-inferior location of the patellar origin relative to the femoral origin; PTA = patellar tilt angle; RF-Q angle = The rectus femoris quadriceps angle.