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Impact of Femoral and Tibial Torsion on Patellofemoral Loading in Individuals With Patellofemoral Instability

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ABSTRACT

Patellofemoral stability is affected by several morphological factors including torsional alignment. To elucidate the impact of factors responsible for the stability of the patellofemoral joint, biomechanical research utilizes the analysis of joint contact forces. At present, there is a paucity of modeling-based research examining the influence of lower limb torsion on patellofemoral joint loading in individuals with patellofemoral instability. The objective of this study was to investigate the impact of the femoral version and tibial torsion on the patellofemoral joint loading. Musculoskeletal simulations were conducted based on 3D motion capture data of 40 individuals with patellofemoral instability using OpenSim. We created three models with different lower limb torsions for each participant: (i) generic torsion, (ii) personalized lower limb (femur and tibia) torsion, and (iii) isolated personalized femoral version model. We correlated femoral version and tibial torsion to differences in patellofemoral joint loading, muscle forces, and lever arms between models. Tibial torsion correlated to differences in mediolateral patellofemoral force ($\rho = 0.39$), whereas the femoral version showed no significant correlation to the differences in mediolateral patellofemoral force ($\rho = 0.01$). Notably, when neglecting individual tibial torsion, the femoral version correlated to differences in mediolateral patellofemoral force ($\rho = 0.65$). The femoral version can increase the lateralizing force on the patella, but this effect diminishes when addressing whole lower limb torsion in musculoskeletal simulations. Studies investigating solely the femoral version should, therefore, be interpreted with caution. Our findings underscore the necessity of evaluating whole lower limb torsion for a comprehensive assessment of its impact on patellofemoral stability and planning treatments.

Level of Evidence: Retrospective cohort study, Level III.

1 | Introduction

Patellofemoral instability is a condition that is prevalent among adolescents and children. It can result in patella dislocations, functional decline, and pain [1, 2]. Several morphological factors have been identified as affecting the stability of the patellofemoral joint. These include abnormal patella height, a shallow femoral trochlea, insufficient function of the medial patellofemoral ligament, and excessive torsions of the femur and tibia [3, 4]. Consequently, surgical interventions are employed to correct deformities that are related to these structures [1]. In addition to procedures like reconstruction of the medial patellofemoral ligament or lateral release, derotation osteotomies are performed at the femur [5–7] and tibia [8, 9] to enhance the stability of the patellofemoral joint. Currently, derotation osteotomies are mainly performed when specific threshold values for the femoral version or tibial torsion are reached. These threshold values are based on literature or the individual preferences of surgeons [5]. As these surgeries are more complex and more invasive than soft tissue procedures like medial patellofemoral ligament reconstruction, it is crucial to possess a comprehensive understanding of their efficacy and indications.

Biomechanical cadaver studies offer theoretical insights into the impact of femoral version and tibial torsion on patellofemoral joint alignment and loading. Cadaver studies showed that an excessive femoral anteversion results in an increase in lateral facet patella pressure [10, 11]. Another study demonstrated that increased femoral anteversion leads to lateralization of the patella [12]. Higher external tibial torsion was associated with a more lateralizing patellofemoral force [10, 13]. However, a limitation of these studies is that muscle forces were applied in an isometric way, which neglected the impact of subject-specific movements and muscle activation on patellofemoral kinematics and loading.

Musculoskeletal simulations enable the investigation of joint reaction loads in relation to individual morphology and movement patterns [14, 15]. In recent years, tools have been developed that facilitate the personalization of lower limb torsions in musculoskeletal models [14, 16, 17]. Investigations based on these tools demonstrated the impact of lower limb torsion on hip and knee joint loads in a healthy person [15] and an elderly individual with a knee implant [17]. Moreover, studies have employed torsion-informed models to demonstrate elevated knee joint loading in individuals with idiopathic rotational deformities [14, 18, 19]. Wheatley et al. [20] found a positive correlation between femoral version and patellofemoral joint loading based on motion capture data from a single healthy participant. However, all these studies only addressed femoral version, and no study has quantified the impact of lower limb torsion, including femoral version and tibial torsion, on joint loads in individuals with patellofemoral instability. Considering that each patient presents a subject-specific torsion at both long bones (femur and tibia), but all previous studies neglected tibial torsion, it is of utmost importance to evaluate the influence of neglecting tibial torsion on patellofemoral joint loads.

The primary objective of this study was to examine the influence of the femoral version and tibial torsion on patellofemoral joint loading in individuals with patellofemoral instability. Based on the findings of previous studies [10, 12, 13, 20], we hypothesized that increased femoral anteversion as well as increased external tibial torsion would increase the lateralizing force exerted on the patellofemoral joint. A second objective of this study was to evaluate how neglecting tibial torsion, as done in previous cadaver [12] and musculoskeletal modeling studies [18, 20], affects our findings. A deeper understanding of how lower limb torsion affects knee joint loading could enhance clinical decision-making and improve the treatment of patellofemoral instability.

2 | Methods

We conducted musculoskeletal simulations based on a retrospective data set derived from our clinical database, which included 3D gait analysis data and rotational magnetic resonance images (MRI) from individuals with patellofemoral instability. The local ethics committee (IRB00002556, 34-181 ex 21/22) approved this study. Due to the retrospective nature of the study, the need for informed consent was waived.

2.1 | Participants

As inclusion criteria, we determined available 3D gait analysis data as well as rotational MRI images and recurrent patellofemoral instability, which was defined as at least three patella dislocations. Individuals with additional leg injuries or neurological diseases were excluded from the study. Based on these criteria, we included 40 participants in this study, who were referred to gait analysis between 2010 and 2022.

2.2 | Retrospective Data

Three-dimensional gait analysis data were collected using a ten-camera infrared-based motion capture system (Vicon Motion Systems, Oxford, UK) operated at a sampling rate of 120 Hz. The data was collected using either the plug-in gait marker set [21, 22] or the modified Cleveland marker set [23]. The hip joint centers were calculated based on the pelvis width [24] according to a modified version of the Harrington equation [25]. The ground reaction forces were captured synchronously to the marker data using four force plates (Advanced Mechanical Technology Inc., Watertown, MA, USA) operated at a sampling rate of 1080 Hz. All participants walked barefoot at a self-selected walking speed.

Rotational MRI scans were recorded for the purpose of planning surgical interventions and were subsequently made available for retrospective analysis. The degree of femoral version was quantified using rotational MRI according to the methodology proposed by Schneider et al. [26] and Guenther et al. [27]. Tibial torsion was measured according to Rosskopf et al. [28].

2.3 | Musculoskeletal Simulations

We created three models for each participant (Figure 1). The first model was a generic-scaled version of the modified Rajagopal model (generic-torsion model) [29]. The modified Rajagopal model represents the Rajagopal model [30] incorporating the more detailed multi-compartment knee joint of the Lerner model [31]. The generic torsion model had a femoral version of 21° and a tibial torsion of 24°. The second model was based on the same generic model but included personalized femoral version and tibial torsion values derived from MRI measurements (torsion-informed model). The third model was based on the same generic model as the prior two, but only the femoral version was personalized based on the MRI measurements (femoral-version model). The Torsion Tool [16] was used

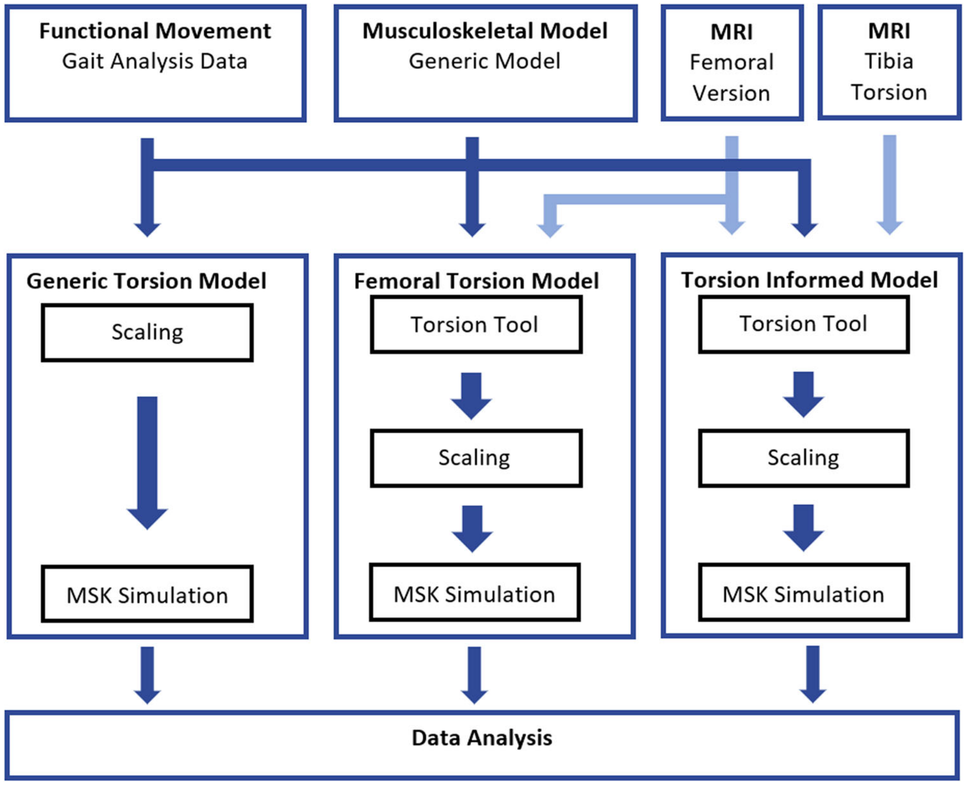

FIGURE 1 | Schematic representation of the simulation workflow. Gait analysis data and the generic modified Rajagopal model were used as input for all three models. MRIs were additionally used as input data for the femoral-version and torsion-informed models. For each participant, three models were created: (1) the generic-torsion model was scaled to each participant's anthropometry, and musculoskeletal simulations were conducted. (2) The femoral-version model was personalized for the femoral version, and scaled and musculoskeletal simulations were performed. (3) The torsion-informed model was personalized for the femoral version and tibial torsion, and scaled and musculoskeletal simulations were performed. MRI = magnetic resonance images, MSK = musculoskeletal.

to personalize the femur and tibial torsion based on the MRI measurements in the second and third models. The Torsion Tool modifies the torsion of the femur and tibia uniformly along their respective shafts. All muscle attachment points were modified to remain in the same position on the corresponding bone. When modified bones were integrated in the models, the knee and hip joint remained in the same neutral position, while the foot progression and the position of the trochanter major relative to the hip joint changed (Figure 2).

Each model was scaled to the corresponding participant's anthropometry using the three-dimensional motion-capturing data [32]. Due to the limited number of foot markers, the metatarsophalangeal and subtalar joints were locked in all models, similar to previous studies [33, 34]. Maximum isometric muscle forces were scaled to each participant according to the squared body weight [35, 36].

Musculoskeletal simulations were conducted for each participant and the corresponding three models using OpenSim 4.4 [37] (Figure 1). Inverse kinematics and inverse dynamics were used to calculate joint angles and moments, respectively. Static optimization was utilized to estimate muscle activations and forces. Based on these results, joint reaction loads were calculated [38]. We compared knee joint loading, muscle forces, and hip muscle lever arms during walking between the generic-torsion model and torsion-informed models from each participant.

2.4 | Data Analysis

For each participant, at least three gait cycles of the affected side were simulated and averaged. For individuals presenting bilateral patellofemoral instability, one leg was randomly selected for the simulations. Simulation results were time-normalized to the gait cycle duration (0%–100% gait cycle). Patellofemoral joint contact forces and muscle forces were normalized to the body weight (BW = body mass × g) of each participant. Patellar dislocations tend to occur in the early stages of knee flexion and under contraction of the quadriceps muscle [39]. Given that these conditions persist at the end of the loading response phase, we focused our analysis on the first 20% of the gait cycle. All analyses were performed on the following three patellofemoral loading parameters obtained during this phase of the gait cycle: (i) maximum mediolateral patellofemoral force, (ii) root mean square (RMS) mediolateral patellofemoral force, as well as (iii) maximum resultant patellofemoral joint reaction force. Individuals with patellofemoral instability are known to walk with a varying gait pattern [40, 41], which can lead to alterations in the patellofemoral loading [42]. To minimize confounding factors (e.g., variability in gait patterns) for each parameter under investigation, we calculated the difference in joint reaction forces between the generic-torsion model and the corresponding personalized torsion model (either torsion-informed or femoral-version model) (Figure 3).

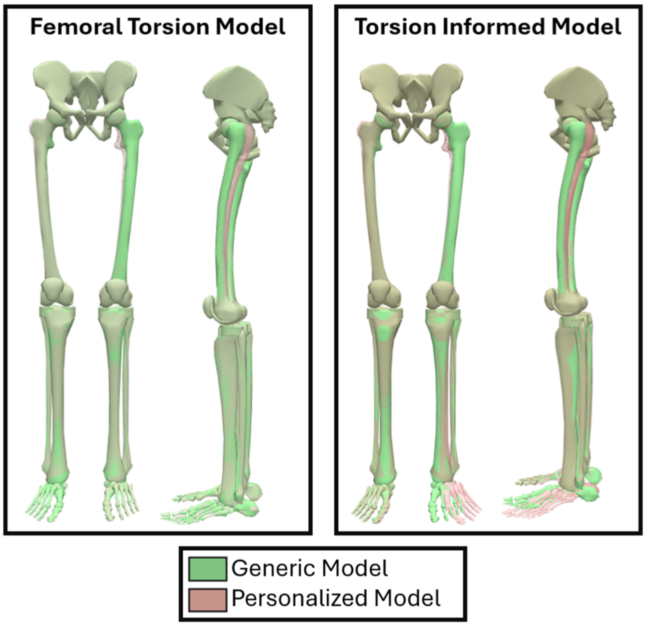

FIGURE 2 | Alignment of the femur and tibia before and after personalization. This figure presents how the Torsion Tool changes the shape and alignment of the femur and tibia (red) compared to the generic torsion model (green). The left side represents the personalization for the femoral torsion model, while the right side represents the personalization for the torsion-informed model. The personalized models represent one participant with a 36° femoral version and 38° tibial torsion.

Correlation analysis (detailed below) between morphological parameters and simulation results (differences between models) were performed using Matlab 2022a (Mathworks Inc., Natick, MA, USA) and SPSS 29 (IBM, New York, USA). To ensure the robustness of the analysis, Spearman's correlations were employed. The α level of significance was set to 0.05.

The following correlation analyses were performed to address our hypotheses. For our primary objective, we investigated the effect of lower limb torsion (femoral version and tibial torsion) on patellofemoral joint loading. For this investigation, the differences in patellofemoral force between the generic-torsion and torsion-informed models were correlated to the femoral version and tibial torsion values. Furthermore, we correlated tibial torsion and femoral version to differences in hip rotation, hip flexion, knee progression, and knee flexion angle between the generic-torsion and torsion-informed models. For our second objective, we evaluate how neglecting tibial torsion affects our findings. For this investigation, the differences in patellofemoral force between the generic-torsion and femoral-version models were correlated to the femoral version values. Afterward, we compared the findings from our primary with our secondary investigation to evaluate how neglecting tibial torsion affects the conclusion of our study.

To investigate a potential relationship between femoral version, tibial torsion, and patellofemoral loading, we performed multiple linear regression analysis for the difference in maximum patellofemoral loading between the generic-torsion and torsion-informed model.

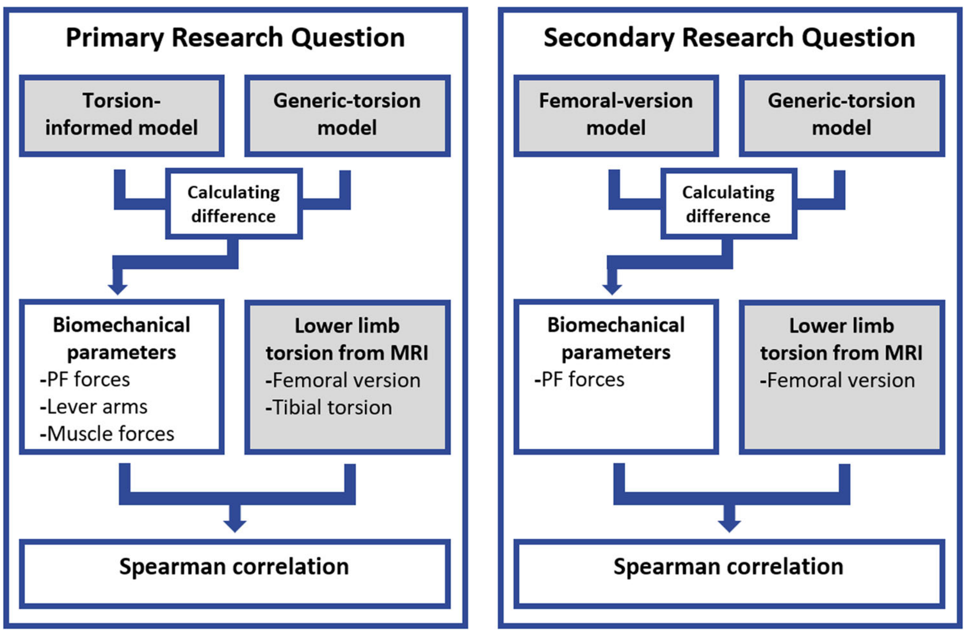

FIGURE 3 | Schematic representation of the data analysis. For the primary research question, the difference between the torsion-informed and generic-torsion model for patellofemoral joint reaction forces, hip muscle lever arms, and muscle forces were calculated. In a second step, Spearman's correlations were calculated between those biomechanical parameters and femoral version and tibial torsion. For the secondary research question, the difference between the femoral-version and generic-torsion model for patellofemoral joint reaction forces were calculated. In the second step, Spearman's correlations were calculated between patellofemoral joint reaction forces and the femoral version. Calculating differences between simulation results allows to investigate the impact of a specific parameter on different biomechanical parameters without confounding factors such as a person's gait pattern. MRI = magnetic resonance images, PF = patellofemoral.

Additionally, the patellofemoral forces calculated in the torsion-informed model were correlated to the tibial torsion and femoral version. Further, we performed a multiple linear regression analysis for the maximum patellofemoral loading in the torsion-informed model using the femoral version and tibial torsion as independent variables. These results are presented in Supporting Information 1.

Previous simulation-based studies demonstrated that the femoral version has an influence on the lever arms and muscle forces of several hip muscles [15, 18, 20]. Consequently, we furthermore investigated the impact of femoral version and tibial torsion on hip muscles' lever arms. We correlated the differences in mean lever arms and RMS of muscle forces normalized to the participant's body weight between the generic-torsion model and the torsion-informed model to femoral version and tibial torsion values.

TABLE 1 | Demographic data.

Demographic data	
Age [years]	15.8 (1.6)
Sex [male/female]	3/37
Height [m]	1.68 (0.07)
Body weight [kg]	59.8 (10.6)
Body-mass-index	21.3 (3.6)
Tibial torsion [°]	35.6 (7.8)
Femoral version [°]	25.0 (10.8)
TT-TG [43] * [mm]	16.5 (4.9)
CDI [44] *	1.18 (0.15)
Dejour [45] * [A/B/C/D]	5/13/10/5
Walking speed [m/s]	1.19 (0.14)
Walking speed normalized to height [1/s]	0.71 (0.08)

Note: Parameters are presented as follows: mean (standard deviation). Parameters marked with an asterisk were only available for 33 participants.
Abbreviations: CDI = Caton-Deschamps index, TT-TG = distance between tibial tuberosity and trochlea groove.

3 | Results

The included participants (37 female, 3 male) had a mean age of 15.8 (SD 1.6) years (Table 1) and exhibited a mean femoral version of 25.0° (SD 10.8°) as well as a tibial torsion of 35.6° (SD 7.8°) (Figure 4). The marker errors for the inverse kinematics were below OpenSim's best practice recommendations [46].

3.1 | Impact of Lower Limb Torsion on Patellofemoral Joint Loading

The participants presented a mean difference of 4.0° (SD 10.8°) for the femoral version and 11.6° (SD 7.8°) for tibial torsion between the generic-torsion and torsion-informed models. This led to a mean absolute difference of 0.09 (0.07 SD) BW and 0.15 (0.10 SD) BW for the maximum mediolateral and overall patellofemoral force, respectively.

The femoral version did not correlate to differences in mediolateral patellofemoral force between the generic-torsion and torsion-informed models ($\rho = 0.01$, $p = 0.967$; $\rho = 0.13$, $p = 0.425$ for maximum and RMS forces, respectively), nor did it correlate to differences in resultant patellofemoral forces ($\rho = -0.27$, $p = 0.087$) (Figure 5a).

Tibial torsion showed a significant positive correlation ($\rho = 0.39$, $p = 0.014$) to differences in maximum mediolateral patellofemoral force between the generic-torsion and torsion-informed models, but did not correlate to differences in RMS mediolateral patellofemoral force ($\rho = 0.15$, $p = 0.328$). Higher tibial torsion

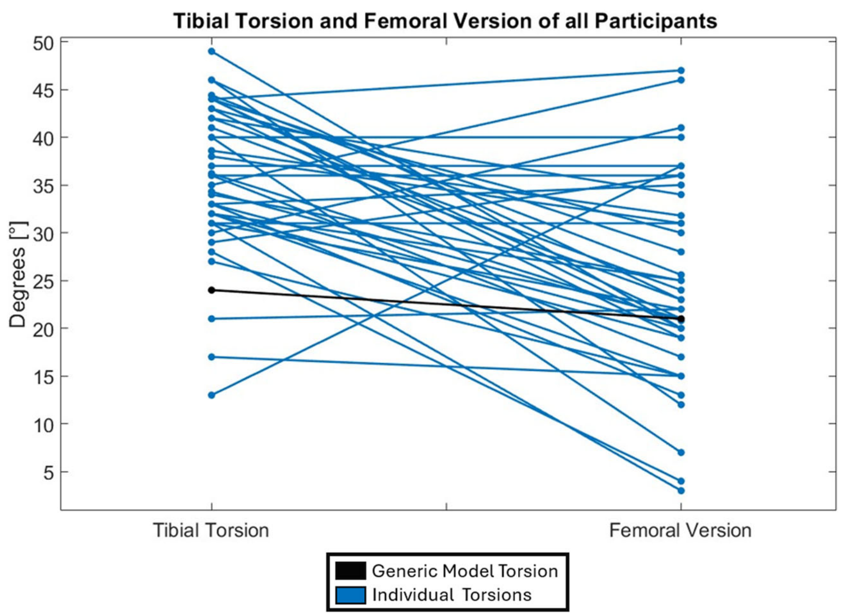

FIGURE 4 | Femoral version and tibial torsion of each participant. Each two dots connected with a line represents the tibial torsion (left dot) and femoral version (right dot) of one participant.

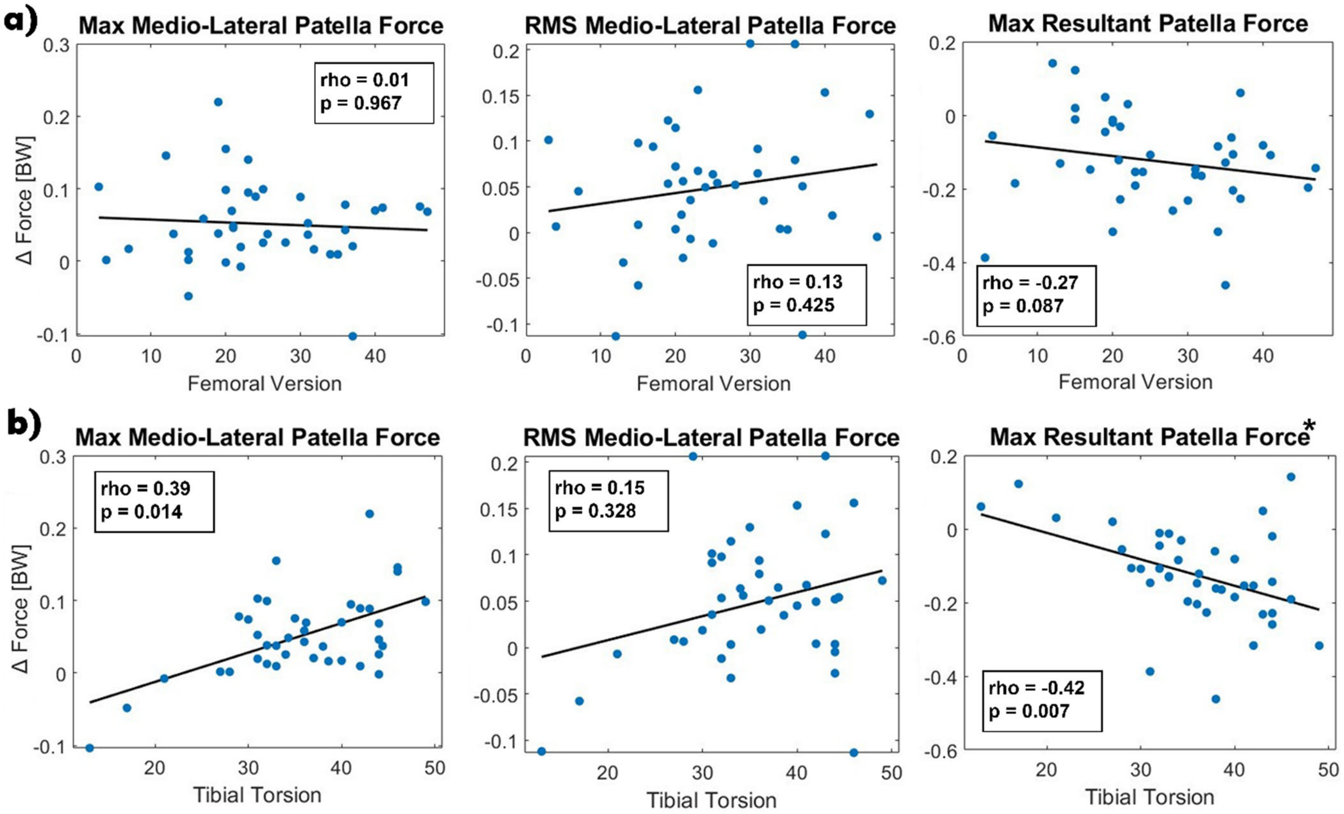

FIGURE 5 | Scatterplots showing the relationship between femoral version (a) and tibial torsions (b) and differences in patellofemoral forces between generic-torsion model and torsion-informed model. The black line represents a least square fitted line. A superscript asterisk represents a significant Spearman's correlation. BW = body weight (mass × g), Max = maximum, RMS = root mean square.

TABLE 2 | Regression coefficients and significance.

Variable	Coefficient	Standard error	*t*-value	*p* value
Constant	−0.112	0.062	−1.81	0.078
Tibial torsion*	0.005	0.002	3.34	0.002
Femoral version	0.001	0.001	0.68	0.498

Note: Significant variables are highlighted with an asterisk.

indicated a higher maximum lateralizing force (Figure 5b). Tibial torsion negatively correlated to differences in resultant patellofemoral loading between models ($\rho = -0.42$, $p = 0.007$).

Tibial torsion correlated to differences in gait pattern between generic-torsion and torsion-informed models. Tibial torsion correlated with changes toward a more internally rotated knee progression angle ($\rho = 0.81$, $p < 0.001$), hip internal rotation ($\rho = 0.93$, $p < 0.001$), reduced hip flexion ($\rho = -0.44$, $p = 0.005$), and reduced knee flexion angle ($\rho = -0.77$, $p < 0.001$). The femoral version did not correlate to any kinematic changes between generic-torsion and torsion-informed models.

A multiple linear regression analysis revealed a significant relationship between difference in maximum mediolateral patellofemoral force to tibial torsion and femoral version with an r^2 of 0.21 ($F = 5.97$, $p = 0.006$). While tibial torsion was significantly related to the difference in maximum mediolateral patellofemoral force, femoral version showed no significant relationship (Table 2).

The maximum and RMS mediolateral patellofemoral forces, as well as resultant patellofemoral force, in the torsion-informed model correlated to tibial torsion (Supporting Information 1). The femoral version did not correlate to the maximum and RMS mediolateral patellofemoral forces, as well as the resultant patellofemoral force, calculated based on the torsion-informed model.

3.2 | Impact of Neglecting Tibial Torsion on Correlation Results

For our second objective, we evaluated how neglecting tibial torsion affects our findings. The participants presented a mean difference of 4.0° (SD 10.8°) for femoral version between the generic-torsion and femoral-version models. This led to a mean absolute difference of 0.02 (0.02 SD) BW and 0.07 (0.06 SD) BW for the maximum mediolateral and overall patellofemoral force, respectively.

When neglecting tibial torsion, femoral version was significantly correlated to differences in mediolateral patellofemoral force ($\rho = 0.65$, $p < 0.001$; $\rho = 0.59$, $p < 0.001$, for maximum and RMS, respectively), as well as differences in resultant patellofemoral force between the generic-torsion and the femoral-version models ($\rho = -0.58$, $p < 0.001$). A higher femoral anteversion was associated with a lower medializing/higher lateralizing force (Figure 6). These findings are in contrast to the findings of our primary investigation.

3.3 | Hip Muscle Lever Arms

The femoral version correlated to differences in mean gluteal muscle lever arms between the generic-torsion and torsion-informed models. Moreover, tibial torsion correlated to differences in gluteal muscle lever arms between models (Table 3). As hip muscle lever arms were not directly affected by tibial torsion in the model's neutral pose, the potential contributing factors were further investigated. A strong correlation ($\rho = 0.97$, $p < 0.001$) was identified between the participants' tibial torsion and differences in hip rotation between the torsion-informed and generic-torsion models, whereas no significant correlation ($\rho = 0.1$, $p = 0.55$) was observed between the participant's femoral version and differences in hip rotation between those models (tibial torsion: $\rho = 0.97$, $p < 0.001$; femoral version: $\rho = 0.1$, $p = 0.55$). A more externally rotated tibia was associated with a more internally rotated hip. After controlling the correlation analysis for hip rotation, the correlation between gluteal muscle lever arms and tibial torsion was no longer statistically significant, while the correlation coefficient between femoral version and gluteal muscle lever arms increased (Table 3).

3.4 | Muscle Forces

Femoral version and tibial torsion correlated to the differences in RMS of normalized muscle forces between torsion-informed and generic-torsion models (Table 4, Supporting Information 2). Specifically, femoral version correlated with differences in gluteus medius and minimus, rectus femoris and vastus lateralis muscle forces between models. Tibial torsion correlated to differences in gluteus maximus, biceps femoris, and rectus femoris muscle forces between models.

4 | Discussion

The objective of this study was to examine the influence of lower limb torsion on patellofemoral joint loading in individuals with patellofemoral instability. Musculoskeletal simulations were conducted on a large sample of 40 participants with recurrent patellofemoral dislocations. Tibial torsion correlated to differences in maximum mediolateral patellofemoral force between the generic torsion and torsion-informed models. The femoral version did not correlate to differences in mediolateral patellofemoral forces between the generic-torsion and torsion-informed models. Interestingly, personalizing solely the femoral version in the musculoskeletal models and thus neglecting each individual's tibial torsion led to a moderate to high correlation between the femoral version and differences in mediolateral patellofemoral force between the generic-torsion and femoral-version models.

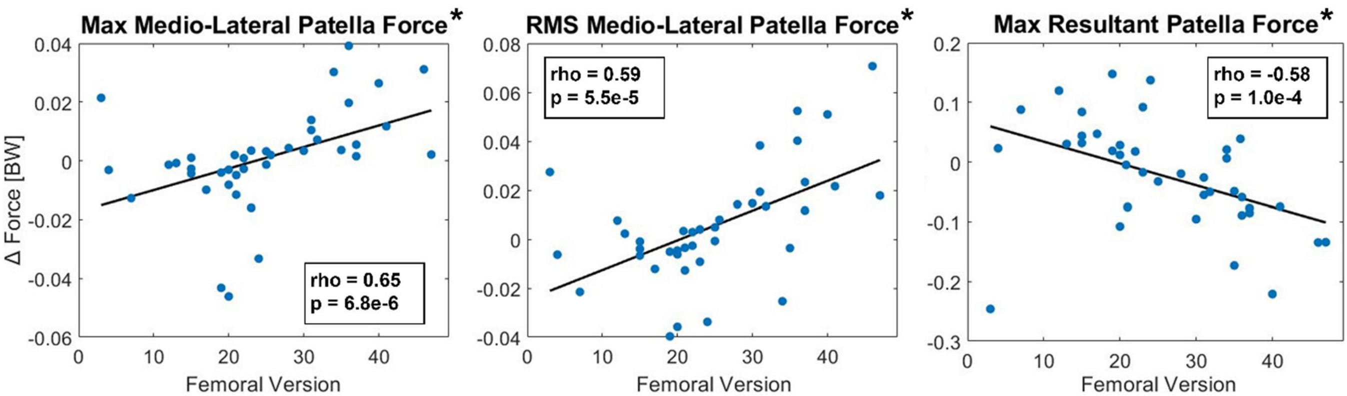

FIGURE 6 | Scatterplots showing the relationship between femoral version and differences in patellofemoral forces between generic-torsion model and femoral-version model. The black line represents a least square fitted line. A superscript asterisk represents a significant Spearman correlation. BW = body weight (mass × g), Max = maximum, RMS = root mean square.

TABLE 3 | Spearman's correlation coefficient between femoral version, tibial torsion, and difference in hip muscle lever arms between generic-torsion and torsion-informed model.

	FV	FV controlled	TT	TT controlled
Gluteus minimus	−0.78**	−0.98**	0.51**	0.07
Gluteus medius	−0.78**	−0.97**	0.51**	0.09
Gluteus maximus	−0.79**	−0.95**	0.47*	0.09
Psoas	0.73**	0.85**	−0.39*	−0.004
Rectus femoris	0.11	0.20	−0.48*	0.23

Note: Controlled correlations were controlled for hip rotation.
Abbreviations: FV = femoral version, TT = tibial torsion; Statistical significance: *$p < 0.05$; **$p < 0.001$.

TABLE 4 | Spearman's correlation coefficient between femoral version, tibial torsion, and difference in RMS muscle forces between generic-torsion and torsion-informed model.

	Femoral version	Tibial torsion
Biceps femoris long head	−0.38*	0.67**
Biceps femoris short head	0.27	0.60**
Gastrocnemius	−0.08	0.27
Gluteus maximus	−0.21	−0.42*
Gluteus medius	−0.57**	0.46*
Gluteus minimus	0.73**	−0.19
Rectus femoris	0.33*	−0.41*
Vastus lateralis	−0.59**	−0.17
Vastus medialis	−0.07	−0.19
Tibialis anterior	0.15	0.49*

Statistical significance:
*$p < 0.05$; **$p < 0.001$.

We found a mean absolute difference of 0.09 BW in peak mediolateral patellofemoral force between the generic-torsion and torsion-informed models. Considering the mean participants' weight of 59.8 kg, the mediolateral force changed about 54 N to a more lateralizing force due to the lower limb torsion. Compared to the maximum strain force of the medial patellofemoral ligament (about 208 N) [47], this is a relevant change which could affect the stability of the patellofemoral joint by increasing the lateralizing traction on the patella.

Tibial torsion correlated to the differences in maximum mediolateral patellofemoral force between the generic-torsion and torsion-informed models, as well as maximum mediolateral patellofemoral force in the torsion-informed models. A higher tibial torsion was, therefore, associated with a more lateralizing force. These findings are corroborated by the results of cadaver studies, which demonstrated an elevation in pressure on the lateral patellar facet [10, 48]. Two effects are of particular relevance when considering the impact of tibial torsion on the stability of the patellofemoral joint. First, tibial torsion may result in alterations to kinematics as a compensatory mechanism [49]. This is supported by the observed correlations between tibial torsion and changes in knee flexion, knee progression, hip flexion, and hip rotation between the generic-torsion and torsion-informed models. Second, even though not altered and therefore not investigated in this study, the position and orientation of the tibial tubercle seem to play a significant role in patella alignment within the context of tibial anatomy [50]. As proximal tibial torsion can result in a displacement of the tibial tuberosity, previous studies indicated that proximal tibial derotation osteotomies may be beneficial for stabilizing the patellofemoral joint [8, 9]. Based on the findings of these previous studies and our results, excessive tibial torsion should be considered as a factor influencing the stability of the patellofemoral joint.

In the models with personalized lower limb torsion, we found no correlation between the femoral version and differences in mediolateral patellofemoral force between models. Furthermore, we found no correlation between the femoral version and the mediolateral patellofemoral forces in the torsion-informed models. This is in contrast to previous musculoskeletal simulations [18, 20] and cadaver studies [11, 12], which solely investigated the impact of femoral version on knee joint loads. These studies revealed an increasing lateralizing patellofemoral loading with increasing femoral anteversion. Given that non-traumatic patella dislocations typically occur in the lateral direction, this seems of particular relevance [2]. For our second objective, we investigated the impact of isolated personalization of femoral version on joint loads. This investigation revealed moderate to strong correlations between the femoral version and differences in mediolateral patellofemoral force between the generic-torsion and femoral-version models. This result would indicate that an excessive femoral version may destabilize the patella, as it pulls the patella more to the lateral side. However, this conclusion would be misleading as tibial torsion was neglected in these simulations. Accounting for personalized tibial torsion additionally to the femoral version altered the impact of the femoral version on differences in patellofemoral joint loads between models. Hence, findings from previous studies solely focusing on femoral version and neglecting tibial torsion should be interpreted with caution.

Tibial torsion was strongly correlated to differences in hip rotation and knee progression angle between the generic-torsion and torsion-informed models, which might explain the impact on patellofemoral joint loads. The hip rotation has an effect on knee progression angle and, thus, on the global orientation of the knee joint. When the ground reaction force remains constant, differences in knee progression angle may result in a redistribution of the force along the anatomical axes of the knee joint. An increased internal hip rotation and knee progression angle, therefore, might lead to a shift toward a more lateralized ground reaction force acting on the knee. A study examining the influence of kinematics and bony morphology on hip joint loading revealed that the impact of kinematics on hip joint loading appears to be more pronounced than that of morphology [51]. Considering the strong correlation between hip rotation, knee progression angle and tibial torsion, this may elucidate why stronger correlations were observed between patellofemoral joint loading and tibial torsion than between patellofemoral joint loading and femoral version in models with subject-specific lower limb torsion.

Alteration of muscles' lever arms and consequently forces might explain the effect of lower limb torsion on patellofemoral joint loads. The differences of hip muscles' lever arms in the torsion-informed models correlated to both femoral version and tibial torsion. The observed correlation between the femoral version and differences in hip muscle lever arms between the generic-torsion and the torsion-informed models is in coherence with previous musculoskeletal simulation studies [15, 52]. After controlling for differences in hip rotation, we observed that the correlations between tibial torsion and differences in gluteal muscle lever arms between the generic-torsion and the torsion-informed models diminished, while the correlations between femoral version and differences in gluteal muscle lever arms increased. Consequently, tibial torsion exerted an indirect influence on hip muscle lever arms by altering hip rotation.

The present study showed contrasting findings in the correlations between differences of mediolateral patellofemoral force and femoral version when personalizing lower limb torsion compared to neglecting tibial torsion. The models with subject-specific lower limb torsion exhibited a higher degree of personalization and, therefore, a more realistic representation of the participants' anatomy. Femoral version can have a destabilizing effect on the patellofemoral joint. However, it is imperative to evaluate this effect in conjunction with other contributing factors, namely the participant's walking pattern and tibial torsion. These factors have the potential to mitigate the destabilizing effects of the femoral version on the patellofemoral joint and, therefore, should be considered in the treatment plan.

In addition to soft tissue procedures, femoral derotation osteotomies are used for the treatment of recurrent patellofemoral instability [5]. At present, the principal method of decision-making in this context is the utilization of femoral version thresholds or application of individual surgeon preferences. For instance, studies suggested a threshold of > 30° femoral version may be a useful indicator [6, 53] to perform a femoral derotation osteotomy. In light of our findings, a threshold criterion based solely on the value of femoral version appears inadequate, as other factors (e.g., tibial torsion, individual movement pattern) appear to influence the impact of femoral version on the mediolateral patellofemoral stability. It is possible that neglecting these other factors could lead to unsatisfactory results after a femoral derotation osteotomy. Therefore, it is essential to investigate not only femoral version, but also the overall torsion of the lower limb. Furthermore, it seems necessary to implement functional assessments, such as 3D gait analysis, as our regression analysis solely based on the lower limb torsion was able to explain only 21% of changes in maximum mediolateral patellofemoral force. This is of particular significance in the context of malalignment syndrome with excessive femoral version and tibial torsion. Especially in such cases, both femoral version and tibial torsion should be taken into account when contemplating a treatment plan [54]. In addition to the clinically established use of 3D gait analysis, the assessment of more demanding movement patterns, which are more likely to predispose patients to patella dislocation, could further enhance the relevance of functional assessment for treatment planning. The objective of future investigations should be to identify a combination of morphological and functional criteria to determine if treating patellofemoral instability with derotation osteotomies should be undertaken.

This study is subject to certain limitations. First, the Torsion Tool [16] was utilized to modify femoral version and tibial torsion for all participants (Figure 2). While the personalized torsion represents the values derived from the rotational MRIs, it may not accurately reflect the precise location of torsion in each participant. Second, the metatarsal and subtalar joints were locked in all models. This is a standard procedure in musculoskeletal simulations when only two markers are available on the foot segment [33, 55, 56]. Nevertheless, as the subtalar joint influences the rotational alignment of the lower limb [57], this may have influenced joint kinematics and consequently patellofemoral joint loads. Furthermore, the analysis did not include axial deviations in the frontal plane or other personalized morphology (e.g., trochlea shape, position of the tibial tuberosity). These additional parameters could also influence the load on the patellofemoral joint.

5 | Conclusion

In conclusion, although an increased femoral version might destabilize the patellofemoral joint, this effect diminishes when accounting for subject-specific tibial torsion. Studies addressing only the femoral version should, therefore, be interpreted with caution. In a clinical context, our results indicate that planning and performing derotation osteotomies based solely on morphological examination of the femur may result in unanticipated and unfavorable clinical outcomes. Treatment decisions should, therefore, be made on an individual basis, considering the overall clinical picture, including femoral version, tibial torsion, and functional movement patterns. Lastly, our study showed that excessive tibial torsion should be considered as a factor influencing the stability of the patellofemoral joint.

Author Contributions

Initial idea: Bernhard Guggenberger, Hans Kainz, and Martin Svehlik. Conceptualization: Andreas Habersack, Bernhard Guggenberger, Hans Kainz, and Martin Svehlik. Data acquisition: Andreas Habersack, Bernhard Guggenberger, Tanja Kraus, and Matthias Sperl. Analysis: Andreas Habersack, Bernhard Guggenberger, Hans Kainz, and Willi Koller. Methodology: Bernhard Guggenberger, Martin Svehlik, Hans Kainz, and Willi Koller. Interpretation: Andreas Habersack, Bernhard Guggenberger, Hans Kainz, Martin Svehlik, Matthias Sperl, Tanja Kraus, and Willi Koller. Writing original draft: Bernhard Guggenberger, Hans Kainz, and Martin Svehlik. Critical revision of final draft: all authors. All authors approved the final version.

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Supporting Information

Additional supporting information can be found online in the Supporting Information section.