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The influence of sagittal trunk posture on the magnitude and rate of patellofemoral joint stress during stair ascent in asymptomatic females

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ABSTRACT

Background: Excessive patellofemoral joint stress (PFJS) is thought to be a contributory factor to patellofemoral pain (PFP). Thus, treatment strategies that minimize PFJS rate and magnitude during painful activities like stair ascent may be useful for optimizing outcomes for PFP patients. Sagittal plane trunk posture has been shown to influence PFJS during running although it is unknown if a similar relationship exists during stair ascent. Research Question: Does altering sagittal plane trunk posture affect PFJS rate and/or magnitude during stair ascent? Methods: Twenty asymptomatic females (23.4 ± 2.5 yr; height: 164.4 ± 7.9 cm; mass: 63.0 ± 12.2 kg) performed 5 stair ascent trials (96 steps/min) during 3 conditions: self-selected trunk (SS), flexed trunk (FLX), and extended trunk (EXT). Three-dimensional kinematics (200 Hz) and ground reaction forces (2000 Hz) were collected during each trial. A previously described mathematical model was used to calculate PFJS that included subject-specific and non-subject-specific model inputs. Dependent variables included sagittal plane trunk angle, and the rates and magnitudes of PFJS, patellofemoral joint reaction force (PFJRF), and PFJ contact area during the stance phase of stair ascent. Results: Compared to SS, peak PFJS decreased during FLX (mean difference (MD) = 2.6 MPa; $p < 0.001$; 95%CI = 2.2 to 2.9; effect size (ES) = 5.2) and increased during EXT (MD = -3.3 MPa; $p < 0.001$; 95%CI = -3.9 to -2.6; ES = -3.4). Similarly, PFJS rate decreased during FLX (MD = 17.8 MPa/sec; $p < 0.001$; 95%CI = 13.6 to 21.9; ES = 3.6) and increased during EXT (MD = -14 MPa/sec; 95%CI = -19.4 to -8.7; $p < 0.001$; ES = -2.2). Significance: Sagittal plane trunk posture influences PFJS rate and magnitude during stair ascent in asymptomatic females. Increasing and decreasing forward trunk flexion resulted in decreased and increased PFJS respectively. Future studies should examine the effects of these movement strategy modifications on pain and function in patients with PFP.

1. Introduction

Patellofemoral pain (PFP) is a complex, multifactorial condition that can limit one's quality of life [1]. The reported annual prevalence for PFP in the general population is 22.7% [2] and females are at greater risk of developing the condition than males [3]. The primary symptom associated with PFP is pain located behind or around the patella that is exacerbated with demanding activities such as stair ascent [4].

Excessive patellofemoral joint stress (PFJS) (force per unit area) is reported to be linked with PFP during these types of activities [5]. Although it has been reported that stair ambulation exacerbates PFP [6], research has not confirmed the stress-theory of PFP during this particular activity. For example, Heino-Brechter and Powers reported there were no differences in peak PFJS between PFP subjects and asymptomatic controls during stair ascent [7]. However, despite this observed similarity the PFP subjects in this study ascended the stairs at a slower cadence (and subsequently slower velocity) than the control subjects which may have influenced peak PFJS. Furthermore, research indicates that persons with PFP have a reduced pain pressure threshold compared to control subjects [8–10] which may be further reduced following a bout of stair ambulation [11]. Thus, for individuals with PFP, an otherwise normal PFJ loading environment during stair ascent may be sufficient for exacerbating symptoms due to a reduced tissue load tolerance. This is consistent with the theory that the mechanical loading threshold for experiencing pain is reduced following symptomatic disruption of joint homeostasis [12,13].

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Besides the peak PFJ load, Silva et al. reported that females with PFP experience an increased rate of lower limb loading compared to control subjects during stair ascent [14] which could result in an increased rate of loading across the PFJ. Considering the viscoelastic nature of patellar bone and articular cartilage it is conceivable that the rate of PFJ loading may also be a contributory factor to PFP symptoms. This premise is supported by Atkins et al. who reported that the average rate of loading across the PFJ was an independent predictor ($r = 0.42$, $p = 0.02$) of the magnitude of increases in PFP across repeated single limb landings [15]. Additionally, previous studies reported that movement retraining which resulted in a decreased rate of lower limb loading was associated with improved symptoms for runners with PFP [16,17].

Therefore, given the plausibility that PFP is related to both the magnitude and rate of PFJS it has been recommended that interventions be identified that minimize both variables (PFJS rate and magnitude) during provocative activities like stair ambulation [15]. Currently however, there is a lack of evidence describing interventions that effectively decrease either PFJ loading variable during stair ascent. For example, Powers et al. described a patellar brace that led to decreased peak PFJS stress during free and fast walking [18] but had no effect on peak PFJS during stair ascent [19]. One potential intervention involves instructing patients to utilize a flexed trunk posture. During running, Teng et al. reported that compared to a self-selected trunk posture, flexed, and extended trunk postures resulted in significant ($p \leq 0.05$) decreases and increases in peak PFJS respectively for healthy individuals [20]. Although promising, it is unknown if a similar relationship exists between trunk posture and the rate and magnitude of PFJS during other gait-related activities (besides running) such as stair ascent.

Increased understanding of the relationship between trunk posture and PFJ mechanics during stair ascent will aid in developing optimal PFP treatment strategies and provide valuable insight regarding the contribution of abnormal stair ascent biomechanics to PFJ dysfunction. The purpose of this study was to examine the influence sagittal plane trunk posture on the rate and magnitude of PFJS during stair ascent. It was hypothesized that during stair ascent PFJS magnitude and rate would decrease when utilizing a forward flexed trunk posture and increase when utilizing an extended trunk posture compared to a natural, self-selected trunk posture.

2. Methods

2.1. Subjects

Twenty asymptomatic females (age: 23.4 ± 2.5 yr; height: 164.4 ± 7.9 cm; mass: 63.0 ± 12.2 kg; body mass index: 23.1 ± 3.0 kg/m^2) were recruited for this study. To be included subjects must have been aged 18–45 years, have a body mass index < 30 kg/m^2, and free of lower limb or low back pain at the time of testing. Potential subjects were excluded if they were non-English speaking, had prior lower limb or low back surgery, were diagnosed with lower limb or low back injury that caused pain during stair ambulation within previous year, had a neuromuscular condition that compromised safety during stair ambulation, or were pregnant. The Institutional Review Board of the affiliated university approved this study. Prior to participation, all subjects provided written informed consent.

2.2. Instrumentation

Three-dimensional kinematics (200 Hz) were recorded using an 8-camera infrared motion capture system (Vicon, Centennial, CO, USA). Three-dimensional ground reaction forces (2000 Hz) were recorded using an in-ground force plate (Bertec, Columbus, OH, USA) embedded into the first step of the stairs (Fig. 1).

2.3. Subject preparation

Participants donned standard shoes (Asics, Irvine, CA, USA) then body height and weight were measured and recorded. Next, double-sided tape was used to secure reflective calibration markers (14 mm in diameter) to the following anatomic landmarks: first, second, and fifth metatarsal heads and the medial and lateral malleoli, femoral epicondyles, iliac crests, and anterior and posterior superior iliac spines. Additionally, rigid tracking clusters consisting of at least three non-collinear markers were secured to the thighs, legs, and feet using tape and elastic Velcro straps (Fig. 1). An additional cluster of four markers was secured to the posterior surface of the upper trunk and aligned such that it was in a plane approximately parallel to the frontal plane of each subject's trunk while they assumed a static standing position. Next, a static standing calibration trial was recorded and used to define joint axes of rotation and local segment coordinate systems. All anatomic calibration markers, except those on the pelvis, were then removed.

2.4. Stair ascent trials

Following preparation subjects performed 5 stair ascent trials during 3 conditions for a total of 15 trials. For the first condition, subjects ascended the stairs using their self-selected (SS) trunk posture. For the second and third conditions, subjects ascended the stairs while either flexing (FLX) or extending (EXT) their trunk as much as comfortably possible (Fig. 1). The FLX and EXT conditions were performed after SS so that it could be confirmed visually that changes were effected compared to SS. The order of the FLX and EXT conditions was randomized to minimize learning effects. During all trials subjects stepped to the beat of a metronome (96 beats per minute) [21] to minimize the influence of varying cadences across conditions. A trial was considered successful if the subject's foot landed within the borders of the force plate embedded in the first step of the stairs (Fig. 1). Prior to recording subjects were allowed several practice trials to familiarize themselves with the task and minimize learning effects.

2.5. Data reduction

Kinematic and kinetic data were reconstructed and low-pass filtered (fourth order, Butterworth, 6 Hz) using Vicon Nexus Software (v2.6.1, Vicon, Centennial, CO, USA). Next these data were imported into Visual 3D software (v6, C-motion, Germantown, MD, USA) where a six-degrees-of-freedom link segment model was applied. The pelvis segment was modeled as a cylinder and the thigh, leg, and foot segments were modeled as frusta of cones. Joint kinematics were calculated as movement of the distal segment relative to the proximal segment. Segment kinematics were calculated as movement of the segment relative to the global coordinate system. Net internal joint moments were calculated using inverse dynamics equations and normalized to each subject's body mass.

2.6. Patellofemoral joint stress model

A previously described model was used to estimate PFJS (Fig. 2) [22]. Subject-specific model inputs included knee flexion kinematic and knee extension moment data obtained during the stance phase of the stair ascent trials. Additional model inputs that were not subject-specific were obtained using data from previous studies and included quadriceps muscle lever arm length [23], a constant describing the ratio of quadriceps muscle force relative to the patellofemoral joint reaction force (PFJRF) [24], and PFJ contact area [25].

The first step in estimating PFJS was to calculate the quadriceps lever arm length which was achieved by fitting a non-linear equation to data from a previous study that reported the length at varying knee flexion angles [23]. The quadriceps muscle force was calculated by dividing the net knee extension moment by the quadriceps lever arm.

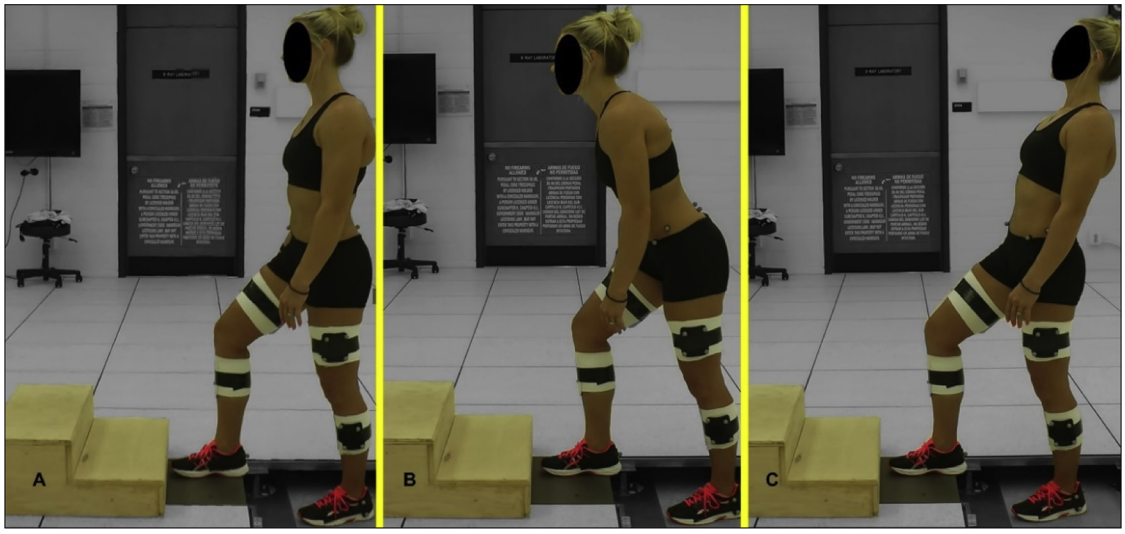

Fig. 1. Example of A) self-selected, B) forward-flexed, and C) extended trunk conditions during the stair ascent trials.

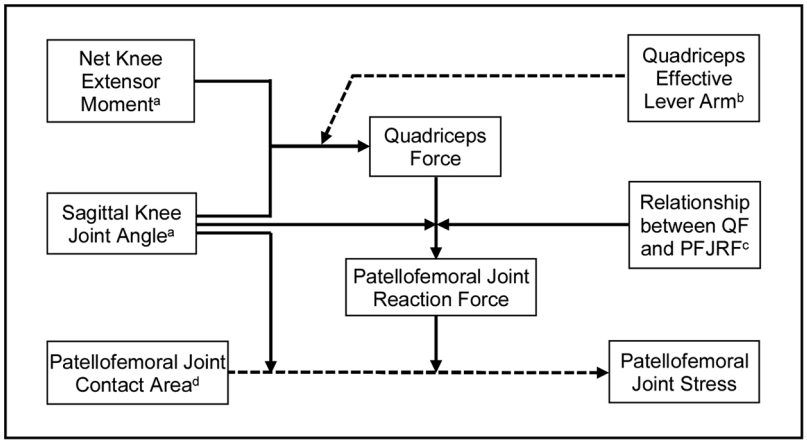

Fig. 2. Flow diagram of patellofemoral joint model. (a) Subject-specific data from motion capture system. (b) Data from van Eijden et al. 1986 (c) Data from van Eijden et al. 1987 (d) Data from Powers et al. 1998.

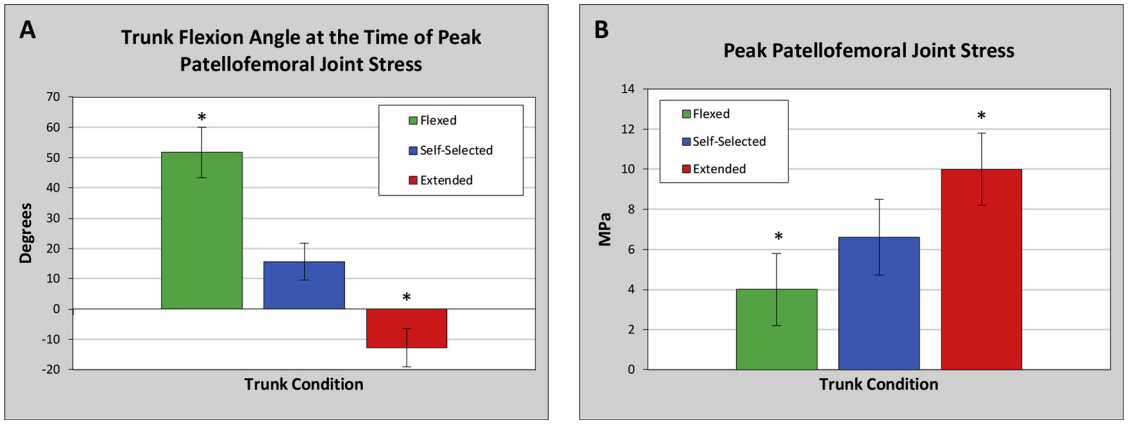

Fig. 3. Comparison of group averages during each condition for (A) trunk flexion angle at the time of peak patellofemoral joint stress and (B) peak patellofemoral joint stress. *significantly different from Self-Selected condition ($p < 0.017$). Error bars represent one standard deviation.

The time-series PFJRF was calculated by multiplying the quadriceps muscle force by a constant that described the ratio of the quadriceps muscle force to the PFJRF as a function of knee flexion angle [24]. Next, PFJ contact area was estimated by fitting a non-linear equation to data from a previous study that reported the contact area between the patella and femur at varying angles of knee flexion [25]. Finally, PFJS was determined by calculating the quotient of the PFJRF divided by PFJ contact area. Model outputs of interest included time-series curves for PFJS, PFJRF, and PFJ contact area during the stance phase of each stair ascent trial.

Primary dependent variables of interest included the sagittal plane trunk angle (degrees) at the time of peak PFJS, average sagittal plane

Table 1
Biomechanical Variables at the Time of Peak Patellofemoral Joint Stress (mean ± SD).

	Flexed	Self-Selected	Extended	Self-Selected - Flexed		Self-Selected - Extended	
				Mean Difference (95% CI)	Effect Size (Cohen's d)	Mean Difference (95% CI)	Effect Size (Cohen's d)
Patellofemoral Joint Reaction Force (N/kg)*	15.4 ± 6.4	25.5 ± 7.1	38.2 ± 6.8	10.1 (8.3 to 11.8)	**4.0**	−12.7 (-15.0 to -10.5)	**-3.7**
Patellofemoral Joint Contact Area (mm^2)	234.7 ± 3.9	237.1 ± 2.6	240.1 ± 0.9	2.4 (0.8 to 4.1)	**1.0**	−3.0 (-4.3 to -1.7)	**-1.3**
Net Knee Extension Moment (Nm/kg)	0.7 ± 0.3	1.1 ± 0.3	1.4 ± 0.2	0.4 (0.3 to 0.5)	**3.8**	−0.3 (-0.4 to -0.3)	**-2.3**
Knee Flexion Angle (°)	46.6 ± 4.0	49.3 ± 4.2	56.5 ± 4.0	2.7 (1.1-4.3)	**1.1**	−7.2 (-9.1 to -5.4)	**−2.6**

*Indicates non-parametric statistical tests used to examine for differences; Negative mean difference/effect size indicates an increase compared to Self-Selected condition; Bold effect size indicates significant difference between respective conditions ($p < 0.017$).

trunk angle (degrees), peak PFJS (MPa), and average rate of PFJS development (MPa/sec) as calculated from initial contact to the time of peak PFJS. Secondary dependent variables of interest included the PFJRF (N/kg), PFJ contact area (mm^2), knee extension moment (Nm/kg), and knee flexion angle (degrees) at the time of peak PFJS. Additionally the average rate of change in each secondary variable was calculated as the magnitude of change from initial contact until the time of peak PFJS, divided by time (positive values indicate an increasing rate).

2.7. Statistical analyses

Initially, descriptive statistics including mean and standard deviation were calculated. Next the normality and sphericity assumptions for each dependent variable were examined by calculating Kolmogorov-Smirnov and Mauchly's tests respectively. Differences in each dependent variable across the three trunk conditions were then examined by calculating one-way repeated measures analyses of variance (ANOVAs) ($\alpha = 0.05$). Post-hoc Bonferonni-adjusted paired t-tests ($\alpha = 0.017$) were calculated for each significant ANOVA. Additionally, mean differences (MD), 95% confidence intervals (95%CI), and Cohen's d effect size (ES) for paired samples were calculated [26]. Dependent variables not normally distributed were examined using Friedman's one-way ANOVA ($\alpha = 0.05$) and Wilcoxon signed rank non-parametric tests ($\alpha = 0.017$) for post-hoc comparisons. A Greenhouse-Geisser transformation was applied to data that did not meet the sphericity assumption. All statistical tests were calculated using SPSS software (v21, IBM Corp., Armonk, NY, USA).

3. Results

3.1. Peak patellofemoral joint stress

Analysis of variance results for the dependent variables (calculated at the time of peak PFJS) indicated there were differences across conditions for trunk flexion angle ($p < 0.001$; Fig. 3), peak PFJS ($p < 0.001$; Fig. 3), PFJRF ($p < 0.001$; Table 1), PFJ contact area ($p < 0.001$; Table 1), knee extension moment ($p < 0.001$; Table 1), and knee flexion angle ($p < 0.001$; Table 1).

Post-hoc testing indicated that at the time of peak PFJS, trunk flexion angle increased during FLX (50.2 ± 8.1°) (MD = −34.7°; 95%CI = -38.9, −30.5; $p < 0.001$; ES = −5.5) and decreased during EXT (−12.9 ± 6.8°) (MD = 28.4°; 95%CI = 25.9,30.9; $p < 0.001$; ES = 7.5) compared to SS (15.5 ± 6.6°) (Fig. 3). In contrast, peak PFJS decreased during FLX (4.2 ± 1.9 MPa) (MD = 2.6 MPa; 95%CI = 2.2, 2.9; $p < 0.001$; ES = 5.2) and increased during EXT (10.1 ± 2.1 MPa) (MD = -3.3 MPa; 95%CI = -3.9,-2.6; $p < 0.001$; ES-3.4) compared to SS (6.8 ± 1.9 MPa) (Fig. 3). The PFJRF at time of peak stress decreased during FLX ($p < 0.001$) and increased during EXT ($p < 0.001$) compared to SS (Table 1). Similarly, PFJ contact area decreased during FLX ($p = 0.006$) and increased during EXT ($p = 0.001$) compared to SS (Table 1). The knee extension moment at the time of peak PFJS decreased during FLX ($p < 0.001$) and increased during EXT ($p < 0.001$) compared to SS (Table 1). Knee flexion angle at the time of peak PFJS decreased during FLX ($p = 0.002$) and increased during EXT ($p < 0.001$) compared to SS (Table 1).

3.2. Rate of patellofemoral joint stress development

Analysis of variance results for PFJS-rate dependent variables indicated there were differences across conditions for the average trunk flexion angle ($p < 0.001$; Fig. 4), and average rates of change in PFJS development ($p < 0.001$; Fig. 4), PFJRF development ($p < 0.001$; Table 2), knee extension moment ($p < 0.001$; Table 2), and knee flexion angle ($p < 0.001$; Table 2).

Post-hoc testing indicated that the average trunk flexion angle

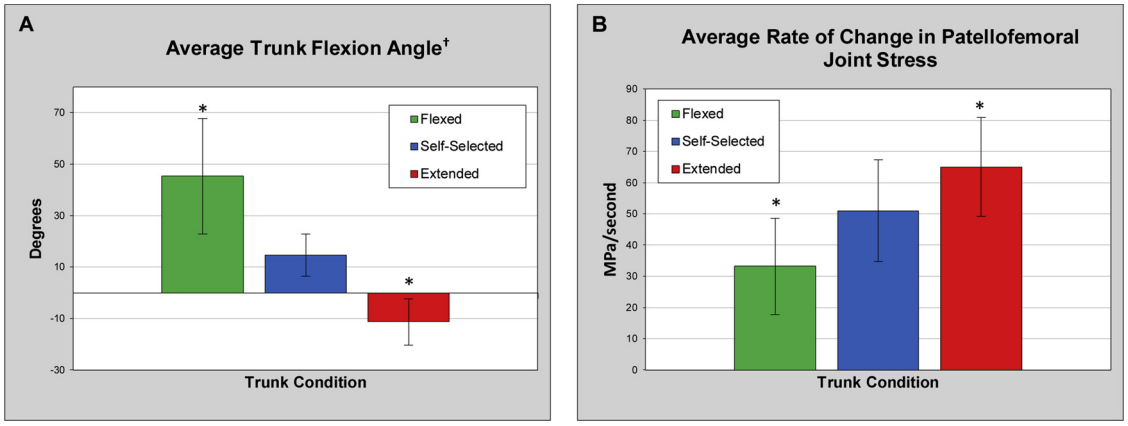

Fig. 4. Comparison of group averages during each condition for (A) average trunk flexion angle from initial contact to the time of peak patellofemoral joint stress and (B) average rate of change of patellofemoral joint stress from initial contact to the time of peak patellofemoral joint stress. †examined using non-parametric statistical tests; *significantly different from Self-Selected condition ($p < 0.017$). Error bars represent one standard deviation.

increased during FLX (45.4° ± 21.8°) (MD = -30.8°; 95%CI = −41.3, −20.3; $p < 0.001$; ES = -3.2) and decreased during EXT (−11.3° ± 8.7°) (MD = 25.8°; 95%CI = 18.5,33.1; $p < 0.001$; ES = 2.9) compared to SS (14.6° ± 8.1°) (Fig. 4). In contrast, PFJS increased at a slower rate during FLX (33.2 ± 15.5 MPa/sec) (MD = 17.8 MPa/s; 95%CI = 13.6,21.9; $p < 0.001$; ES = 3.6) and a faster rate during EXT (65.0 ± 15.8 MPa/sec) (MD = -14.0; 95%CI = −19.4, −8.7; $p < 0.001$; ES = −2.2) compared to SS (51.0 ± 16.3 MPa/s) (Fig. 4). Additionally, the PFJRF increased at a slower rate during FLX ($p < 0.001$) and faster rate during EXT ($p < 0.001$) compared to SS (Table 2). Similarly, the knee extension moment increased at a slower rate during FLX ($p < 0.001$) and a faster rate during EXT ($p = 0.006$) compared to SS (Table 2). Conversely, the knee flexion angle decreased at a slower rate during FLX ($p < 0.001$) and faster rate during EXT ($p = 0.002$) compared to SS.

4. Discussion

Findings from this study support the hypothesis that sagittal trunk posture influences the magnitude and rate of PFJS during stair ascent. In particular, PFJS rate and magnitude decreased with FLX and increased EXT during stair ascent.

In the current study increased trunk flexion resulted in a 38% decrease in peak PFJS whereas trunk extension led to a 49% increase in PFJS during stair ascent. This finding is consistent with Teng et al. who reported that increased trunk flexion and extension while running resulted in a 6% decrease and 7% increase in peak PFJS respectively. Albeit a similar relationship, there are notable differences between studies in the magnitude of changes in peak PFJS. This difference is likely attributable to differences in the magnitude of changes in trunk position between studies. For example, in the current study, the increase in average trunk flexion from SS to FLX was approximately 5 times that reported by Teng et al. [20]. It is possible that the increased demands associated with running compared to stair ascent constrained the magnitude of changes in trunk position that were available. This is consistent with the premise that the availability of movement strategies is reduced as task demands increase [27]. Nonetheless, findings from both studies suggest the relationship between trunk posture and PFJS (for running and stair ascent) may be linear. In other words, the greater the degree of trunk flexion, the greater the decrease in peak PFJS and vice versa. This is important as some patients may not be able to assume the amount of increased trunk flexion observed in this study. It is possible that a smaller degree of trunk flexion during stair ascent may sufficiently reduce PFJS. This premise is consistent with another finding by Teng et al. who reported that forward trunk flexion was inversely correlated ($r = -0.06$, $p = 0.002$) with peak PFJS during running [20].

The changes in peak PFJS observed in this study can be largely attributed to changes in the PFJRF as opposed to changes in PFJ contact area. Specifically, in comparison to the SS condition, the PFJRF (at the time of peak PFJS) decreased by 40% during FLX and increased by 50% during EXT whereas PFJ contact area varied no more than 2% during FLX and EXT conditions when compared to SS. Furthermore, the observed changes in PFJRF resulted from changes in both the net knee extension moment and the knee flexion angle. Specifically, the knee extension moment and knee flexion angle decreased during FLX and increased during EXT (Table 1).

The FLX and EXT conditions led to a decrease and increase in the rate of PFJS by 35% and 27% respectively, compared to SS. This decrease in the PFJS rate during FLX, was due to a 37% decrease in the rate of PFJRF development as PFJ contact area decreased more rapidly during FLX versus SS. In contrast, the increased rate of PFJS development observed during EXT was the result of more rapid rates of PFJRF increase (28%) and a PFJ contact area decrease (95%).

The findings from this study have several important clinical implications. Previous research indicates that PFJ loading rate and magnitude are independent predictors of changes in PFP [15] and a 1.0 MPa decrease in peak PFJS has been associated with a 56% decrease in average self-reported PFP [18]. Additionally, running retraining for PFP patients that resulted in reduced lower limb loading rates has been associated with improved symptoms [16,17]. Thus, it is conceivable that the decreased PFJS observed during FLX would be effective for minimizing PFP symptoms during stair ascent. Furthermore, patellofemoral osteoarthritis has been associated with increased PFJS [28] and while it has yet to be tested, it is possible that decreased PFJS may reduce the risk for development or progression of this condition. Additionally, the increased PFJS associated with EXT may be used clinically to identify individuals exposed to increased PFJ loads. However, additional research is needed that examines associations between stair ascent trunk posture and PFP symptoms.

This study has several limitations. First, PFJS was estimated using a two-dimensional PFJ model that did not account for frontal and transverse plane knee kinematics which could influence PFJ mechanics [29]. Second, this study consisted of asymptomatic subjects and extrapolation of these findings to patients should not be done. Furthermore, it is unknown if sagittal trunk posture influences loading of joints or body segments besides the knee. For example, altered sagittal trunk posture may increase the risk for developing other musculoskeletal disorders such as low back pain or Achilles tendinopathy. Finally, only the first step was examined in this study which represents an adaptation from level walking to stair ascent and it is unknown if the relationships observed in this study exist during subsequent steps.

Table 2
Average Rate of Change of Biomechanical Variables from Initial Contact to Time of Peak Patellofemoral Joint Stress (mean ± SD).

	Flexed	Self-Selected	Extended	Self-Selected - Flexed		Self-Selected - Extended	
				Mean Difference (95% CI)	Effect Size (Cohen's d)	Mean Difference (95% CI)	Effect Size (Cohen's d)
Patellofemoral Joint Reaction Force Rate of Change (N/kg/sec)	122.0 ± 50.4	192.4 ± 56.8	246.3 ± 50.0	70.4 (53.3 to 87.5)	**3.5**	−53.9 (-75.3 to -32.5)	**−2.1**
Patellofemoral Joint Contact Area Rate of change (mm^2/sec)	−25.9 ± 16.2	−22.7 ± 20.7	−44.3 ± 42.4	3.2 (-8.4 to 14.8)	0.2	21.6 (-10.4 to 53.6)	0.5
Net Knee Extension Moment Rate of Change (N·m/kg/sec)	5.3 ± 2.1	8.1 ± 2.2	9.1 ± 1.9	2.8 (2.2 to 3.3)	**4.1**	−1.0 (-1.9 to -0.2)	**−1.0**
Knee Flexion Angle Rate of Change (°/sec)	−31.1 ± 21.1	−60.2 ± 14.5	−76.6 ± 23.3	−29.1 (-38.5 to -19.7)	**−2.7**	16.4 (5.4 to 27.5)	**1.3**

Negative mean values indicate a decreasing rate. Negative mean difference/effect sizeindicates a more rapid rate compared to Self-Selected condition. Bold effect size indicates significant difference between respective conditions ($p < 0.017$).

5. Conclusion

Changing stair ascent trunk posture affected PFJS rate and magnitude in asymptomatic females. Specifically, FLX and EXT were associated with decreased and increased PFJS magnitude and rate respectively. Future studies should examine whether these changes affect symptoms in individuals with PFP.

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CRediT authorship contribution statement

Lee T. Atkins: Conceptualization, Methodology, Software, Formal analysis, Investigation, Writing - original draft, Writing - review & editing, Visualization, Supervision, Project administration, Data curation, Resources, Validation. **Cevan Smithson:** Conceptualization, Methodology, Writing - review & editing, Data curation, Investigation, Project administration, Software, Writing - original draft. **Daniel Grimes:** Conceptualization, Methodology, Writing - review & editing, Data curation, Investigation, Project administration, Writing - original draft. **Nancy Heuer:** Conceptualization, Methodology, Writing - review & editing, Data curation, Investigation, Project administration, Writing - original draft.

Declaration of Competing Interest

The authors have no conflicts of interest to disclose.

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