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The influence of patella alta on patellofemoral joint stress during normal and fast walking

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Abstract

Objective: To determine if persons with patella alta exhibit elevated patellofemoral joint stress compared to pain-free controls during normal and fast walking speeds.

Subjects: Twenty-four subjects (13 patella alta, 11 pain-free controls) participated.

Methods: Sagittal and axial magnetic resonance images of the knee were obtained to quantify subject specific knee extensor mechanics and patellofemoral joint contact area. Instrumented gait analysis was used to quantify knee joint kinematics and kinetics. MRI and gait data were used as input variables into a model of patellofemoral joint stress. Analysis of variance with repeated measures was used to compare group differences and group × gait speed interactions for each dependent variable during stance.

Results: During normal speed gait there were no group differences in peak knee flexion angle, knee extensor moment, joint reaction force, or stress. However, the patella alta group had significantly less contact area. During fast speed gait there were no group differences in peak knee flexion angle, knee extensor moment, or joint reaction force. However, the patella alta group demonstrated significantly less contact area and significantly greater stress compared to controls.

Conclusion: Persons with patella alta demonstrated greater calculated patellofemoral stress during fast walking. This was the result of reductions in contact area as joint reaction forces were similar between groups.

Relevance

Persons with patella alta may be predisposed to patellofemoral dysfunction through elevations in joint stress. Therefore, treatments aimed at increasing the load-bearing surface area between the patella and femur, such as bracing, may be beneficial in this patient population.

Keywords: Patella alta; Patellofemoral joint; Patellofemoral stress

1. Introduction

Patella alta is a condition characterized by a superior patellar position relative to the trochlear groove of the femur (i.e. high riding patella) (Insall and Salvati, 1971). This condition has been associated clinically with patellofemoral dysfunction (Insall et al., 1972) and is considered a predisposing factor for the development of patellofemoral pain (PFP) (Kujala et al., 1986). Although the etiology of PFP is frequently debated, a commonly proposed hypothesis is related to elevated patellofemoral joint (PFJ) stress (force per unit area). Elevated PFJ stress is believed to lead to articular cartilage degeneration (Moller et al., 1989; Heino and Powers, 2002) and subsequent pain.

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It has been suggested that persons with patella alta have altered knee extensor mechanics that predispose such individuals to elevated PFJ reaction forces (Yamaguchi and Zajac, 1989; Singerman et al., 1994). Additionally, investigators have theorized that increased PFJ malalignment, as a result of patella alta, may predispose these individuals to diminished PFJ contact area (Kannus, 1992). Despite the fact that the combination of elevated PFJ reaction forces and reductions in PFJ contact area would appear to contribute to elevated PFJ stress in this population, such an assumption has not been tested experimentally.

Models of PFJ stress have been used to compare persons with PFP and pain-free controls (Heino and Powers, 2002; Brechter and Powers, 2002). Although subject specific PFJ contact areas were obtained in these investigations, extensor mechanism variables were obtained from the literature. However, the use of normative data in persons with patella alta may not be appropriate as previous studies suggest that the knee extensor lever arms and the relationship between quadriceps force and joint reaction force may be different in this population when compared to persons with normal patellar position (Yamaguchi and Zajac, 1989; Singerman et al., 1994).

The purpose of this study was to test the hypothesis that subjects with patella alta would exhibit elevated PFJ stress during level walking. In order to quantify PFJ stress in this population, an imaging based, subject specific model of PFJ stress was used.

2. Methods

2.1. Subjects

Twenty-four subjects (22 female and 2 male) between the ages of 19 and 34 participated in this study. Based on the procedures outlined below, 13 were determined to have patella alta and 11 had normal patellar position. Ten of the 13 subjects with patella alta (77%) had a recent history of pain originating from the patellofemoral articulation, while none of the subjects in the control group had a history of pain. Subject groups were similar in terms of height and weight (Table 1). An a priori power analysis revealed that 10 subjects per groups were needed to detect a 20% difference in patellofemoral joint stress between populations (80% power, $\alpha = 0.05$).

Table 1
Subject characteristics

Mean (SD)	Control ($n = 11$)	Patella alta ($n = 13$)
Age (years)[a]	28.4 (4.3)	25.0 (3.5)
Height (cm)	164.4 (5.3)	162.9 (4.8)
Weight (kg)	57.2 (7.2)	58.9 (9.6)
Insall–Salvati index (L_{pl}/L_p)[a]	1.02 (0.08)	1.32 (0.08)

[a] indicates significant differences between groups.

Subjects were screened by physical examination to rule out the presence of tibiofemoral instability or meniscal injury. Additionally, subjects were excluded if they reported previous knee surgery or any implanted biological devices, such as pacemakers, cochlear implants, or clips which could interact with the magnetic field during imaging. Prior to participation all subjects were informed as to the nature of the study, procedures, and risks. Each subject signed a human subjects consent approved by the Institutional Review Board of the University of Southern California.

2.2. Instrumentation

2.2.1. Magnetic resonance imaging

Magnetic resonance images were obtained with a 1.5-TGE Signa scanner. Sagittal images of the knee were obtained using a T1 weighted spin echo pulse sequence (TR 350 ms, TE 10 ms, NEX 1, FOV $20\,cm \times 20\,cm$, matrix 256×256, slice thickness 10 mm, and two 5-inch receive only coils). Axial images of the patellofemoral joint were obtained using a fat suppressed FSPGR pulse sequence (TR 8.2 ms, TE 1.5 ms, NEX 1, spectral inversion for fat suppression, FOV $20\,cm \times 20\,cm$, matrix 512×512, slice thickness 2 mm, two 5-inch receive only coils).

2.2.2. Gait analysis

Three-dimensional kinematics of the lower extremity were acquired using a six-camera motion analysis system (Vicon 370, Plug-in-Gait model v1.7, Oxford Metrics Ltd, Oxford, UK) (Davis et al., 1990). Kinematic data from an 18 point marker set (20-mm spheres) were sampled at 60 Hz. Ground reaction forces were sampled at 600 Hz using four AMTI force plates (AMTI, model OR6-6-1, Newton, MA, USA).

2.3. Procedures

Each subject completed two phases of data collection. First, sagittal and axial MR images were obtained to determine subject specific knee extensor mechanics and PFJ contact area. Second, instrumented gait analysis was performed to determine sagittal plane kinematics and kinetics at the knee. Data from both testing sessions were used as input variables to a previously developed biomechanical model to estimate PFJ stress.

2.3.1. Magnetic resonance imaging

Prior to imaging, subjects completed a standard MRI safety screening form. Subjects were placed supine on the MR gantry and secured within a custom made non-ferromagnetic leg press apparatus (Captain Plastic, Shoreline, WA, USA). The lower extremity of interest was placed on the footplate of the loading device and two receive only coils were taped to the knee. Epoxy weights totaling 25% of bodyweight were then secured to the load carriage. Preliminary studies demonstrated that this load was sufficient to remove any slack in the knee extensor mechanism and was tolerable to patients with PFP.

Subjects were first positioned at 0° of knee flexion and were instructed to hold this position during imaging. Once the desired knee flexion angle was achieved (determined using a standard goniometer), the knee was landmarked, a triplanar scout scan was acquired, sagittal images of the knee were obtained, followed by axial imaging. This procedure was repeated at 20°, 40°, and 60° of knee flexion. The acquisition times for the sagittal and axial images were 120 s and 39 s respectively.

2.3.2. Gait analysis

Subjects were appropriately attired to permit placement of markers directly onto the skin. Anthropometric measurements of leg length, knee width, and ankle width were obtained bilaterally using calipers. Reflective markers were then placed on the following anatomical landmarks bilaterally; posterior superior iliac spine (PSIS), anterior superior iliac spine (ASIS), lateral thigh, lateral femoral epicondyle, lateral shank, lateral malleoli, second metatarsal heads, fifth metatarsal heads, and posterior calcanei. Following marker placement, a six second static trial was obtained.

Subjects were asked to walk along a 10-m walkway with the middle 6-m being used for data collection. Subjects were instructed to walk along the walkway at two pre-determined gait velocities, normal speed (85 m/min ± 5%) and fast speed (120 m/min ± 5%). Three trials were obtained at each gait speed. A successful trial was one in which the lower extremity of interest landed completely within one of the four force plates, without targeting.

2.4. Data analysis

2.4.1. Magnetic resonance imaging

Prior to analysis, all images were magnified (1.5×) and calibrated (pixel dimensions specified) using Scion Image software (Scion Corp., Frederick, MD, USA). To determine the presence of patella alta, images were screened to determine which contained the maximum patellar ligament length and patellar length. Measurements of patellar ligament length were made along the posterior surface from the tibial tuberosity to the patellar apex, while measurements of patellar length were made from the apex of the patella to the most posterior superior aspect of the patellar base (Fig. 1) (Insall and Salvati, 1971). In all cases, the longest patellar ligament length and longest patellar length were used even if they were on separate images in a series, as previous research has suggested that this method produces the strongest reliability and validity with measurements obtained from lateral radiographs of the knee (Miller et al., 1996). The length of the patellar ligament was then divided by the length of the patella to yield the Insall–Salvati index (Insall and Salvati, 1971). Insall–Salvati ratios greater than or equal to 1.2 indicated the presence of patella alta, while indices between 0.80 and 1.19 identified subjects as having normal patellar position (Insall and Salvati, 1971).

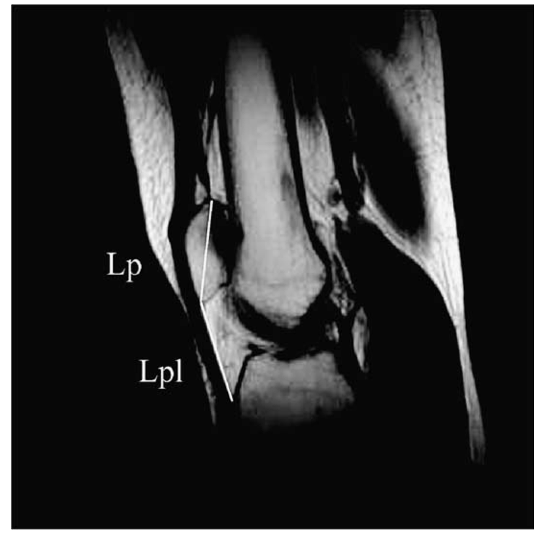

Fig. 1. The Insall–Salvati index was computed by dividing length of the patellar ligament (L_{pl}) by the longest diagonal length of the patella (L_p).

Sagittal images of the knee were then screened to determine which image contained the midsection of the knee. This was determined by identifying the image containing the intersection of the cruciate ligaments. This image was used to measure all four indices of knee extensor mechanics according to the methods described by Yamaguchi and Zajac (1989).

The actual moment arm (M_{act}) was measured as the perpendicular distance from the tibiofemoral axis of rotation, estimated at the intersection of the cruciate ligaments (O'Connor et al., 1989), to the patellar ligament (Fig. 2). The intersection of the cruciate ligaments was chosen to estimate the axis of rotation as previous literature has suggested that it produces smaller errors than using the tibiofemoral contact point when compared to the helical axis method (Baker and Ronsky, 2001).

Measurements of the quadriceps moment arm (M_q) and patellar ligament moment arm (M_{pl}) were measured as the perpendicular distances from the patellofemoral contact point to the respective tendons (Fig. 3). When

Fig. 2. The actual moment arm (M_{act}) was measured as the distance from the axis of rotation (C) to the patellar ligament (A–B).

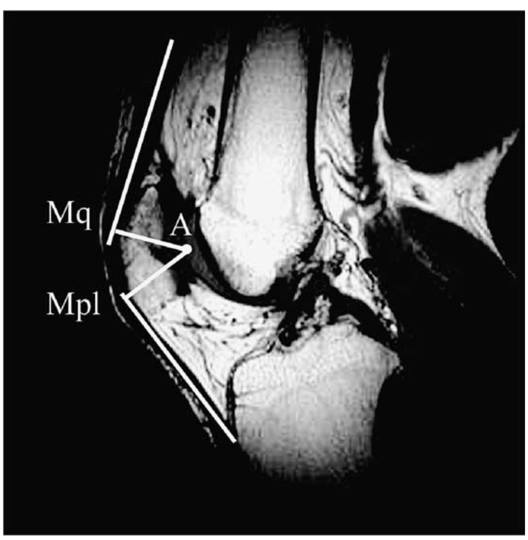

Fig. 3. The quadriceps tendon moment arm (M_q) and the patellar ligament moment arm (M_{pl}) were measured as the perpendicular distances from the patellofemoral contact point (A) to the lines of action of their respective tendons.

the patellofemoral contact point could not be identified by a single point, the line of contact between the patella and femur was bisected and the midpoint of this line was used as the axis of rotation. The force in the patellar ligament/quadriceps tendon force (F_{pl}/F_q) ratio was then computed using the following equation:

$$F_{pl}/F_q = M_q/M_{pl} \quad (1)$$

The quadriceps effective moment arm was then calculated using the following equation (van Eijden et al., 1986; Yamaguchi and Zajac, 1989):

$$M_{eff} = F_{pl}/F_q(M_{act}) \quad (2)$$

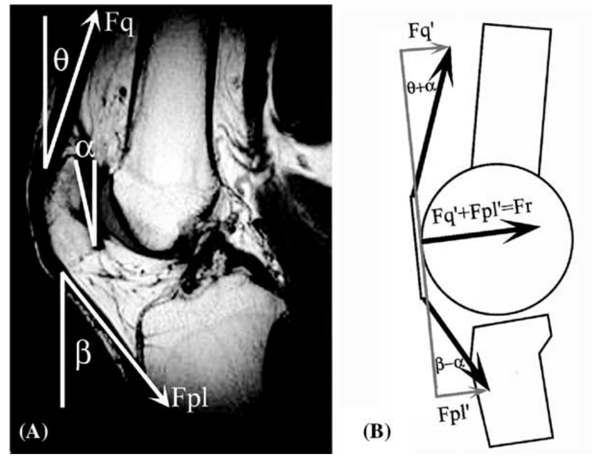

Fig. 4. (A) Measurements of the quadriceps tendon (θ), retropatellar surface (α), and patellar ligament (β) relative to vertical. (B) These angles were used to compute the joint reaction force (black line, F_r) from the quadriceps tendon force (gray line, F'_q) and the patellar ligament force (gray line, F'_{pl}).

To estimate how much force from the patellar ligament and quadriceps tendon would be directed towards compression, the angles the quadriceps tendon (θ), retropatellar surface (α), and patellar ligament (β) formed with vertical were measured (Fig. 4). Using the following equation, these angles and the F_{pl}/F_q ratio were used to calculate the ratio of patellofemoral joint reaction force to quadriceps force (F_r/F_q),

$$F_r = \frac{F_q[\sin(\theta+\alpha)] + [F_q(F_{pl}/F_q)][\sin(\beta-\alpha)]}{F_q} \quad (3)$$

where F_q is assigned a value of 1 N. This ratio represents the magnitude of the expected joint reaction force, in Newtons per unit quadriceps force.

PFJ contact area was measured from the sequential axial plane images of the patellofemoral joint obtained in supine, with the lower extremity loaded in a simulated weightbearing condition. Contact was defined as areas of patella and femur approximation in which no distinct separation could be found. A curvilinear line of contact between the patella and femur was drawn and then measured on each slice. The length of contact on each slice was then multiplied by the 2-mm slice thickness to yield an intraslice contact area. The areas of contact from each sequential image were summed to obtain a total PFJ contact area (Salsich et al., 2003; Brechter et al., 2003). All contact area measurements were reported in mm^2.

This MRI method to quantify PFJ contact area has been shown to be reliable and comparable to contact area measurements obtained using Fuji pressure sensitive film in cadaver specimens (Brechter et al., 2003). All contact area measurements were made twice by the same investigator and averaged for final analysis.

A custom written macro for Scion Image was used to measure all perpendicular distances and angles. All measurements were made twice by the same investigator and averaged for statistical analysis.

2.4.2. Gait analysis

Markers were identified manually and then automatically digitized using Vicon Workstation software (v 4.5). Marker trajectory data were filtered using a Woltring quintic spline function with a predicted mean square error of 20 mm (Woltring, 1986) and normalized to 100 equally spaced points representing 1 full stride. Sagittal plane joint angles were calculated as reported by Davis et al. (1990) and moments were calculated as described by Kadaba et al. (1987) using the Vicon Plug-in-Gait model (v 1.7). Angles were reported in degrees and net joint moments were reported in N m/kg to facilitate comparison between groups. The average of three trials was used for statistical analysis.

2.4.3. Patellofemoral model

A previously described model of the PFJ was used to estimate stress during walking (Heino and Powers, 2002; Brechter and Powers, 2002). This model was modified to allow subject specific knee extensor mechanics to be used as input variables. Input variables for the model were; knee joint moment, knee joint angle, PFJ contact area, quadriceps effective moment arm (M_{eff}), and the joint reaction force/quadriceps force (F_r/F_q) ratio. Matlab was used as the computational engine for the model steps described below.

Prior to running the model, a shape-preserving cubic spline function (Curvefitting Toolbox, Matlab 6.5, Natick, MA, USA) was fit to the contact area, M_{eff}, and F_r/F_q ratio data. Next, contact area, M_{eff}, and F_r/F_q ratio data were interpolated as a function of knee flexion angle.

The first step of the model estimated quadriceps force (F_q). This was calculated by dividing the net knee extensor moment (non-normalized) by M_{eff} for each time point in the gait cycle (i) (Eq. (4)):

$$F_{qi} = \text{Knee extensor moment}_i / M_{\text{eff}i} \quad (4)$$

The second step of the model estimated joint reaction force (F_r). This was calculated by multiplying quadriceps force (F_{qi}) by the joint reaction force/quadriceps force (F_r/F_q) ratio at each time point in the gait cycle (i) (Eq. (5)):

$$F_{ri} = F_{qi}(F_r/F_q)_i \quad (5)$$

The third step of the model estimated PFJ stress (PFJS). This was calculated by dividing joint reaction force (F_{ri}) by PFJ contact area (CA) at each time point in the gait cycle (i) (Eq. (6)):

$$\text{PFJS}_i = F_{ri}/\text{CA}_i \quad (6)$$

The model outputs were PFJ reaction force, utilized contact area, and PFJ stress, all expressed as a percentage of the gait cycle.

2.5. Statistical analysis

Initial statistical analysis included the Shapiro-Wilk's and Levene's tests to screen the data for assumptions of normality and homogeneity of variances. A 2 × 2 (group × gait velocity) ANOVA with repeated measures were used to test for main effects and interactions between group and gait speed. This analysis was repeated for each dependent variable. The variables of interest in this study were the knee flexion angle, knee extensor moment, PFJ reaction force, and PFJ contact area that corresponded with the peak stress value during the stance phase of the gait cycle.

Differences between groups and group × gait speed interactions were of interest in this study. In the event of a significant interaction, post-hoc Tukey's tests were used to determine where differences existed between groups. All statistical analyses were performed using SPSS statistical software with a significance level of $P < 0.05$.

3. Results

There was no significant group effect or interaction for peak knee flexion angle (Fig. 5) or peak knee exten-

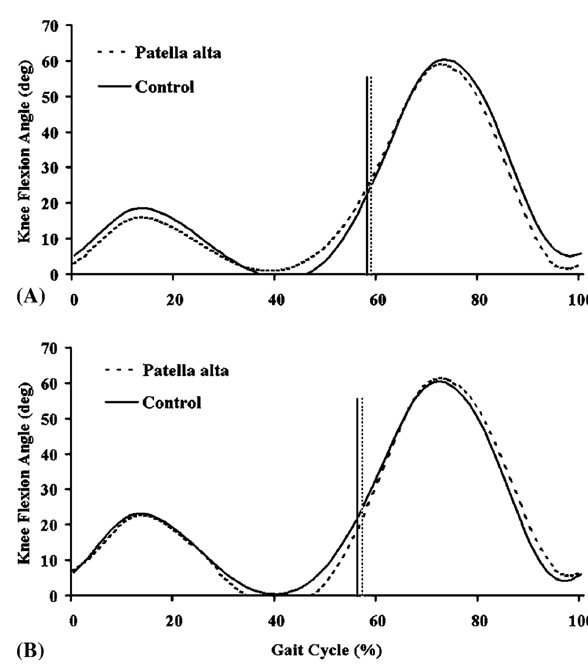

Fig. 5. Comparison of knee flexion angle between the patella alta and control groups during normal (A) and fast (B) gait speeds. Vertical lines separate stance and swing for each group. No significant differences were observed between groups.

sor moment (Fig. 6) during stance. Similarly, there was no significant group effect or interaction for peak PFJ

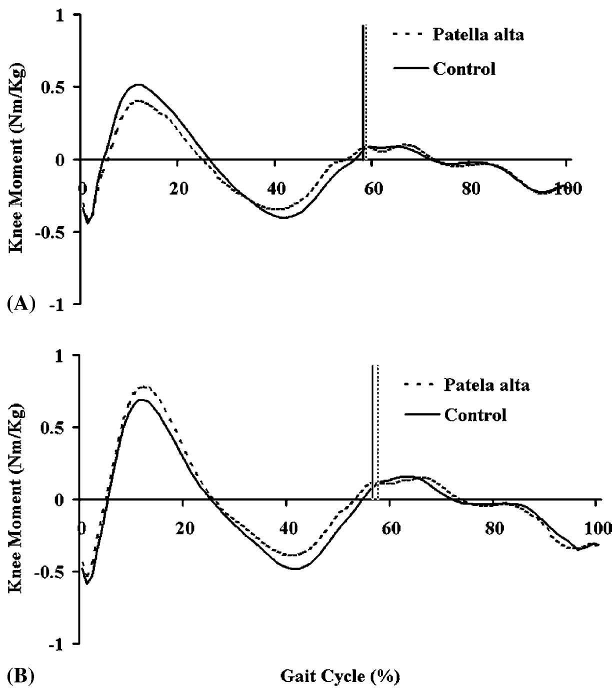

Fig. 6. Comparison of knee extensor moment between the patella alta and control groups during normal (A) and fast (B) gait speeds. Vertical lines separate stance and swing for each group. No significant differences were observed between groups.

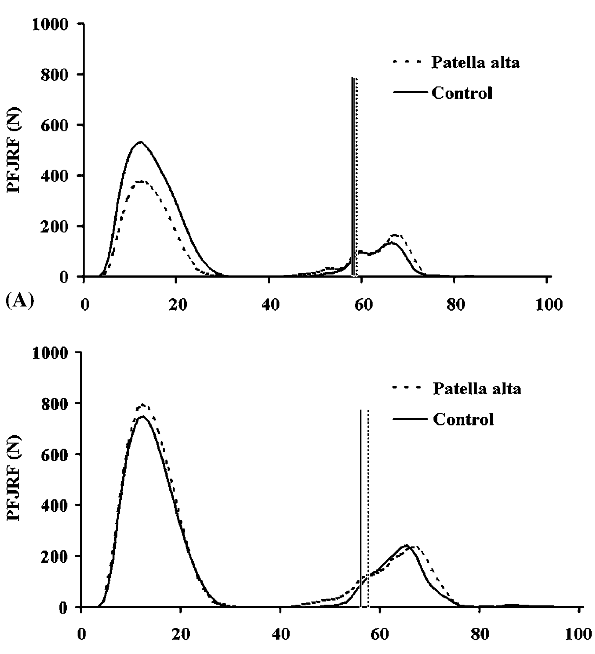

Fig. 7. Comparison of patellofemoral joint reaction force between the patella alta and control groups during normal (A) and fast (B) gait speeds. Vertical lines separate stance and swing for each group. No significant differences were observed between groups.

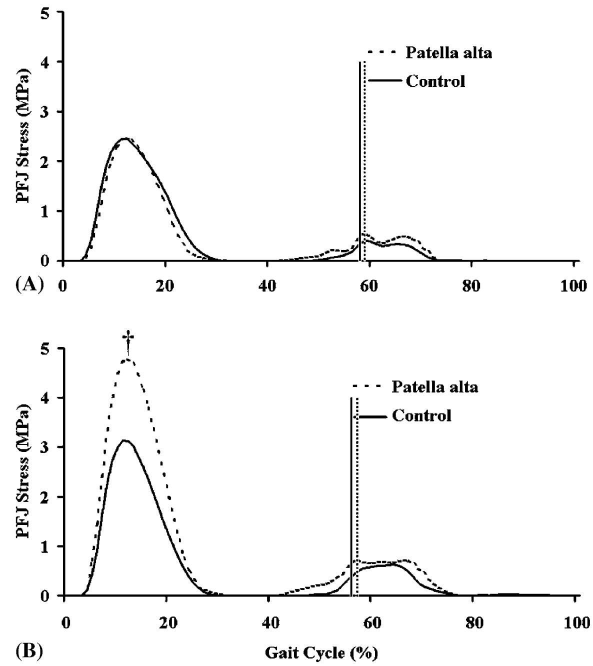

Fig. 8. Comparison of patellofemoral joint stress between the patella alta and control groups during normal (A) and fast (B) gait speeds. Vertical lines separate stance and swing for each group. † indicates a significant difference in peak patellofemoral joint stress.

reaction force (Fig. 7). However, subjects with patella alta demonstrated a trend towards decreased peak PFJ reaction force during the normal speed gait (409.5 N (SD, 175.7) vs. 506.3 N (SD, 248.8)) compared to the control group.

There was a significant group effect for the utilized contact area during stance ($F_{1,22} = 5.314, P < 0.05$) and no group × gait speed interaction. Subjects with patella alta demonstrated reduced contact areas at the point of peak stress during the normal gait speed (163.1 mm^2 (SD, 54.8) vs. 212.2 mm^2 (SD, 50.5)) and fast gait speed (178.4 mm^2 (SD, 66.0) vs. 229.3 mm^2 (SD, 49.2)) compared to the control group.

There was no group effect for PFJ stress, however there was a significant group × gait speed interaction ($F_{1,22} = 8.158$, $P < 0.05$) (Fig. 8). Post-hoc testing revealed that subjects with patella alta demonstrated elevated PFJ stress during the fast gait speed (4.80 MPa (SD, 1.33) vs. 3.12 MPa (SD, 1.34), $P < 0.05$) but did not have elevated PFJ stress during the normal gait speed (2.68 MPa (1.10) vs. 2.37 MPa (SD, 0.93)).

4. Discussion

The results of this study found that subjects with patella alta demonstrated higher peak PFJ stress during fast walking speeds. The increase in PFJ stress during fast walking was the result of reduced utilized contact area as no significant difference in PFJ reaction force was observed. Given that PFP has been linked to increases in PFJ stress (Grana and Kriegshauser, 1985; Moller et al., 1989; Heino and Powers, 2002), these data support the clinical observation that subjects with patella alta may be predisposed to the development of PFP (Kujala et al., 1986).

In contrast to fast walking, no significant differences in PFJ stress were observed during normal walking speeds. Although persons with patella alta demonstrated a reduction in contact area during the slower walking speed, this did not result in elevated PFJ stress. The lack of a difference in peak stress during this task can be explained by the trend towards reduced PFJ reaction force in the patella alta group. This reduction in PFJ reaction force was due to the combined effect of a smaller knee flexion angle and reduced knee extensor moment. Although not statistically significant, the tendency towards a reduced knee flexion angle and knee extensor moment in the patella alta group may have been a compensatory strategy to reduce PFJ reaction force and stress during normal walking speeds. This premise is consistent with the data of Salsich et al., who reported similar findings in a population of persons with PFP during stair ambulation (Salsich et al., 2001). The fact that this pattern was not observed during the fast gait speed suggests a decreased ability to compensate during higher demand tasks.

Previous investigations have suggested that the presence of patella alta may influence the mechanics of the knee extensor mechanism in a way that may predispose these individuals to elevated joint reaction forces (Yamaguchi and Zajac, 1989; Singerman et al., 1994). The results of this study do not support this premise as no significant differences in PFJ reaction forces were observed. In fact, persons with patella alta had a 21% reduction in PFJ reaction force during the normal gait speed and only a 5% increase in PFJ reaction force during the fast gait speed. It should be noted that previous studies suggesting a relationship between patella alta and increased PFJ reaction force utilized mathematical modeling and cadaver specimens (Yamaguchi and Zajac, 1989; Singerman et al., 1994). The results of the current study indicate that extrapolation of these data to an in vivo population may not be valid.

During the normal walking speed, subjects with patella alta had 23% less available contact area over which to distribute the PFJ reaction force. Similarly, these individuals had 22% less available contact area during the fast walking speed. Analysis of the PFJ contact area that was input into the stress model revealed that subjects with patella alta had a systematic reduction in contact area from 0° to 60° of knee flexion compared to those without patella alta. This observation is consistent with the hypothesis proposed by Kannus (1992) who predicted smaller PFJ contact areas in subjects with patella alta on the basis of increased patellofemoral malalignment. It is also in agreement with the data of Heino and Powers (2002), who demonstrated that elevated PFJ stress in subjects with PFP pain was due to reductions in contact area.

There are several limitations of this study that should be noted. First, the model used assumed a planar representation of the PFJ for calculation of PFJ stress. Although the error associated with this assumption is not know, any error would be consistent across both groups, making comparisons between populations possible. Secondly, our model did not account for the potential effects of co-contraction at the knee (i.e. simultaneous hamstring or gastrocnemius activity with quadriceps activity). As a result, the absolute values for the quadriceps force and PFJ kinetics would likely be underestimated. Thirdly, our model assumes that the relationship between the patella and femur obtained in supine during imaging is the same as the relationship between these structures during gait. Although the error associated with this assumption is also unknown, an apparatus was used during the imaging process that loads the knee in a simulated weightbearing condition in an attempt to minimize such errors (Salsich et al., 2003).

In summary, persons with patella alta demonstrated increased patellofemoral joint stress during fast walking speeds. The primary cause of elevated patellofemoral joint stress was a reduction in patellofemoral joint contact area and not an increase in patellofemoral joint reaction force. Therefore, interventions aimed at increasing patellofemoral joint contact area (i.e. bracing) may be efficacious in this patient population.

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