Full length article

The effects of pediatric obesity on patellofemoral joint contact force during walking

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ABSTRACT

Background: Obesity increases a child's risk of developing knee pain across the lifespan, potentially through elevated patellofemoral joint loads that occur during habitual weight-bearing activities.
Research question: Do obese children have greater absolute and patellar-area-normalized patellofemoral joint forces compared to healthy weight children during walking?
Methods: We utilized a cross-sectional design to address the aims of this study. Experimental biomechanics data were collected during treadmill walking in 10 healthy-weight and 10 obese 8–12 year-olds. We used radiographic images to develop subject-specific musculoskeletal models, generated walking simulations from the experimental data, and predicted patellofemoral joint contact force using established techniques.
Results: We found that the obese children had 1.98 times greater absolute (p = 0.002) and 1.81 times greater patellar-area-normalized (p = 0.008) patellofemoral joint contact forces compared to the healthy-weight children. We observed a stronger relationship between absolute patellofemoral joint contact force and BMI ($r^2 = 0.58$) than between patellofemoral joint contact force and body fat percentage ($r^2 = 0.38$).
Significance: Our results indicate that obese children walk with increased patellofemoral loads in absolute terms and also relative to the area of the articulating surfaces, which likely contributes to the increased risk of knee pain in this pediatric population. This information, which provides a baseline comparison for future longitudinal studies, also informs the type and frequency of physical activity prescription aimed at reducing the risk of knee injury and improving long-term outcomes.

1. Introduction

Pediatric obesity remains one of the top public health concerns of the 21st century. One in five children currently suffers from obesity in the United States [1]. Being obese increases a child's risk of developing life-long pain and orthopedic disorders of the lower-extremity [2], particularly of the knee joint [3,4]. Moreover, knee pain and injury during childhood growth is positively associated with weight gain [5] and reduced physical function [6]. The health and development of a child's articulating joint surfaces and growth plates have health implications across the lifespan.

The mechanical loading environment encountered during activities of daily living affects joint health and function [7]. Walking is the most common form of daily physical activity and a fundamental component in the treatment of pediatric obesity [8]. Larger body mass is associated with greater ground reaction forces, joint moments, and tibiofemoral joint contact forces during walking [9,10]. Knowledge of how pediatric obesity affects joint loads during walking may inform the prescription of physical activity and contribute to understanding the mechanisms of musculoskeletal disorders of the knee joint in this population.

The patella plays an important role in lower-extremity movements by increasing the mechanical advantage of the quadriceps muscles, and the patellofemoral joint is the primary source of symptoms associated with knee pain [11]. Peak patellofemoral joint contact forces have been reported in the range of 0.5–1.5 times body-weight during walking in healthy adults, with loads up to 2–3 times body-weight during daily activities involving high knee flexion [11,12]. Based on current literature, children with excess body mass and body fat percentage may have greater patellofemoral joint loading both in absolute terms and also relative to the area of the articulating surface. Such joint loading information may aid the prevention or treatment of childhood obesity through better informed and safer prescription of physical activity.

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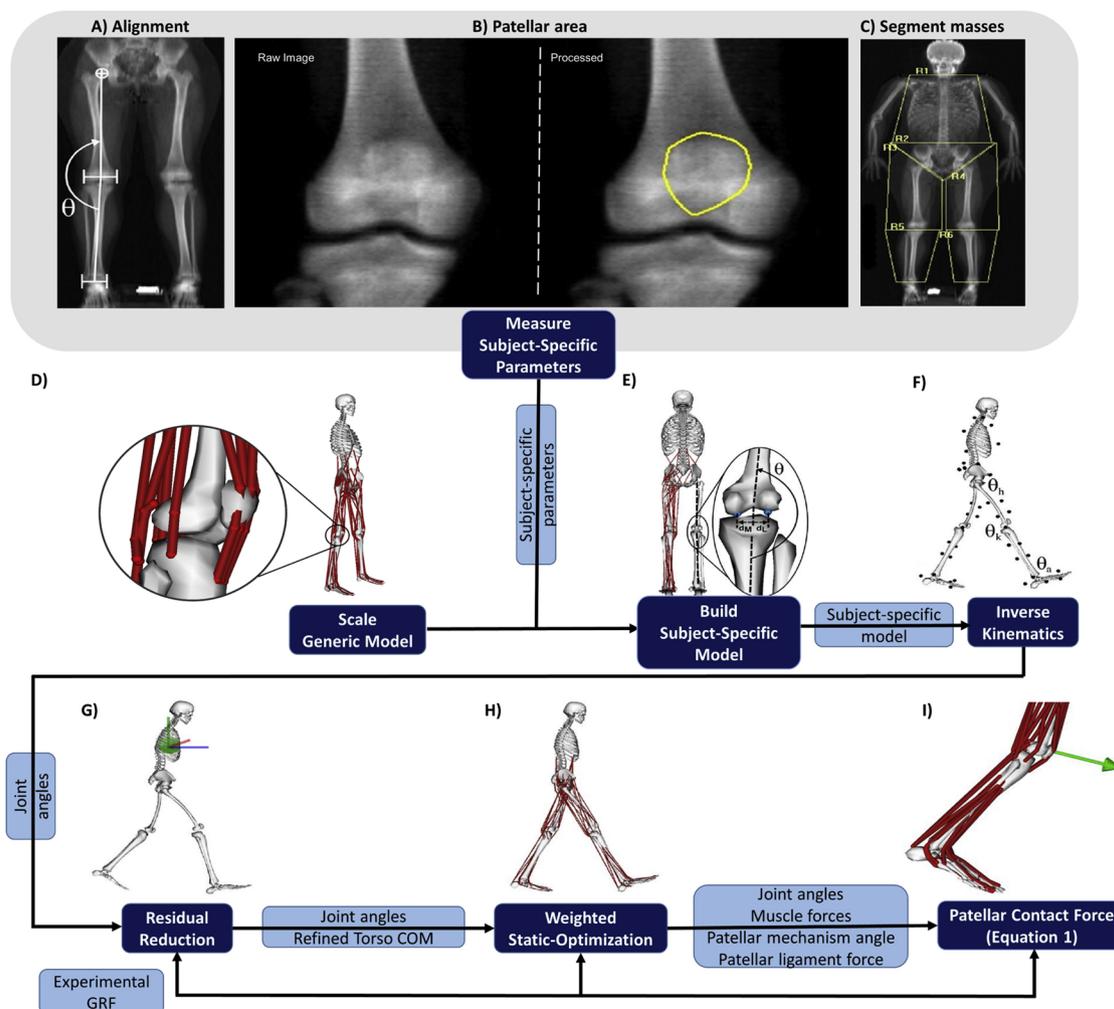*E-mail address:* Zachary.Lerner@nau.edu (Z.F. Lerner).

Fig. 1. Schematic depiction of the musculoskeletal modeling workflow used to calculate patellofemoral joint contact forces. (A), (B), and (C) represent measured subject-specific parameters. (D) Generic musculoskeletal model used in the study. Tibiofemoral joints and patellofemoral joints were modeled as planar joints with translations and rotations coupled to the knee flexion angle. Quadriceps forces were transmitted through the patella to the tibia. (E), (F), (G), (H), and (I) represent the process of computational analysis to estimate patellofemoral joint contact forces via Eq. 1.

The purpose of this study was to compare patellofemoral joint contact forces between obese children and healthy weight children during walking. Our first hypothesis was that obese children would exhibit larger absolute and patella-area-normalized patellofemoral joint contact forces during walking compared to healthy-weight children. Our second hypothesis was that body fat percentage and body mass index (BMI) would be directly related to absolute and patellar-area-normalized patellofemoral joint contact forces averaged across the gait cycle. We tested these hypotheses by developing personalized musculoskeletal models and conducting motion analysis during walking in obese and healthy-weight children.

2. Methods

2.1. Participants

Prior to data collection, all subjects provided verbal assent while their parents provided written informed consent. The Institutional Review Board at the Colorado State University approved this study. Twenty individuals between the ages of 8 and 12 participated in this study (Supplemental Table 1). Ten participants were obese (6 male) with a BMI z-score greater than the 95th percentile and 10 participants were of healthy-weight (5 male). Exclusion criteria for all groups included a history of musculoskeletal, neuromuscular, or cardiovascular

disorder, or any other disorder that would preclude safe participation in this study. We quantified body composition of each participant using dual x-ray absorptiometry (DXA, Whole-Body Scan, Hologic Discover, Bedford, MA). A custom foot placement jig was used to passively maintain the frontal-plane orientation of each participant's lower extremity. A higher resolution scan (1 mm point resolution) was taken of each participant's right knee.

2.2. Experimental walking protocol

Subjects walked on an instrumented treadmill (Bertec Corp, Columbus, OH) at a constant rate (1.0 m/s) that was close to the preferred speeds reported in children [13]. Participants walked for 5 min prior to data collection allowing time to acclimate to treadmill locomotion; data reported in this study were collected during the 6th min of walking. Ground reaction force data from the instrumented treadmill was recorded at 1000 Hz. A three-dimensional motion capture system (10 cameras, Nexus, Vicon, Centennial, CO) recorded marker trajectories at 100 Hz. An obesity-specific marker set was used for the collection of kinematic data [14]. Reflective markers were placed on the 7th cervical vertebrae, acromion processes, right scapular inferior angle, sternoclavicular notch, xyphoid process, 10th thoracic vertebrae, posterior-superior iliac spines, medial and lateral epicondyles of femurs, medial and lateral malleolus, calcanei, first metatarsal heads,

second metatarsal heads, and proximal and distal heads of the 5th metatarsals. We used a digitizing pointer to mark the skeletal locations of the anterior superior iliac spines (ASIS) and iliac crests by probing through overlying soft-tissue. Marker clusters, four non-collinear markers on a small body-worn plate, were attached to the sacrum, thighs, and shanks. Post-processing (Visual 3D, C-Motion, Germantown, MA) was used to account for adiposity surrounding the pelvis by defining the digital ASIS and iliac crest landmarks relative to the sacral cluster and generating virtual markers for model scaling and motion tracking. Force plate data were low-pass filtered at 12 Hz while marker data were low-pass filtered at 5 Hz, both using a 4th order zero-lag Butterworth filter.

2.3. Personalized model building

We used a previously reported full-body musculoskeletal gait model [15] for this investigation in OpenSim 3.3 [16]. The model, which has been used to examine joint loading in children [10,17], had 18 body segments, 21 degrees of freedom, and 92 muscle-tendon actuators. The knee joint included a patella that articulated along the femoral trochlea with motion that was coupled with knee flexion-extension [18]. As first reported in [19], the quadriceps muscles passed through the patella and inserted on the tibial tuberosity, replicating the function of the patellar ligament. A tibial plateau body and a distal femoral component body were included in this model, which enabled us to designate the frontal-plane knee alignment while maintaining the translations of the tibia and patella relative to the femur as well as sagittal plane rotation of the knee.

The musculoskeletal model was scaled for each participant using the positions of the reflective markers located on the aforementioned anatomical and digital landmarks during a static standing trial. Muscle attachments, muscle moment arms, segment inertial properties, and muscle length properties were scaled accordingly. Next, each scaled model was customized to include lower-extremity alignment and segment masses measured from each participant's DXA radiographs using standard techniques [20]; subject-specific model building was reported in greater detail in [15]. Mass properties of the head, arms and torso were included in a combined head-torso model segment (Fig. 1).

2.4. Prediction of muscle and joint contact forces

We used OpenSim to simulate each participant's walking trial and compute joint contact forces (Fig. 1). Joint angles were calculated using OpenSim's Inverse Kinematics analysis. Residual Reduction Algorithm (RRA) was used to maximize the dynamic consistency between model kinematics and kinetics by adjusting the torso's mass and center of mass location of each subject-specific model. Following adjustment, average residuals were less than 6% body-weight for all participants. We found that residuals were not significantly affected by group. The muscle redundancy problem was solved using a static optimization approach to predict individual muscle forces. The objective function of the optimization scheme minimized the sum of squared muscle activations. Individual muscle weighting constants were applied to the objective function to refine muscle force predictions based on our validation procedure (below) using in-vivo measurements from an instrumented joint implant (Lerner et al., 2015b; Steele et al., 2012).

Patellofemoral joint contact forces (PFJCF) were computed using the approach from [21], based on the equation from [22]:

$$PFJCF = \sqrt{F_Q^2 + F_p^2 + 2F_Q F_p \cos\gamma} \qquad (1)$$

where F_Q is the quadriceps force, F_p is the patellar ligament force, and γ is the angle between the quadriceps muscle and the patellar ligament (patellar mechanism angle). Quadriceps muscle force and knee flexion angle were obtained from OpenSim. Experimentally-determined force data from [23] were used to quantify patellar ligament force (F_p) as a

function of quadriceps force and knee angle, and patellar mechanism angle (γ) as a function of knee angle, as in [21].

We used ImageJ, medical imaging analysis software, to trace the perimeter of the right patella on each subject's knee radiograph (Fig. 1B). Planar patellar area was calculated as the area enclosed by perimeter of the patella. We divided patellarfemoral contact force by the planar patellar area to approximate the contact force relative to the size of the articulating surface and obtain patellar-area-normalized contact force.

In addition to our investigation on pediatric obesity, we also evaluated the importance of modeling the loading dynamics of the patellar ligament when computing patellofemoral joint contact forces. We calculated patellofemoral loads directly using OpenSim's Joint Reaction analysis, and compared them to the approach from Fok et al. (Eq. 1). Joint Reaction analysis computes the resultant bone-on-bone joint contact force acting on the patellofemoral joint due to the muscle, inertial, and external loads applied to the model. The primary difference between approaches was that patellar ligament force was equal to the quadriceps force in our OpenSim calculation, which ignored the mechanical properties and loading response of the patellar ligament. However, quadriceps force and patellar ligament force are different depending on knee flexion angle. Therefore, we employed the approach from Fok et al. to account for the mechanical properties and loading response of the patellar ligament.

2.5. Validation of model and optimization scheme

The musculoskeletal model and optimization scheme used in this study was previously validated using in-vivo data from a subject with a load cell embedded within the tibial insert of their knee endoprosthesis [24]. Muscle weighting factors used in static optimization were established to minimize the difference between measured and estimated knee joint contact force; factors included 1.5 for the gastrocnemii, 2 for the hamstrings, and 1 for all other muscles [15]. The model and weighting factors used in this study were previously used to investigate knee (tibiofemoral) and hip joint contact forces in the same cohort as in this study [10,17]. We previously ensured that the muscle activation patterns estimated from static optimization were similar to experimentally measured muscle activation patterns (electromyography) from these exact walking trials as well as with electromyography data reported in the literature [10].

2.6. Statistical analysis

Data from three representative gait cycles were averaged for each subject. Outcome measures were averaged across participants in each group. Independent sample t-tests were used to test for differences in outcomes between the obese and healthy-weight cohorts, where $\alpha < 0.05$ indicated statistical significance. Linear regression analysis was used to determine the relationships between gait cycle average patellofemoral joint contact force and participant anthropometrics. R^2 values were interpreted as follows, weak: < 0.3; moderate: $0.3\text{-}0.5$; strong: > 0.5. Statistical Parametric Mapping [25] (SPM 1D version 0.4) independent *t*-test was used to compare knee angle between the two groups across the gait cycle. Matlab version 2017a (Mathworks Inc., Natick, MA) was used to perform data and statistical analysis.

3. Results

There were no significant differences in step length, and patella width and area between groups (Supplemental Table 1). Nor were there significant differences in knee flexion angles across the gait cycle (Supplemental Fig. 1). On the other hand, absolute and body-weight normalized quadriceps forces during walking were affected by pediatric obesity (Supplemental Fig. 2). Second peak absolute ($p < 0.001$) and body-weight normalized ($p < 0.001$) quadriceps force was 2.10 and

Table 1
Absolute and body-weight normalized peak quadriceps force.

	First peak (N)			Second peak (N)		
	Healthy-weight	Obese	p	Healthy-weight	Obese	p
Absolute (N)	262.61 (127.97)	386.59 (248.69)	0.178	119.83 (63.54)	252.08 (75.40)	< **0.001**
Normalized (BW)	0.85 (0.40)	1.33 (0.96)	0.156	0.38 (0.15)	0.84 (0.28)	< **0.001**

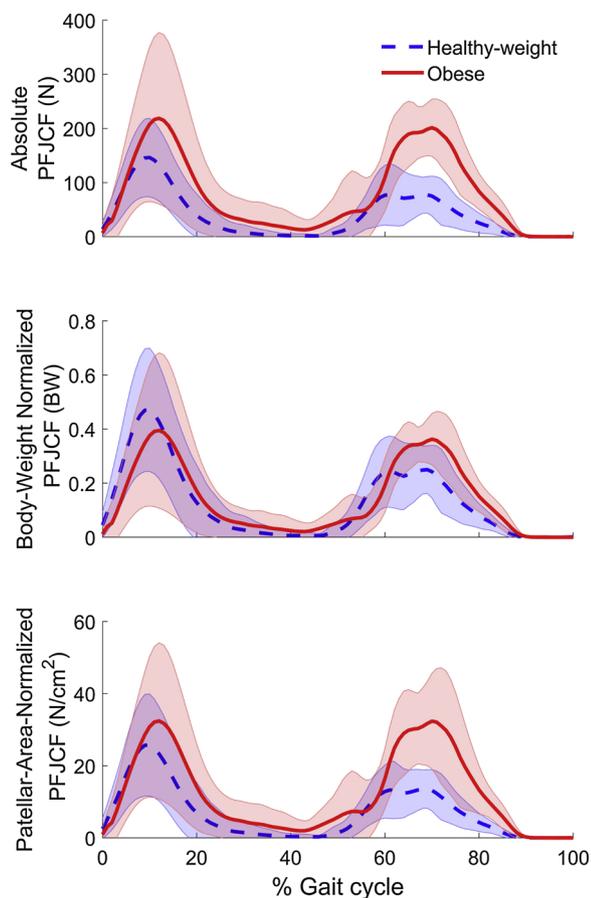Notes: *p*-values are from independent *t*-test. Values are mean (SD).

Fig. 2. Absolute, body-weight normalized, and patellar-area-normalized patellofemoral joint contact force (PFJCF) in the obese (red lines) and healthy-weight (blue dashed lines) participants across the gait cycle. Shading depicts ± 1 standard deviation. (For interpretation of the references to colour in this figure legend, the reader is referred to the web version of this article.)

2.18 times greater, respectively, in the obese versus healthy-weight children (Table 1). No significant differences were found for first peak absolute and body-weight normalized quadriceps forces between the two groups.

Pediatric obesity affected absolute, body-weight normalized, and patellar-area-normalized patellofemoral joint contact forces during walking (Fig. 2). Second peak absolute patellofemoral joint contact forces were 2.27 times greater in the obese participants compared to the healthy-weight participants (p < 0.001, Table 2); no statistically significant difference was found for the first peak. Averaged across the gait cycle, absolute patellofemoral joint contact forces were 1.98 times greater in the obese children compared to the healthy-weight children (p = 0.003). Body-weight normalized patellofemoral joint contact forces were similar between the two groups (Table 2). Normalized to the area of the patella, second peak patellofemoral joint contact force

was 2.21 times greater in the obese compared to healthy participants (p = 0.002); first peak patellar-area-normalized patellofemoral joint contact forces were not significantly different (Table 2). Averaged across the gait cycle, patellar-area-normalized patellofemoral joint contact forces were 1.81 times greater in the obese children compared to the healthy-weight children (p = 0.008).

Body fat percentage, BMI, and total body mass were significant predictors of absolute and patellar-area-normalized patellofemoral joint contact forces across the gait cycle (Fig. 3). There were weak-to-moderate relationships between body fat percentage and absolute ($p = 0.004$, $r^2 = 0.38$) and patellar-area-normalized ($p = 0.006$, $r^2 = 0.35$) patellofemoral joint contact forces. There were moderate-to-strong relationships between body mass index (BMI) and absolute ($p = < 0.001$, $r^2 = 0.58$) and patellar-area-normalized ($p = 0.001$, $r^2 = 0.45$) patellofemoral joint contact forces. There were weak-to-moderate relationships between total body mass and absolute ($p = < 0.001$, $r^2 = 0.53$) and patellar-area-normalized ($p = 0.008$, $r^2 = 0.33$) patellofemoral joint contact forces.

Patellofemoral joint contact forces computed using the standard approach from Fok et al. (Eq. 1) were generally similar compared to the contact forces computed directly using OpenSim (Fig. 4). Gait cycle peak and average root mean square differences were 73.58 N and 12.77 N, respectively, across all participants.

4. Discussion

This study investigated how pediatric obesity affects patellofemoral joint contact forces during walking. We partially confirm our first hypothesis; obese children exhibited larger second peak and average absolute and patella-area-normalized patellofemoral joint contact forces during walking compared to healthy-weight children. However, there were no significant differences in first peak forces due to the high variability across participants. Body fat percentage and BMI were positively related to absolute and patellar-area-normalized patellofemoral joint contact forces averaged across the gait cycle, confirming our second hypothesis. Our findings indicate that pediatric obesity increased the mechanical loads applied to the patella. On the other hand, our measurements indicate that the planar area of the patella did not significantly increase with obesity. Together, these findings show that patellofemoral stress during walking is greater for obese children compared to healthy-weight children.

In this study, peak body-weight-normalized patellofemoral joint contact force was around one-half body-weight during walking across all participants, which is similar to findings from a study by [12], which reported peak patellofemoral contact force from 0.5 to less-than 1.0 body-weight during walking in patients with a total knee replacement. Another study reported peak patellofemoral joint contact forces between 1.3–1.4 times body-weight in healthy adults and adult patients with patellofemoral pain (37 ± 10 years) during walking with two different self-selected speeds (˜1.4 and ˜1.8 m/s) [26]. Our reported patellofemoral contact forces are relatively small compared to this previous study. These discrepancies may be attributed to differences in subject characteristics (e.g. age), walking speed, and modeling procedures. We used validated muscle weighting factors that have been demonstrated to accurately predict knee loads.

Absolute and patellar-area-normalized patellofemoral joint contact forces were positively associated with body fat percentage, total body mass, and BMI. Among these three parameters, BMI ($p = < 0.001$, $r^2 = 0.58$) and total body mass ($p = 0.001$, $r^2 = 0.53$) were stronger predictors of absolute patellofemoral joint contact forces than body fat percentage ($p = 0.004$, $r^2 = 0.38$). These results corroborate findings from previous studies that have found that higher BMI was correlated with incidences of developing patellofemoral pain syndrome [27–29]. Our findings, and these prior studies, provide evidence supporting the potential detriment of pediatric obesity on the growing musculoskeletal system.

Table 2
Patellofemoral joint contact force.

	First peak (N)			Second peak (N)			Gait cycle average (N)		
	Healthy-weight	Obese	p	Healthy-weight	Obese	p	Healthy-weight	Obese	p
Absolute (N)	147.5 (72.6)	228.9 (160.5)	0.161	99.45 (48.9)	222.4 (62.1)	< 0.001	38.19 (14.3)	79.05 (32.8)	0.003
Body-weight Normalized (BW)	0.48 (0.2)	0.41 (0.3)	0.598	0.32 (0.1)	0.40 (0.1)	0.117	0.12 (0.0)	0.14 (0.1)	0.474
Patellar-Area-Normalized (N/cm^2)	26.02 (14.3)	33.81 (21.8)	0.358	17.09 (6.5)	35.57 (14.9)	0.002	6.70 (3.2)	12.13 (4.8)	0.008

Notes: p-values are from independent t-test. Values are mean (SD).
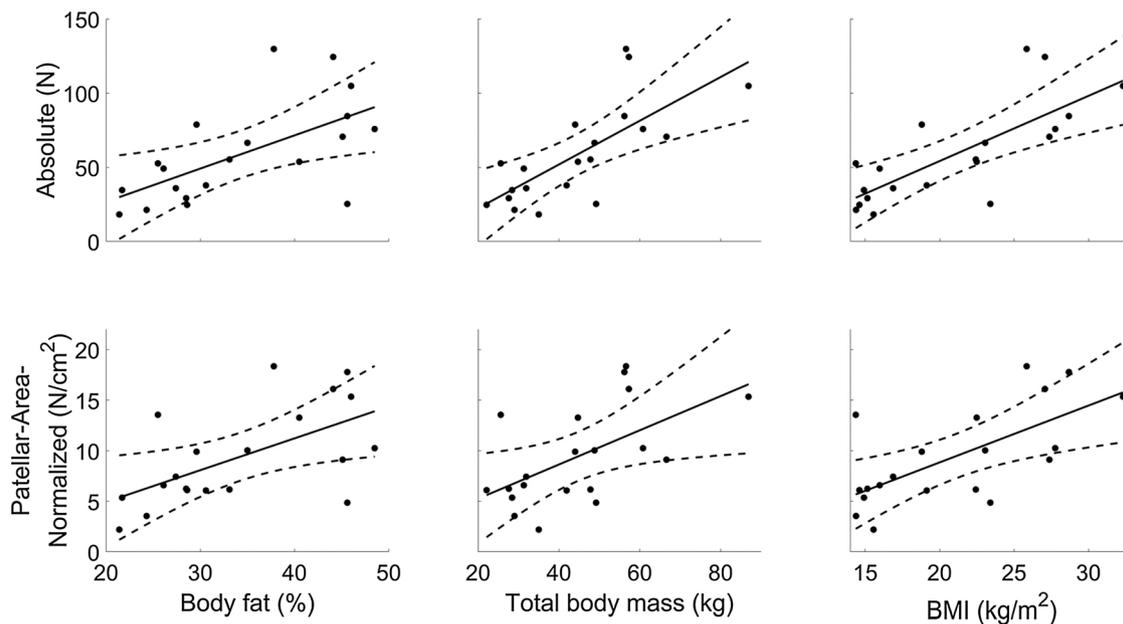

Fig. 3. Scatter grams presenting the relationships between gait cycle average absolute (top row) and patellar-area-normalized (bottom row) patellofemoral joint contact force and body fat % (left), total body mass (middle), and BMI (right). The solid lines represent the linear regression and dashed lines represent the 95% confidence intervals. Regression (r^2), intercept (b$_0$), and slope (b$_1$) coefficients describing the fit and behavior of each linear regression equation are noted on each of the plots.
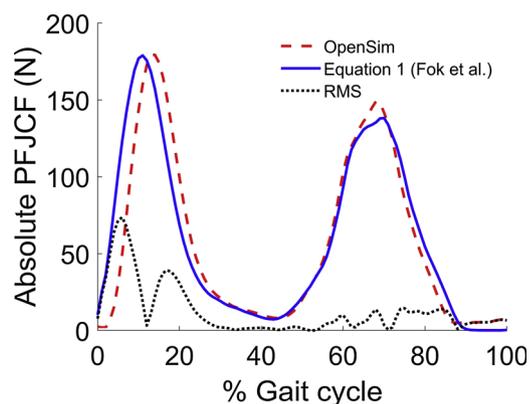

Fig. 4. Comparison of absolute patellofemoral joint contact forces (PFJCF) between Eq. 1 (blue solid line) and OpenSim (red dashed line) across all study participants. The black dotted line represents the root mean square (RMS) difference between approaches. (For interpretation of the references to colour in this figure legend, the reader is referred to the web version of this article.)

To determine the importance of modeling the loading dynamics of the patellar ligament when computing patellofemoral joint contact forces, we compared joint contact forces calculated from the frequently employed approach by Fok et al. [21] (Eq. 1) to those predicted directly using OpenSim's built-in joint reaction analysis. In the OpenSim calculation, the patellar ligament force was equal to the quadriceps force, which ignored the mechanical properties and loading response of the

patellar ligament. We found that predictions of patellofemoral contact force from the two methods were generally similar in terms of pattern and magnitude. This suggests that OpenSim's joint reaction analysis combined with the model from DeMers et al. [19] could potentially be used to predict patellar loading during walking.

There were several limitations in this study. First, the sample size was relatively small. Second, it was not feasible to directly measure patellar kinematics and the contact area between patellar and femur, and the relationship between patella perimeter and contact surface area was unknown. Future studies should use higher resolution imaging techniques to assess patellar kinematics and contact, including kinematic magnetic resonance imaging (KMRI) [30]. In theory, dynamic in vivo fluoroscopy could also be used to quantify patellofemoral kinematics [31], but administering large doses of radiation in children is a major concern. Lastly, as in prior studies, patellar ligament force-to-quadriceps ratio and patellar mechanism angle were based on cadaveric measurements [23].

This study had several strengths. We used a specifically designed kinematic marker set and methodology to improve segment tracking for individuals with considerable subcutaneous adipose tissue [32]. Radiographic images were used to develop subject-specific model segment masses and lower-extremity alignment. It is currently not possible to directly validate patellofemoral contact force because no instrumented patellofemoral joint implant exists. However, our modeling approach was indirectly validated using in-vivo gait data from an individual with an instrumented tibiofemoral joint. Because muscle forces are the primary contributor to joint contact forces [33], accurate

tibiofemoral contact forces must consequently result in accurate patellofemoral contact forces. Moreover, by employing the same approach across groups, we believe the relative effects of obesity on patellofemoral loads and, therefore, the primary conclusions of this study, are accurate.

This study may help guide physical activity prescription for obese children. Increased exposure to the elevated or abnormal patellofemoral joint loads may contribute to the development of patellofemoral pain or chondromalacia patella because magnitude, frequency, and the duration of the applied load has been associated with joint degeneration [26,34]. However, patellofemoral loads during walking are likely smaller than those incurred during dynamic movements with greater knee flexion, like running [12]. Longitudinal investigations designed to study the relationships between joint loading and disease progression in children with obesity are needed. Until those studies are completed, the cardiovascular and metabolic benefits of walking physical activity must be weighed against the potential detriment to the patellofemoral joint in obese children. Incorporating reduced weight-bearing exercises such as swimming or elliptical training may be beneficial for obese children with existing symptoms of patellofemoral injury.

In summary, we found that obese children have greater patellofemoral joint contact forces compared to healthy-weight children during walking. Elevated patellofemoral joint contact forces relative to the area of the articulating surface may have long-term negative consequences for the patellofemoral joint because elevated mechanical loads have been associated with cartilage degeneration and patellofemoral pain [26]. These findings may help clinicians develop exercise or rehabilitation programs specifically for obese children to reduce the risk of patellofemoral injury.

Declaration of Competing Interest

None declared for any authors on this manuscript.

Acknowledgements

Appendix A. Supplementary data

Supplementary material related to this article can be found, in the online version, at doi:https://doi.org/10.1016/j.gaitpost.2019.07.307.

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