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Compressive and shear hip joint contact forces are affected by pediatric obesity during walking

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Abstract

Obese children exhibit altered gait mechanics compared to healthy-weight children and have an increased prevalence of hip pain and pathology. This study sought to determine the relationships between body mass and compressive and shear hip joint contact forces during walking. Kinematic and kinetic data were collected during treadmill walking at $1 \text{ m}\cdot\text{s}^{-1}$ in 10 obese and 10 healthy-weight 8–12 year-olds. We estimated body composition, segment masses, lower-extremity alignment, and femoral neck angle via radiographic images, created personalized musculoskeletal models in OpenSim, and computed muscle forces and hip joint contact forces. Hip extension at mid-stance was 9° less, on average, in the obese children ($p < 0.001$). Hip abduction, knee flexion, and body-weight normalized peak hip moments were similar between groups. Normalized to body-weight, peak contact forces were similar at the first peak and slightly lower at the second peak between the obese and healthy-weight participants. Total body mass explained a greater proportion of contact force variance compared to lean body mass in the compressive ($r^2 = 0.89$) and vertical shear (perpendicular to the physis acting superior-to-inferior) ($r^2 = 0.84$) directions; lean body mass explained a greater proportion in the posterior shear direction ($r^2 = 0.54$). Stance-average contact forces in the compressive and vertical shear directions increased by 41 N and 48 N, respectively, for every kilogram of body mass. Age explained less than 27% of the hip loading variance. No effect of sex was found. The proportionality between hip loads and body-weight may be implicated in an obese child's increased risk of hip pain and pathology.

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CONFLICT OF INTEREST

The authors declare no conflict of interest.

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Keywords

Obesity; Gait Biomechanics; Musculoskeletal Modeling; Personalized; Joint Loading; OpenSim; Orthopedics

INTRODUCTION

Obese children are at an increased risk of developing orthopedic disorders of the lower-extremity. At the hip, pediatric obesity is positively associated with joint pain, a reduction in femoral neck anteversion angle, and slipped capitol femoral epiphysis, the most common hip disorder affecting children (Stovitz et al., 2008; Wearing et al., 2006; Wills, 2004). The increased prevalence of lower-extremity pain and pathology associated with pediatric obesity may lead to reduced physical activity and a cycle of weight gain into adulthood (Smith et al., 2014; Taylor et al., 2006).

The manner and frequency by which mechanical loads are applied to the skeleton during adolescent growth affect bone development and growth plate mechanics (Hind and Burrows, 2007; Jones et al., 2003; Villemure and Stokes, 2009). Obese children exhibit altered hip mechanics during gait compared to healthy-weight children. In a prior study, we found that body fat percentage was positively associated with hip abductor muscle forces during walking in children (Lerner et al., 2014b), while (Shultz et al., 2009) found that obese children walk with larger frontal, sagittal, and transverse plane hip moments relative to their healthy-weight peers. Researchers have theorized that elevated hip joint loading in obese children during repetitive daily physical activity (e.g. walking) contributes to their increased risk of developing pain and pathology of the hip joint (Wearing et al., 2006; Wills, 2004). However, no studies have quantified hip joint contact forces during walking in this population.

Knowledge of how pediatric obesity affects the magnitude, direction (i.e. shear vs. compression), and loading rates of hip joint contact forces during walking may lead to safer physical activity guidelines and improved surgical treatment. Researchers have found that indices of axial and bending strength of the proximal femur are more closely related to lean body mass, not total body mass in children (Petit et al., 2005). Therefore, we must understand how hip joint contact forces scale with both total and lean body masses to allow therapists to properly assess the risk-benefit of walking on the musculoskeletal systems of obese children.

The goal of this study was to determine the relationships between compressive and shear hip joint contact forces and both total body mass and lean body mass. We hypothesized that total body mass would have a strong positive correlation with compressive and shear hip joint contact forces and explain more of the hip loading variance compared to lean body mass. To accomplish this goal, we used personalized musculoskeletal models and experimentally collected biomechanics data to predict hip joint contact forces during walking in both obese and healthy-weight children.

METHODS

Study Participants

We recruited 10 obese (6 male) children with a body mass index (BMI) z-score greater than the 95th percentile and 10 healthy-weight (5 male) children with a BMI z-score between the 5th and 85th percentiles for our study (Table 1). Colorado State University's Human Research Institutional Review board granted approval for this study. Informed written assent and consent was obtained from the participants and their parents, respectively. Participants were excluded if they had any disorder that would preclude safe participation. We quantified body composition for each participant using dual x-ray absorptiometry (DXA) (Whole-Body Scan, Hologic Discover, Bedford, MA). We used a custom foot placement jig to standardize the orientation of each participant's lower-extremity in the frontal-plane. Femoral neck diameter at the physis was measured using the Hologic software.

Experimental Walking Protocol

We used reflective markers and a custom calibration procedure to capture lower-extremity, pelvis and trunk kinematics that sought to attenuate the impact of subcutaneous adiposity obscuring the motion of the skeleton. Reflective markers were placed over the 7th cervical vertebrae, acromion processes, right scapular inferior angle, sterno-clavicular notch, xyphoid process, 10th thoracic vertebrae, posterior-superior iliac spines, medial and lateral epicondyles of the femurs, medial and lateral malleoli, calcanei, first metatarsal heads, second metatarsal heads, and proximal and distal heads of the 5th metatarsals. We used a digitizing pointer (C-Motion, Germantown, MD) to probe through overlying soft-tissue and mark the anterior superior iliac spines (ASIS) and iliac crests on their bony locations. Marker clusters (four non-collinear markers affixed to a rigid plate) were adhered to the thighs, shanks, and sacrum. To account for adiposity surrounding the pelvis, post-processing (Visual 3D, C-Motion, Germantown, MA) was used to define the digital ASIS and iliac crest landmarks relative to the sacral cluster and generate virtual markers for subsequent segment tracking. For a more detailed description of this procedure see (Lerner et al., 2014a).

Each participant walked on an instrumented treadmill (Bertec Corp, Columbus, OH) at 1 m•s⁻¹ while we collected the kinematic and kinetic data. Allowing for a 5 minute acclimation period, the experimental data collected during the 6th minute of each walking trial was used in the computational analysis for this study. Marker trajectories, collected using a three-dimensional motion capture system (10 cameras, Nexus, Vicon, Centennial, CO), were recorded at 100Hz and low-pass filtered at 5 Hz using a fourth order zero-lag Butterworth filter (Winter et al., 1974). Ground reaction forces, collected from the instrumented treadmill, were recorded at 1000 Hz and low-pass filtered at 12 Hz using a fourth order zero-lag Butterworth filter.

Personalized Model Building

Reflective markers located at anatomical and digitized landmarks of each segment were used to scale a full-body musculoskeletal gait model in OpenSim to the size of each participant. Muscle moment arms, muscle attachments, and muscle length properties (muscle-tendon, tendon slack, and optimal fiber lengths) were scaled based on segment lengths. The 18

segment, 21 degree of freedom model with 92 muscle-tendon actuators (Anderson and Pandy, 1999; Delp et al., 1990) has been previously used to quantify muscle function and joint loading in children (Lerner et al., 2014b; Steele et al., 2012). The model included a knee mechanism that allowed the specification of subject-specific lower-extremity alignment (Lerner et al., 2015b). The model included a distal femoral component body and a tibial plateau body, which allowed us to specify the frontal-plane alignment of the knee while maintaining the sagittal plane rotation and translations of the tibia and patella relative to the femur. We created personalized models for each participant by refining each scaled model and specifying lower-extremity alignment and segment masses determined from the DXA radiographs. The angle formed between the intersection of the mechanical axes of the femur and tibia was used to specify personalized model alignment (Moreland et al., 1987), while the whole body DXA image was sectioned at the pelvis, thigh, and shank to obtain individual segment masses. The mass of the head, arms, and torso were included in the combined head-torso segment.

Prediction of Muscle and Joint Contact Forces

We used the method of inverse kinematics to determine joint angles during each walking trial. Inverse dynamics was used to compute joint moments. Next, we conducted a residual reduction algorithm (Delp et al., 2007) to improve the dynamic consistency of each simulated walking trail. Based on suggestions from the residual reduction algorithm, we subsequently refined the torso's mass and center of mass position for each child's model. After running the residual reduction algorithm, we found that the average residuals (<6% body-weight) were not different between groups. This provides confidence that our simulations were not affected by weight status.

Predictions of hip joint contact forces are affected by the optimization approach used to solve the neuromuscular redundancy problem (Modenese et al., 2011; Wesseling et al., 2015). Wesseling et al. found that static optimization was the most favorable approach for obtaining accurate predictions, and Bosmans et al. used this approach while predicting hip joint contact forces in children with cerebral palsy (Bosmans et al., 2014). Accordingly, we also used a static optimization method to estimate the individual muscle forces that reproduced the kinematics and kinetics of the measured walking motions; the optimization objective function sought to minimize muscle activation squared. Additionally, we introduced individual muscle weighting constants into the objection function via a custom OpenSim plugin to allow for refinement of muscle force predictions. This approach has been used previously to improve the accuracy of knee load predictions based on in-vivo measurements from instrumented joint implants (Lerner et al., 2015b; Steele et al., 2012). Hip joint contact forces were calculated using OpenSim's joint reaction analyses. The joint reaction analysis determined the resultant forces acting on the hip joint structure due to the muscle forces and external and inertial loads applied to the model.

Validation of Model and Optimization Scheme

We sought to examine the general accuracy of our hip load predictions, and evaluate the validity of using muscle weighting factors in our optimization scheme. We used our musculoskeletal model to predict the hip joint contact forces for individual with an

instrumented hip prosthesis, and subsequently compared our predictions to the *in-vivo* measurements. For this analysis, we selected the subject (IBL) from a freely-available dataset on the basis of having an average gait speed ($1.08 \text{ m}\cdot\text{s}^{-1}$) closest to that of our study ($1.0 \text{ m}\cdot\text{s}^{-1}$) (Bergmann et al., 2001) (Fig. 1). The experimental marker trajectories and ground reaction forces from the walking trial of this participant were used in our simulation workflow as described in the previous section. The coordinate systems of the predicted hip joint loads were aligned with the reported *in-vivo* hip data based on a provided schematic, and we compared our predictions to the measurements from the prosthesis. We evaluated predictions for both weighted and unweighted static optimization.

In a prior study investigating knee joint contact forces in the same cohort as in this study (Lerner et al., 2015a), we used static optimization with weighting factors of 1.5 for the gastrocnemius, 2 for the hamstrings, and 1 for all other muscles, which were established from *in-vivo* knee data (Lerner et al., 2015b). The previously established weighting factors improved the accuracy of the predicted hip forces; the difference between the measured and predicted peak hip contact force magnitude was 2% body-weight for the weighted-static optimization, and 10.5% body-weight for unweighted static optimization. Varying weighting factors between 1 and 3 for the primary muscles crossing the hip did not improve prediction accuracy. Using the previously established weighting factors, the peak axial, anteroposterior, and mediolateral contact force predictions were within 4.3%, 5.3%, and 11.3% body-weight, respectively, of the measured peak values. Therefore, we elected to use the previously established weighting factors in this study because the predictions were more accurate than unweighted static optimization, and to maintain consistency with our prior work.

Data and Statistical Analysis

To compute joint contact forces acting in compression and shear on the femoral epiphysis, we transformed the contact forces from the reference frame of the model's hip joint definition to a reference frame corresponding to the orientation of each child's femoral head and neck (Fig. 2). We calculated compressive and shear hip joint loading rates as the estimated slope of each loading response by taking the difference between the local maxima and minima of the compressive and shear contact force during weight acceptance (the first 20% of the gait cycle) and dividing by the time elapsed between those extremes.

We averaged our outcome measures across three representative gait cycles for each participant. Student's t-tests were used to test for differences between the obese and healthy-weight participants, where $p < 0.05$ defined significance. Linear regression analysis was used to determine the relationships between total and lean body mass and hip loads averaged across the stance phase. Matlab version 2015a (Mathworks Inc., Natick, MA) was used to perform data analysis, while SigmaPlot version 11.0 (Systat Software, Inc., San Jose, CA) was used to perform statistical analysis.

RESULTS

Pediatric obesity affected compressive and shear hip joint loading (Fig. 3, Table 2). Normalized to body-weight, the hip joint contact force load vectors had similar directions

and magnitudes in both the sagittal and transverse planes for the obese compared to the healthy-weight children (Fig. 4). Normalized to each physis cross-sectional area, first peak compressive and vertical shear forces were 1.77 and 1.81 times greater in the obese compared to the healthy-weight participants, respectively; first peak anteroposterior shear forces were 1.48 times greater (Fig. 4).

Total body mass and lean body mass were both significant predictors of the hip joint contact forces in each direction averaged across the stance phase (Fig 5). Total body mass had very strong positive correlations with compressive ($r=0.94$) and vertical shear ($r=0.92$) forces, and explained a greater proportion of contact force variance in these directions compared to lean body mass. Contact forces in the compressive and vertical shear directions increased by 41 N and 48 N, respectively, for every kilogram of body mass. Lean body mass explained a greater proportion of the anteroposterior shear force variance compared to total body mass, which had a moderate positive correlation ($r=0.70$) in this direction.

Normalized to body-weight, first peak forces were similar, while second peak forces were smaller, in the obese compared to the healthy-weight children (Table 2). Body-weight normalized loading rates were lower in compressive and anteroposterior directions, but similar in the vertical shear direction, in the obese compared to the healthy-weight children (Table 2). No differences were found between the sexes.

We found differences in peak sagittal, but not frontal plane hip joint kinematics between groups (Supplemental Fig. 1). Peak hip extension, which occurred during mid to late stance, was 9° less, on average, in the obese compared to healthy-weight participants ($p<0.001$). Peak bodyweight normalized hip extensor and abductor moments and muscle forces were similar between groups, except peak body-weight normalized hip flexor moments and the psoas and iliacus muscles, which were smaller in the obese children (Table 3). Absolute psoas and iliacus muscles forces were similar between groups (Supplemental Fig. 2)

DISCUSSION

In this study, we sought to determine how pediatric obesity affects hip joint loading during walking. We partially confirm our hypothesis that total body mass would have a strong positive correlation with hip joint contact forces and explain a greater proportion of the contact force variance compared to lean body mass. Total body mass had a strong correlation with compressive and vertical shear contact forces, and explained a greater proportion of the variance in these directions compared to lean body mass. However, total body mass had only a moderate correlation with anteroposterior contact force, with lean body mass explaining a greater proportion of the variance in this direction. Our finding of reduced hip extension during mid to late stance in the obese children was similar to that found in obese adults (Lerner et al., 2013). Similar to (Shultz et al., 2009) and (McMillan et al., 2010), we found no difference in frontal plane hip joint angles between groups.

The associations between total and lean body mass and joint loading were strongest in the compressive and vertical shear directions, and weakest in the anteroposterior direction. During single support, the full weight of pelvis and upper body must be supported by the

stance limb hip joint, which may explain the stronger associations between the forces acting in compression and vertical shear and total body mass. On the other hand, acceleration of the body's center of mass in the anterior and posterior directions is primarily accomplished by the ankle plantar flexors and knee extensors, respectively, which likely explains the lower associations between hip loading in this direction and total and lean body masses. There were no differences in body-weight normalized contact forces during the first peak when hip kinematics were similar between groups compared to during the second peak when the obese children exhibited reduced hip extension and lower body-weight normalized hip flexor moments. The more upright posture during this interval in the obese children likely contribute to the lower body-weight normalized iliacus and psoas muscle forces and the lower body-weight normalized hip joint contact forces.

The results of this study may provide useful information for improving our understanding of the increased prevalence of hip pain and pathology in pediatric obesity. Greater compressive and shear forces at the hip during walking in obese children may be implicated in their increased risk of developing orthopedic conditions of the hip (Wills, 2004). The directions of the applied shear forces (inferior and posterior) are consistent with the direction of femoral head slippage in the slipped capital femoral epiphysis condition (inferior-posterior). Femoral neck diameters, measured from the radiographs at the physis, were similar between the obese and healthy-weight children in this study (Table 1). This suggests that our findings of greater hip contact forces in the obese children likely represent similarly elevated compressive and shear stresses at the proximal femoral physis. Hip loading during walking in both obese and healthy-weight children may be relatively small compared to other forms of physical activity such as running. Our results provide insight into potentially greater hip joint loads in obese children during more dynamic activities.

Resultant hip loads increased more closely with total body mass than with lean body mass. If femoral bone strength does not scale with total body mass in children, as has been found previously (Petit et al., 2005), this imbalance (i.e. greater loads relative to underlying skeletal structure) may further explain the increased risk of hip pain and pathology in pediatric obesity. It might be natural to expect that bone strength and joint loading scale together, however, the loading history must be considered when relating biomechanical loading to skeletal adaptation. Reduced physical activity in obese children may be implicated in this imbalance (Trost et al., 2001). Another potential implication may be the disparity between longitudinal bone growth and microarchitecture that occurs during adolescent growth (Wills, 2004). Because this was a cross-sectional study, we were not able to evaluate causal relationships between pediatric obesity, joint loading, and hip pain and pathology. Future studies should be conducted to examine these potential relationships.

Since a relatively small sample size was a limitation of this study, we investigated how variability in the main groups affected the correlations. There were no differences in hip loads between the sexes, and only a weak association was found between hip loads and age ($r^2=0.27$). Furthermore, the relationship between hip loading and total body mass remained, even when the hip joint contact force magnitudes were normalized by height ($r^2=0.89$).

The accuracy of hip contact force predictions may be affected by several parameters. In particular, predictions are likely affected by how well each musculoskeletal model represents the subject, the quality of the experimental data used in the walking simulations, and the optimization techniques used to estimate muscle forces. We attempted to address the specific challenges presented by obesity. An inherent concern regarding the quality of the experimental data used in the simulations was that excess adiposity can cause motion artifact, obscure the motion of the underlying skeleton, and affect predictions of hip and knee joint contact forces (Lerner et al., 2014a). To address this limitation, we used a kinematic marker set and methodology developed in a prior study that was specifically designed for use in obese individuals (Lerner et al., 2014a). Scaling a generic musculoskeletal model to the size and anthropometrics of each child participant may result in non-representative segment inertial properties, skeletal alignment, and muscle properties. To address these issues, we used radiographic imaging data to build models for each subject that incorporated subject-specific segment masses and lower-extremity alignment. Several limitations remained. First, femoral neck length and angle, parameters shown to influence predictions of hip loads (Lenaerts et al., 2008), were not personalized. Additionally, muscle properties were not altered according to age or sex, and personalized moments of inertia were not modeled. Since prior research has indicated that muscle contractile properties and relative strength are not affected by pediatric obesity (Maffiuletti et al., 2008; Maffiuletti et al., 2013), and hip geometries and lower-extremity length were similar between groups, the remaining modeling limitations (i.e. generic hip geometry and scaling of muscle properties) should affect both obese and healthy-weight children equally. Different objective functions or muscle weighting factors may influence the magnitudes of the muscle forces and therefore the hip joint contact forces and the muscle weighting factors established and/or validated using elderly individuals with instrumented joints may not be applicable to children. However, prior sensitivity analyses suggest that the relative differences between groups remain when the same optimization approach is used in both groups, as was this case in this study (Steele et al., 2012). In summary, while the magnitudes and directions of the predicted loads should be interpreted with caution, the relative differences in the hip joint contact forces between the obese and healthy-weight children are likely representative.

We found that obese children have elevated hip joint loading compared to healthy-weight children during walking. At the same walking speed, obese children had greater compressive and shear contact forces and loading rates. The directions of the applied loads are consistent with the pathogenesis of the orthopedic conditions at the hip associated with pediatric obesity. This study provides evidence in support of the belief that elevated hip joint loading during walking in obese children may be implicated in their increased risk of hip pain and pathology. These results may help clinicians weigh the risk-benefit ratio of increased physical activity on the musculoskeletal system in obese children.

Supplementary Material

Refer to Web version on PubMed Central for supplementary material.

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References

- Anderson FC, Pandy MG. A Dynamic Optimization Solution for Vertical Jumping in Three Dimensions. *Computer Methods in Biomechanics and Biomedical Engineering*. 1999;201–231. [PubMed: 11264828]
- Bergmann G, Deuretzbacher G, Heller M, Graichen F, Rohlmann A, Strauss J, Duda GN. Hip contact forces and gait patterns from routine activities. *Journal of Biomechanics*. 2001; 34:859–871. [PubMed: 11410170]
- Bosmans L, Wesseling M, Desloovere K, Molenaers G, Scheys L, Jonkers I. Hip contact force in presence of aberrant bone geometry during normal and pathological gait. *Journal of Orthopaedic Research*. 2014; 32:1406–1415. [PubMed: 25087777]
- Delp SL, Anderson FC, Arnold AS, Loan P, Habib A, John CT, Guendelman E, Thelen DG. OpenSim: open-source software to create and analyze dynamic simulations of movement. *IEEE Transaction on Biomedical Engineering*. 2007; 54:1940–1950.
- Delp SL, Loan JP, Hoy MG, Zajac FE, Topp EL, Rosen JM. An interactive graphics-based model of the lower extremity to study orthopaedic surgical procedures. *IEEE Transaction on Biomedical Engineering*. 1990; 37:757–767.
- Hind K, Burrows M. Weight-bearing exercise and bone mineral accrual in children and adolescents: A review of controlled trials. *Bone*. 2007; 40:14–27. [PubMed: 16956802]
- Jones G, Bennell K, Cicuttini F. Effect of physical activity on cartilage development in healthy kids. *British Journal of Sports Medicine*. 2003; 37:382–383. [PubMed: 14514526]
- Lenaerts G, De Groote F, Demeulenaere B, Mulier M, Van der Perre G, Spaepen A, Jonkers I. Subject-specific hip geometry affects predicted hip joint contact forces during gait. *Journal of Biomechanics*. 2008; 41:1243–1252. [PubMed: 18346745]
- Lerner ZF, Board WJ, Browning RC. Effects of obesity on lower extremity muscle function during walking at two speeds. *Gait & Posture*. 2013; 39:978–984. [PubMed: 24412270]
- Lerner ZF, Board WJ, Browning RC. Effects of an Obesity-Specific Marker Set on Estimated Muscle/Joint Forces in Walking. *Medicine and Science in Sports and Exercise*. 2014a; 46:1261–1267.
- Lerner ZF, Board WJ, Browning RC. Pediatric obesity and walking duration increase medial tibiofemoral compartment contact forces. *Journal of Orthopaedic Research*. 2015a; 34:97–105.
- Lerner ZF, DeMers MS, Delp SL, Browning RC. How tibiofemoral alignment and contact locations affect predictions of medial and lateral tibiofemoral contact forces. *Journal of Biomechanics*. 2015b; 48:644–650. [PubMed: 25595425]
- Lerner ZF, Shultz SP, Board WJ, Kung S, Browning RC. Does adiposity affect muscle function during walking in children? *Journal of Biomechanics*. 2014b; 47:2975–2982. [PubMed: 25064426]
- Maffiuletti N, Jubeau M, Agosti F, Col A, Sartorio A. Quadriceps muscle function characteristics in severely obese and nonobese adolescents. *European Journal of Applied Physiology*. 2008; 103:481–484. [PubMed: 18389275]
- Maffiuletti NA, Ratel S, Sartorio A, Martin V. The Impact of Obesity on In Vivo Human Skeletal Muscle Function. *Current Obesity Reports*. 2013; 2:251–260.
- McMillan AG, Pulver AM, Collier DN, Williams DS. Sagittal and frontal plane joint mechanics throughout the stance phase of walking in adolescents who are obese. *Gait & Posture*. 2010; 32:263–268. [PubMed: 20573511]
- Modenese L, Phillips ATM, Bull AMJ. An open source lower limb model: Hip joint validation. *Journal of Biomechanics*. 2011; 44:2185–2193. [PubMed: 21742331]

- Moreland JR, Bassett LW, Hanker GJ. Radiographic analysis of the axial alignment of the lower extremity. *The Journal of Bone & Joint Surgery*. 1987; 69:745–749. [PubMed: 3597474]
- Petit MA, Beck TJ, Shults J, Zemel BS, Foster BJ, Leonard MB. Proximal femur bone geometry is appropriately adapted to lean mass in overweight children and adolescents. *Bone*. 2005; 36:568–576. [PubMed: 15777684]
- Shultz SP, Sitler MR, Tierney RT, Hillstrom HJ, Song J. Effects of pediatric obesity on joint kinematics and kinetics during 2 walking cadences. *Archives of Physical Medicine and Rehabilitation*. 2009; 90:2146–2154. [PubMed: 19969183]
- Smith S, Sumar B, Dixon K. Musculoskeletal pain in overweight and obese children. *International Journal of Obesity*. 2014; 38:11–15. [PubMed: 24077005]
- Steele KM, DeMers MS, Schwartz MH, Delp SL. Compressive tibiofemoral force during crouch gait. *Gait & Posture*. 2012; 35:556–560. [PubMed: 22206783]
- Stovitz SD, Pardee PE, Vazquez G, Duval S, Schwimmer JB. Musculoskeletal pain in obese children and adolescents. *Acta Pædiatrica*. 2008; 97:489–493.
- Taylor ED, Theim KR, Mirch MC, Ghorbani S, Tanofsky-Kraff M, Adler-Wailes DC, Brady S, Reynolds JC, Calis KA, Yanovski JA. Orthopedic complications of overweight in children and adolescents. *Pediatrics*. 2006; 117:2167–2174. [PubMed: 16740861]
- Trost SG, Kerr LM, Ward DS, Pate RR. Physical activity and determinants of physical activity in obese and non-obese children. *International Journal of Obesity*. 2001; 25:822–829. [PubMed: 11439296]
- Villemure I, Stokes IA. Growth plate mechanics and mechanobiology. A survey of present understanding. *Journal of Biomechanics*. 2009; 42:1793–1803. [PubMed: 19540500]
- Wearing SC, Hennig EM, Byrne NM, Steele JR, Hills AP. The impact of childhood obesity on musculoskeletal form. *Obesity Reviews*. 2006; 7:209–218. [PubMed: 16629876]
- Wesseling M, Derikx LC, de Groote F, Bartels W, Meyer C, Verdonschot N, Jonkers I. Muscle optimization techniques impact the magnitude of calculated hip joint contact forces. *Journal of Orthopaedic Research*. 2015; 33:430–438. [PubMed: 25492510]
- Wills M. Orthopedic complications of childhood obesity. *Pediatric physical therapy*. 2004; 16:230–235. [PubMed: 17057553]
- Winter DA, Sidwall HG, Hobson DA. Measurement and reduction of noise in kinematics of locomotion. *Journal of Biomechanics*. 1974; 7:157–159. [PubMed: 4837552]

Figure 1.

A) Axial (top), medial-lateral (middle), and anteroposterior (bottom) hip joint contact forces measured *in-vivo* (solid black line) and predicted from our musculoskeletal model (dashed gray line) using static optimization with the previously published weighting factors (Lerner et al., 2015b). The peak axial, anteroposterior, and mediolateral contact force predictions from weighted static optimization had 1.4%, 9.7%, and 14.3% percent error, respectively. B) First peak hip joint load vectors in the sagittal (left) and transverse (right) planes measured *in-vivo* (solid black lines) and predicted from our musculoskeletal model using weighted static optimization (dashed gray lines).

Figure 2. Measurement of femoral neck angle (θ) from a representative participant's DXA radiograph (inset) and subsequent transformation of the hip joint contact forces to resolve compressive forces parallel to the femoral neck, and shear forces acting perpendicular to the femoral neck and along the growth-plate.

Figure 3.

Absolute compressive (top row), vertical shear (middle row), and anteroposterior shear (bottom row) hip joint contact forces in the obese (solid black lines) and healthy-weight (dashed gray lines) participants.

Figure 4.

A) First peak hip joint load vectors normalized by body-weight in the sagittal (left) and transverse (right) planes for the obese (solid black lines) and healthy-weight (dashed gray lines) children. B) First peak hip joint load vectors normalized to the cross sectional area of the femoral neck at the physis in the sagittal (left) and transverse (right) planes. * denotes significant differences in magnitude.

Figure 5.

Scatter grams depicting the modeled relationships between the average compressive (top row), vertical shear (middle row), and anteroposterior shear (bottom row) hip joint contact forces and total body mass (left) and lean body mass (right). The solid lines represents the linear regression and the dashed lines represent the 95% confidence intervals. Noted on each plot are regression (r^2), intercept (b_0), and slope (b_1) coefficients describing the fit and behavior of each linear regression equation.

Table 1

Mean (standard deviation) participant characteristic values for the obese and healthy-weight children. Bold denotes a significant difference between groups. Alignment was measured in the frontal-plane.

	Obese	Healthy-Weight
Age (years)	9.5 (0.9)	9.6 (1.4)
Body Mass (kg)	57.5 (11.7)	31.7 (6.6)
BMI (Percentile)	98 (2)	34 (23)
BMI (kg/m ²)	26.0 (3.1)	16.0 (1.7)
Body Fat (%)	42.1 (5.0)	26.4 (3.0)
Leg Length (m)	0.72 (0.05)	0.68 (0.06)
Leg Alignment (°)	178.1 (2.5)	177.6 (0.6)
Femoral Neck Diameter (cm)	3.05(0.27)	3.07(0.31)
Femoral Neck Angle (°)	52.6 (2.8)	52.2 (3.0)

Table 2

Mean (standard deviation) first and second peak body-weight normalized hip joint contact forces, early stance loading rates. Significant differences ($p < 0.05$) are bolded.

Direction of Applied Load		Obese	Healthy-Weight	<i>p</i>
Compression	1 st Peak (BW)	2.92 (0.25)	3.03 (0.49)	0.560
	2 nd Peak (BW)	2.70 (0.45)	3.40 (0.49)	0.004
	Rate (BW•s ⁻¹)	18.4 (3.47)	22.6 (4.58)	0.031
Vertical Shear	1 st Peak (BW)	1.12 (0.14)	1.14 (0.21)	0.0779
	2 nd Peak (BW)	1.17 (0.23)	1.55 (0.35)	0.010
	Rate (BW•s ⁻¹)	6.43 (1.22)	7.92 (2.55)	0.113
Anteroposterior Shear	1 st Peak (BW)	0.68 (0.12)	0.83 (0.21)	0.080
	2 nd Peak (BW)	0.77 (0.23)	1.18 (0.43)	0.015
	Rate (BW•s ⁻¹)	4.28 (1.02)	6.04 (1.22)	0.004

Table 3

Mean (standard deviation) peak body-weight normalized joint moments and muscle forces during stance. Significant differences ($p < 0.05$) are bolded.

	Obese	Healthy-Weight	<i>p</i>
Hip Extensor Moment (Nm/kg)	0.73 (0.08)	0.69 (0.12)	0.451
Hip Flexor Moment (Nm/kg)	0.33 (0.04)	0.41 (0.12)	0.017
Hip Abductor Moment (Nm/kg)	0.50 (0.08)	0.45 (0.08)	0.224
Knee Extensor Moment (Nm/kg)	0.36 (0.10)	0.31 (0.17)	0.267
Semitendinosus (N/kg)	0.94 (0.21)	1.00 (0.24)	0.583
Semimembranosus (N/kg)	5.37 (1.02)	5.18 (1.46)	0.741
Iliacus (N/kg)	5.59 (1.94)	10.8 (3.14)	<0.001
Psoas (N/kg)	5.95 (2.16)	10.0 (2.55)	0.001
Gluteus Minimis (N/kg)	2.34 (0.44)	2.37 (0.49)	0.905
Gluteus Maximus (N/kg)	6.18 (1.42)	5.96 (2.28)	0.799
Gluteus Medias (N/kg)	12.6 (1.34)	13.2 (2.47)	0.566
Biceps Femoris (N/kg)	5.26 (1.38)	5.29 (1.27)	0.956
Rectus Femoris (N/kg)	3.99 (0.86)	3.67 (1.42)	0.545