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Musculoskeletal Models with Generic and Subject-Specific Geometry Estimate Different Joint Biomechanics in Dysplastic Hips

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Abstract

Optimizing the geometric complexity of musculoskeletal models is important for reliable yet feasible estimation of joint biomechanics. This study investigated the effects of subject-specific model geometry on hip joint reaction forces (JRFs) and muscle forces in patients with developmental dysplasia of the hip (DDH) and healthy controls. For nine DDH and nine control subjects, three models were created with increasingly subject-specific pelvis geometry, hip joint center locations and muscle attachments. Hip JRFs and muscle forces during a gait cycle were compared among the models. For DDH subjects, resultant JRFs from highly specific models including subject-specific pelvis geometry, joint locations and muscle attachments were not significantly different compared to models using generic geometry in early stance, but were significantly higher in late stance ($p=0.03$). Estimates from moderately specific models using CT-informed scaling of generic pelvis geometry were not significantly different from low specificity models using generic geometry scaled with skin markers. For controls, resultant JRFs in early stance from highly specific models were significantly lower than moderate and low specificity models ($p=0.02$) with no significant differences in late stance. Inter-model JRF differences were larger for DDH subjects than controls. Inter-model differences for JRF components and muscle forces were similar to resultant JRFs. Incorporating subject-specific pelvis geometry significantly

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affects JRF and muscle force estimates in both DDH and control groups, which may be especially important for reliable estimation of pathomechanics in dysplastic hips.

Keywords

musculoskeletal modeling; subject-specific; hip; developmental dysplasia of the hip; biomechanics; joint reaction force

Introduction

Developmental dysplasia of the hip (DDH) is a structural disease characterized by a shallow acetabulum, insufficient femoral coverage, and abnormal intra-articular loading (Leunig et al. 2001, Gala et al. 2016, Henak et al. 2014). Abnormal hip loads may contribute to acetabular labrum and articular cartilage damage (Cooperman 2013), which often progresses to early osteoarthritis (Jessel et al. 2009, Harris-Hayes and Royer 2011). Reliable quantification of hip loads, including joint reaction forces (JRFs) and muscle forces, may improve our understanding of tissue damage and the pathogenesis of osteoarthritis among patients with DDH.

Musculoskeletal models are valuable for quantifying biomechanical variables that are difficult to measure *in vivo*, including JRFs and muscle forces. Model accuracy depends on many factors, including kinematics and force input, joint representation, and passive and active muscle properties. One major factor that can affect model accuracy is representation of the musculoskeletal geometry, which dictates joint center and muscle attachment locations. To reliably estimate hip JRFs and muscle forces for specific populations, it is important to understand the level of model complexity needed with regard to bone and muscle geometry (Blemker et al. 2007). The majority of musculoskeletal modeling research has used generic bony geometry derived from cadavers, which are scaled to match anthropometrics of individuals (Neptune et al. 2001, Hamner et al. 2010, Lenhart et al. 2014). This generic scaling approach is straightforward and may be adequate for investigating healthy adult gait (Correa et al. 2011, Valente et al. 2014). However, for populations with structural hip disease, models with subject-specific geometry may be necessary due to the potential associations among abnormal geometry, joint biomechanics, and tissue damage (Heller et al. 2001, Lenaerts et al. 2008, Lenaerts et al. 2009). Yet, there is a direct relationship between model pre-processing time and computational demands with geometric complexity of the models (Valente et al. 2014, Viceconti et al. 2006, Blemker et al. 2007). Comparing estimates from models with varying levels of geometric detail is one method that can help researchers optimize specificity and complexity for reliable, yet feasible, estimation of joint biomechanics in a given population.

Geometric deformities relevant to DDH include lateralized hip joint centers (HJCs), abnormal pelvis bony geometry, and associated alterations in muscle paths (Wyles et al 2017, Clohisy et al 2011, Maquet 1999), making this condition a microcosm for determining the sensitivity of JRF and muscle force estimates to the level of geometric detail in a model. However, previously reported biomechanical estimates from models of DDH are limited in number, and findings have not been in full agreement (Skalshoi et al. 2015; Harris et al.

2017). A study using generic geometry to model gait in patients with DDH (Skalskoi et al. 2015) reported smaller hip JRFs versus controls, whereas a separate study using subject-specific pelvis geometry (Harris et al. 2017) found no difference in resultant hip JRFs, but larger medial JRFs in patients with DDH. The levels of pelvis geometry complexity may have contributed to the contrasting findings in these studies. Specifically, generic geometry may not adequately represent biomechanical differences in DDH compared to healthy hips because they omit the abnormal pelvis geometry and HJCs. Comparison between prior studies of DDH is further complicated by the use of different subject demographics and data collection protocols. By comparing generic and subject-specific models created for the same subject group, we can directly assess the effects of geometric specificity on estimation of hip biomechanics in patients with DDH.

The objective of this study was to compare JRFs and muscle forces among models with low, moderate and high levels of pelvis geometry specificity in patients with DDH and healthy controls. We hypothesized that hip JRFs and muscle forces would be significantly different for models with highly specific pelvis geometry, HJCs, and muscle paths compared to those of lower specificity. We also hypothesized that JRFs and muscle forces would be different, but to a lesser extent, for moderately specific models with scaling informed by imaging versus low specificity models scaled with skin markers.

Methods

Subjects

With Institutional Review Board approval and informed consent, eighteen subjects were recruited and reported previously (Harris et al. 2017). Nine subjects had symptomatic DDH ('DDH' group; 6 female, 3 male, 26 ± 7 years old, body mass index: 22.7 ± 3.1 kg/m²), with lateral center edge angles smaller than 20° (Wiberg 1939) confirmed by an orthopaedic surgeon and a musculoskeletal radiologist. Six patients had radiographic evidence of bilateral DDH, but all patients had unilateral symptoms at the time of data collection. The other 9 subjects were healthy controls ('CONT' group; 6 female, 3 male, 26 ± 4 years old, body mass index: 23.8 ± 4.5 kg/m²) who had no history of DDH, hip injury or other hip diseases as confirmed by radiographic inspection.

CT geometry and gait motion analysis

Computed tomography (CT) images of each subject's pelvis and proximal femurs were collected, segmented, and reconstructed in 3D with Amira v6.1 software (FEI, Hillsboro, OR, USA), using previously reported methods (Harris et al. 2012). Kinematic data during gait were collected for each subject in a motion capture laboratory and previously reported (Harris et al. 2017). Briefly, twenty-one retro-reflective markers were placed on the pelvis, thighs, shanks, feet and upper trunk and subjects walked barefoot at their self-selected speed along a 10m runway, while marker trajectories were recorded at 100 Hz and ground reaction forces were recorded at 1000 Hz. A residual analysis (Winter 2004) was performed on marker and force data separately to determine appropriate cutoff frequencies and a 4th-order, zero-lag Butterworth low-pass filter was then applied to the signals with 6 Hz and 20 Hz

cutoffs, respectively. Filtered data were imported to OpenSim musculoskeletal modeling software v3.3 (Delp et al. 2007).

Musculoskeletal modeling

A 23 degree-of-freedom model with pelvis, lower limbs, torso, head segments and 96 muscle-tendon actuators was used as the baseline model for this study, with modifications made to the path, maximum isometric force and tendon slack length of muscles around the hip (Delp et al. 1990, Shelburne et al. 2010). A virtual marker set corresponding to experimental markers was placed on the model. The baseline model (Figure 1-A) was personalized for each subject in OpenSim via three different methods: (1) marker-based isotropic scaling ('Generic'), (2) imaging-informed anisotropic scaling of the pelvis ('Nonuniform'), or (3) a previous method using subject-specific pelvis geometry, HJCs, and muscle paths from CT reconstructions ('CT-Geometry') (Harris et al 2017). These methods represented low, moderate, and high levels of model geometric specificity, respectively (Table 1).

'Generic' model - Marker-based isotropic scaling—Generic models were created by adjusting the size and inertial properties of the baseline model with one isotropic scale factor for each segment, derived from distance ratios between experimental and corresponding virtual markers. Specifically for the pelvis, a set of distances were measured and averaged between the left and right anterior superior iliac spine (ASIS) and posterior superior iliac spine (PSIS) markers (Figure 1-B). Hip joint center and muscle attachment locations were moved automatically as the generic geometry was scaled (Table 1). After scaling, a least squares optimization was used to fit the model to an experimentally captured static pose, and virtual marker positions were adjusted to match experimental marker placement specific to each subject.

'Nonuniform' model - Imaging-informed anisotropic scaling of the pelvis—Nonuniform models were also created by scaling the generic pelvis geometry of the baseline model, but used anisotropic scale factors informed by CT-based pelvis reconstructions (termed 'CT pelvis') instead of skin marker measurements. First, the CT pelvis of each subject was adjusted in Amira to match the spatial orientation of the OpenSim model pelvis geometry. Then, the anteroposterior (AP, horizontal distance between ASIS and PSIS), superoinferior (SI, vertical distance from top of iliac crest to lower border of ischial tuberosity), and mediolateral (ML, distance between left and right iliac crests) dimensions of each geometry were measured (Figure 1-B). Three distinct scale factors (AP, SI, and ML) for the model pelvis segment were calculated as the ratios of each CT pelvis dimension to the corresponding baseline pelvis dimension. The generic pelvis geometry was scaled using the AP, SI, and ML ratios, which again moved the HJC and muscle attachment locations automatically with the model pelvis (Table 1). The other model segments were scaled with marker-based measurements similar to the Generic method. By incorporating a CT-informed pelvis size, Nonuniform models were incrementally more subject-specific than the Generic models.

‘CT-Geometry’ model - Subject-specific pelvis geometry, HJCs, and muscle paths—CT-Geometry models built upon the Nonuniform models, by incorporating fully subject-specific pelvis geometry and HJC locations derived from CT. Specifically, the generic OpenSim pelvis was removed from the model and replaced by the exact CT pelvis geometry at the corresponding location and orientation in the model. HJCs for both sides were moved to subject-specific locations, determined in PreView software (<https://febio.org/preview/>) as the centroid of a sphere fit to the 3D-reconstructed femoral head geometry. Attachment locations for twenty-nine muscles crossing the hip were then updated using the subject-specific pelvis geometry (Figure 1-B, Table 1) and anatomical descriptions (Netter 2014). A sensitivity analysis was performed to test the robustness of model estimates against uncertainty in muscle attachment point placement (Appendix 1). The other model segments were scaled with markers, similar to the Generic and Nonuniform methods. As the most subject-specific models, CT-Geometry were considered the reference standard for inter-model analyses.

Biomechanical analysis and data processing

For the three models (Generic, Nonuniform, CT-Geometry) created for each subject, hip joint angles, moments, muscle forces and JRFs were computed in OpenSim (Figure 1-C). Analyses were performed on a full gait cycle of a representative trial for each subject. Data were analyzed for the symptomatic side of DDH subjects, and a randomly chosen side of CONT subjects. Hip joint angles were calculated by inverse kinematics, using least squares optimization to minimize virtual to experimental marker differences; internal net hip joint moments were then calculated by inverse dynamics (Winter 2004). A residual reduction algorithm was used to reduce nonphysical compensatory forces in the full-body inverse dynamics solution, which may originate from marker placement errors, data filtering artifacts and model body-joint assumptions (Delp et al 2007). Muscle forces were estimated by static optimization using the minimized total of squared muscle activation criterion (Crowninshield and Brand 1981), which has been shown to be appropriate for estimating muscle forces during gait (Anderson and Pandy 2001). Lastly, hip JRF components were computed from static optimization results (Steele et al. 2012).

Hip JRFs, muscle forces, angles, and moments from each model were extracted and processed using Matlab R2016a (MathWorks, Natick, MA, USA). Data were time-normalized to percentage of a gait cycle. JRFs and muscle forces were normalized by body weight; moments were normalized by body mass. JRF components were expressed in the pelvis frame to represent hip loads acting on the acetabulum. Individual hip muscles were grouped according to their functional roles in gait, and net muscle force magnitudes were summed algebraically for each functional group (hip flexors, extensors, abductors, adductors, internal rotators, external rotators). Variables were analyzed at the time points when resultant hip JRF peaked, first during early stance (termed ‘JRF1’) and again during mid-to-late stance (termed ‘JRF2’).

Statistical analysis

Inter-model differences for each outcome variable were statistically analyzed using SPSS Statistics v24 (IBM, Armonk, NY, USA). Resultant hip JRFs and components were the

primary outcome variables. Secondary outcome variables included HJC locations in each model, as well as hip angles, moments, individual and grouped muscle forces at the time of JRF1 and JRF2. Normality of data was tested using the Shapiro-Wilk test; variables with normally distributed data in all 3 models were compared using one-way ANOVA with repeated measures, while variables with any model violating data normality were compared using the Friedman test. Homogeneity of variance among inter-model differences (sphericity) was examined using Mauchly's test, with Greenhouse-Geisser corrections for data violating sphericity. Level of significance was set as $\alpha = 0.05$ for all statistical tests. For variables with significant differences among the 3 models, *post hoc* pairwise *t*-tests or Wilcoxon signed-rank tests were performed with Bonferroni corrections. Root mean square errors (RMSE) among the three model types were calculated for the resultant JRFs, JRF components, and muscle group forces at JRF1 and JRF2. All statistical tests were implemented separately for DDH and CONT subject groups.

Results

Hip JRFs

For DDH subjects, CT-Geometry models estimated larger resultant and superior JRFs at JRF2 versus Generic and Nonuniform models, while there were no significant inter-model differences at JRF1 (Figure 2; Table 2). In contrast, CT-Geometry models for CONT subjects estimated smaller medial JRFs at both JRF1 and JRF2, and slightly yet statistically different resultant, superior and posterior JRFs at JRF1 only (Figure 3; Table 3). Lastly, JRF estimates were not different between Nonuniform and Generic models for either group at JRF1 or JRF2. Root-mean-square error (RMSE) of hip JRFs between CT-Geometry and the other two models (Supplemental Tables 6 and 7) were generally larger for DDH compared to CONT, especially at JRF2, while much smaller between Nonuniform and Generic models.

Hip muscle forces

For DDH, CT-Geometry models estimated larger hip flexor and internal rotator forces at both JRF1 and JRF2, as well as smaller forces in the extensors and external rotators at JRF1, compared to Generic and Nonuniform models (Figure 4; Table 2). Similarly for CONT, CT-Geometry models estimated larger flexor forces at both JRF1 and JRF2 (Figure 5; Table 3). However, CT-Geometry models for CONT estimated smaller extensor and external rotator forces at both JRF1 and JRF2 compared to Generic and Nonuniform models, as well as larger internal rotator forces, larger adductor forces, and smaller abductor forces at JRF1 (Figure 5; Table 3). Inter-model differences were larger in DDH than CONT for hip flexor, internal rotator and early-stance extensor forces, but smaller for external rotator and mid-to-late stance extensor forces (Tables 2 and 3). Lastly, compared to Generic, Nonuniform models estimated marginally larger hip extensor and external rotator forces at JRF2 for CONT (Table 3), and no significant differences in hip muscle group forces for DDH. RMSE values of hip muscle group forces followed similar trends (Supplemental Tables 6 and 7). Inter-model differences for individual muscle forces are reported in Appendix 2.

HJC locations

For DDH, HJC locations in CT-Geometry models were more lateral than both Generic and Nonuniform models, more anterior than Nonuniform, but not significantly different from either Generic or Nonuniform in the SI direction (Table 4). For CONT, HJCs in CT-Geometry models were more anterior and inferior than both Generic and Nonuniform, but not significantly different in the ML direction (Table 4). HJC locations were not different between Nonuniform and Generic models for either subject group.

Hip angles and moments

Inter-model differences in hip angles and moments were found for both DDH and CONT groups at JRF1 and/or JRF2 (Appendix 3). Briefly, CT-Geometry models had larger hip adduction at JRF1 and a larger flexion moment throughout stance for both groups, but a larger abduction moment for DDH only and not for CONT (Appendix 3). Hip angles and moments were not significantly different between Nonuniform and Generic models.

Discussion

Optimizing the level of geometric detail needed for musculoskeletal models to reliably estimate biomechanics can be challenging, especially when studying hips with structural diseases such as DDH. The objective of this study was to compare hip JRFs and muscle forces among models with low, moderate and high levels of geometric specificity in patients with DDH and healthy controls. Overall, models with highly subject-specific pelvis geometry, hip joint locations, and muscle paths (CT-Geometry) estimated significantly different hip JRFs and muscle forces compared to models with moderate (Nonuniform) or low specificity (Generic) during gait. Specifically, using subject-specific pelvis geometry and HJC resulted in larger resultant and medial JRFs at mid-to-late stance for patients with DDH, but smaller medial JRFs for controls. In contrast, moderately subject-specific Nonuniform models did not differ substantially from low-specificity Generic models. The larger hip JRFs and related muscle forces due to patient-specific geometry support theories of pathomechanics in DDH (Maquet 1999, Delp and Maloney 1993), and emphasize the importance of accurately representing abnormal geometry when modeling joint biomechanics in patients with DDH.

As hypothesized, a high level of subject-specific geometric detail caused significant changes in hip JRF estimates for both subject groups, but there were distinct patterns for each group. For controls, CT-Geometry models estimated a slightly smaller resultant hip JRF at its early-stance peak and smaller medial JRFs throughout stance, which might be influenced by HJCs being more anterior and inferior than less specific models. HJC location has been reported to significantly affect hip JRF estimates (Lenaerts et al. 2009) due to altered flexor and extensor muscle moment arm lengths (MALs) and muscle forces that contribute to joint torques (Delp and Maloney 1993). Indeed, CT-Geometry models estimated reduced early-stance extensor forces and net extension moments, as the HJC moved anteriorly and away from extensor muscle lines of action. As more anterior HJCs and changed hip flexor-extensor forces were also found in CT-Geometry models for DDH subjects, we speculate

these traits might have not been closely represented by the coarse geometry of the generic models, for either healthy or deformed hips.

The primary effects of subject-specific geometry on models of DDH were larger resultant and superior hip JRFs at mid-to-late stance, which were not found for controls. Also unique to DDH, CT-Geometry models had significantly more lateral HJCs, which supports radiographic studies (Wyles et al. 2017). The more lateral HJCs reduced hip abductor muscle MALs, which increased demands on the abductors to generate larger force and provide stabilizing torque prior to push-off (Delp and Maloney 1993). This finding coincided with the DDH-specific trend of increased abductor muscle forces and significantly larger net abduction moments. Elevated hip abductor forces, together with increased flexor forces, then contributed to larger medial, superior, and resultant hip JRFs (Figure 2). If we consider models with high geometric specificity as the ‘reference standard’, the generic models might have underestimated potentially abnormal JRFs in DDH hips, because they could not adequately characterize the aberrant HJC locations or excessive abductor muscle demands. Therefore, it may be particularly essential to include structural details to reliably estimate hip biomechanics in the DDH population.

Significantly different muscle forces caused by subject-specific model geometry were not always reflected in changes to hip JRFs, possibly due to the changing function and coordination of individual muscles at various joint positions. For example in the DDH group, CT-Geometry models estimated larger anterior gluteus medius forces throughout stance (Supplemental Table 1) but larger resultant hip JRFs only in mid-to-late stance. While hip abductor forces increased in early stance, simultaneously decreased extensor forces might neutralize the overall effect on peak hip JRF. In contrast, during mid-to-late stance, increased hip flexor forces combined with larger abductor forces to raise the hip JRF peak. Moreover, effects from other individual muscles (Supplemental Tables 1 and 2) could account for specific changes in hip JRFs as well. As such, the effects on muscle force estimates should be interpreted cautiously with regard to their potential clinical relevance.

Joint kinematics can also affect JRF estimates. For instance, changes in hip extension angle may alter anterior JRFs during gait (Lewis et al. 2010). However, neither peak hip extension nor the AP component of hip JRF was substantially different among the models in the current study. The CT-Geometry models did have larger hip adduction during early stance (Supplemental Figure 1) especially for DDH, which could be related to the lateralized HJCs. These inter-model kinematic differences demonstrate the influence of subject-specific geometry, in particular HJC locations, on motion tracking by the models, which may affect accuracy of model estimation.

Nonuniform models were created to improve upon marker-based scaling, and represented a method that could be feasibly replicated with multi-view 2D radiographic images of the pelvis. However, the only notable differences between Nonuniform and Generic models were mid-to-late stance hip extensor and external rotator muscle forces, whose functions were minor at that time point. Even with improved scaling, the pelvis geometry in Nonuniform models was still derived from a single generic shape, and did not incorporate important geometric traits such as unique muscle lines of action relative to HJC locations.

Therefore, using 2D radiographs to scale model geometry may be inadequate for improving the accuracy of JRF and muscle force estimates.

Some limitations of this study should be recognized. First, muscle attachment points were adjusted on the pelvis using reconstructed bony geometry and canonical descriptions of muscle anatomy. As found in the sensitivity analysis, hip JRF estimates were mostly sensitive to attachment perturbations of six hip flexor and extensor muscles in ML. Since the attachment surface area of those six muscles are either small (e.g. rectus femoris) or primarily along the sagittal plane (e.g. iliacus), it was unlikely that uncertainty in muscle placement during CT-Geometry model creation meaningfully altered hip muscle force and JRF estimation. Magnetic resonance imaging could be used in future studies to improve estimation of subject-specific muscle paths (Wesseling et al. 2016, Bosmans et al. 2014, Handsfield et al. 2014). Second, we did not acquire subject-specific muscle properties such as physiological cross-sectional areas or fiber-tendon length. It is not known how muscle properties may differ between DDH and healthy groups, but because our DDH subjects were young, capable adults, and were similar to control subjects in height, weight and BMI (Harris et al. 2017), we assumed muscle properties to be the same as controls and to the baseline model (Delp et al. 1990). Third, full femur geometry was not available from CT. Femoral deformities in DDH have been reported in the literature (Wyles et al. 2017), but their influence on DDH biomechanics is unknown. Nonetheless, we used image-based femoral head geometry to determine subject-specific HJC locations in the CT-Geometry models. Another limitation is a relatively small sample size. However, our two groups were matched demographically, inter-model comparisons were made on the same gait trials for each subject, and the comparisons demonstrated statistically significant differences in key biomechanical variables such as JRFs. As experimental hip JRF and muscle force data were not available for our subjects, direct validation of the model estimates was not possible. While highly subject-specific models could provide more accurate estimation, results should be interpreted with caution in the absence of directly referable experimental force data. However, hip JRFs estimates and muscle activations from models similar to the CT-Geometry models have been shown to be comparable to electromyography signals and published force estimates in healthy hips (Harris et al. 2017). Lastly, assumptions inherent to the baseline model behavior (e.g. rotation-only hip joint, muscle force objective function) affect the accuracy of any simulation (Delp et al. 2007, Hicks et al. 2015). Thus, model results should not be considered an exact representation of musculoskeletal behavior and may not elucidate all biomechanical differences between healthy and pathologic populations. Because the objective of the current study was to quantify changes due to increasingly specific model geometry, the inter-model differences could still indicate relative improvement in model accuracy.

In conclusion, models with highly subject-specific, CT-based pelvis geometry, hip joint locations and muscle paths estimated significantly different hip JRFs and muscle forces compared to models of lower specificity. Moderately specific models with improved image-informed scaling, despite being simpler to implement than highly specific models, did not estimate hip biomechanics differently than generic models. Inter-model differences due to highly specific model geometry were greater for patients with DDH compared to healthy controls. Therefore, we recommend incorporating image-based subject-specific geometric

details for musculoskeletal models of dysplastic hips. Future research may focus on improving automation of image-based subject-specific musculoskeletal models, and exploring the appropriate levels of model complexity for other structural hip diseases.

Supplementary Material

Refer to Web version on PubMed Central for supplementary material.

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Appendix 1. Sensitivity of hip JRFs to muscle path in CT-Geometry model

An analysis was performed to test the sensitivity of hip JRF estimates to the uncertainty in muscle attachment point placement during creation of the CT-Geometry model. To demonstrate the boundary of sensitivity for both DDH and CONT groups, CT-Geometry models for one DDH and one CONT subject were chosen for the analysis, which had the largest inter-model differences in estimated resultant hip JRF (versus Generic and Nonuniform). Muscle attachment points on the CT pelvis for each of the 16 muscles generating > 0.1 xBW force during gait were perturbed by ± 5 mm in anterior, posterior, superior, inferior, medial, and lateral directions. The ± 5 mm perturbation followed a 10mm uncertainty range, which was representative of the errors when marking muscle origins and insertions with bone pins on cadaveric specimens (Brand et al. 1982; White et al. 1989). Static optimization analysis was again performed for each perturbed model; hip JRF estimates were then recomputed, and compared to the unperturbed CT-Geometry model.

For the two CT-Geometry models analyzed, a ± 5 mm perturbation of muscle attachment point in AP, SI or ML direction resulted in a large change of estimated JRFs only for a few muscles or muscle portions. Specifically, hip JRF resultant or component values differed by > 0.1 xBW at JRF1 or JRF2 among original and perturbed models (Supplemental Tables 1 and 2) for six major hip flexors and extensors: iliacus (in ML), rectus femoris (in AP, SI and ML), tensor fasciae latae (in ML), sartorius (in ML), anterior portion of gluteus medius (in SI), and semimembranosus (in SI and ML). JRF estimates at JRF1 or JRF2 were not sensitive to attachment perturbations on other hip muscles.

Appendix 2. Inter-model differences in individual hip muscle forces

For both DDH and CONT subjects, CT-Geometry models estimated larger forces (than Generic and/or Nonuniform) in iliacus and tensor fasciae latae, as well as smaller forces in gluteus maximus, posterior gluteus medius, and gluteus minimus, at JRF1 and/or JRF2 (Supplemental Tables 3 and 4). Additionally, CT-Geometry estimated larger forces in gluteus medius (anterior, middle, and total) and smaller force in semimembranosus for DDH only (Supplemental Table 3), as well as smaller forces in long head of biceps femoris and

piriformis for CONT (Supplemental Table 4). For muscles generating large force (> 0.5 xBW at JRF1 or JRF2), including gluteus maximus, gluteus medius, and iliacus, estimates from CT-Geometry were more different from the other models in DDH than in CONT, both by value and statistically (Supplemental Tables 3 and 4). Lastly, compared to Generic, Nonuniform only estimated marginally smaller forces in the middle portion of gluteus maximus at JRF1 for DDH (Supplemental Table 3).

Appendix 3. Inter-model differences in hip angles and moments

For DDH, CT-Geometry models had smaller hip flexion and larger adduction at JRF1, as well as larger extension at JRF2, compared to Generic and/or Nonuniform models; while for CONT, CT-Geometry only had larger hip adduction at JRF1 (Supplemental Figure 1; Supplemental Table 5). Hip angles were not different between Nonuniform and Generic models at JRF1 or JRF2 for either DDH or CONT.

For DDH, CT-Geometry models had a smaller hip extension moment and larger abduction moment at JRF1, as well as a larger flexion moment and larger abduction moment at JRF2, compared to Generic and/or Nonuniform (Supplemental Figure 2; Supplemental Table 5). For CONT, CT-Geometry had a smaller extension moment and larger internal rotation moment at JRF1, as well as a larger flexion moment at JRF2 (Supplemental Figure 2; Supplemental Table 5). Inter-model differences in flexion moments were similar for DDH and CONT, whereas CT-Geometry models had significantly different abduction moments only for DDH and not for CONT. Hip moments were not different between Nonuniform and Generic at JRF1 or JRF2 for either group.

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Figure 1.

Flowchart showing development of models with low, moderate, and high levels of geometric specificity. (A) Top: The baseline OpenSim model. Bottom: Subject-specific pelvis geometry reconstructed from CT. (B) I: Set of experimental (blue) and virtual (pink) marker-based measurements for isotropic pelvis scaling in a Generic model. II: 3D measurements on pelvis bony geometry for anisotropic scaling in a Nonuniform model. III: Subject-specific pelvis geometry, adjusted HJC and hip muscle attachment locations in a CT-Geometry model. (C) Subsequent biomechanical analysis to estimate hip JRFs and muscle forces during gait.

Figure 2.

Average resultant hip JRFs and components during gait for DDH subjects, expressed in pelvis frame (i.e. acting on acetabulum). Shaded area represents ± 1 standard deviation; highlighted vertical bands indicate time of JRF1 and JRF2.

Figure 3.

Average resultant hip JRFs and components during gait for CONT subjects, expressed in pelvis frame (i.e. acting on acetabulum). Shaded area represents ± 1 standard deviation; highlighted vertical bands indicate time of JRF1 and JRF2.

Figure 4.

Average hip muscle group forces during gait for DDH subjects. Shaded area represents ± 1 standard deviation; highlighted vertical bands indicate time of JRF1 and JRF2.

Figure 5.

Average hip muscle group forces during gait for CONT subjects. Shaded area represents ± 1 standard deviation; highlighted vertical bands indicate time of JRF1 and JRF2.

Table 1.

Differences among Generic, Nonuniform, and CT-Geometry model types.

	Generic	Nonuniform	CT-Geometry
Pelvis Geometry	Generic	Generic	Subject-specific from CT reconstruction
Pelvis Scaling	Uniform: 1 scale factor, derived from skin markers	Nonuniform: 3 scale factors, derived from CT dimensions (height, width, depth)	Step 1: Nonuniform Step 2: Substitute in subject-specific
HJC Locations and Hip Muscle Attachments	Automatically moved from baseline locations via uniform scaling	Automatically moved from baseline locations via nonuniform scaling	Subject-specific from CT reconstruction

Table 2.

Hip JRFs and muscle forces (mean $\pm$ 1 standard deviation) at JRF1 and JRF2 for DDH subjects, normalized by body weight (xBW).

	At JRF1				At JRF2			
	Generic	Nonuniform	CT-Geometry	p-value	Generic	Nonuniform	CT-Geometry	p-value
JRF: resultant	3.05 $\pm$ 0.40	3.05 $\pm$ 0.42	3.19 $\pm$ 0.55	n.s.	4.18 $\pm$ 1.24	4.15 $\pm$ 1.33	* $\dagger$ 5.47 $\pm$ 1.27	*=.033 $\dagger$ =.023
JRF: (+) AP (-)	<0.01 $\pm$ 0.41	0.01 $\pm$ 0.42	0.07 $\pm$ 0.39	n.s.	1.30 $\pm$ 1.02	1.28 $\pm$ 1.09	1.59 $\pm$ 0.86	n.s.
JRF: (+) SI (-)	2.97 $\pm$ 0.39	2.97 $\pm$ 0.40	3.13 $\pm$ 0.54	n.s.	3.87 $\pm$ 0.94	3.85 $\pm$ 0.99	* $\dagger$ 5.14 $\pm$ 1.07	*=.001 $\dagger$ =.001
JRF: (+) ML (-)	0.57 $\pm$ 0.10	0.57 $\pm$ 0.11	0.50 $\pm$ 0.15	n.s.	0.56 $\pm$ 0.36	0.56 $\pm$ 0.35	0.74 $\pm$ 0.38	n.s.
Flexors	0.15 $\pm$ 0.18	0.17 $\pm$ 0.20	* $\dagger$ 0.48 $\pm$ 0.44	*=.023 $\dagger$ =.033	2.31 $\pm$ 1.09	2.23 $\pm$ 1.16	* $\dagger$ 3.27 $\pm$ 1.11	*=.001 $\dagger$ =.001
Extensors	1.05 $\pm$ 0.31	1.00 $\pm$ 0.27	* $\dagger$ 0.64 $\pm$ 0.34	*<.001 $\dagger$ <.001	0.21 $\pm$ 0.14	0.22 $\pm$ 0.13	0.14 $\pm$ 0.14	n.s.
Abductors	1.79 $\pm$ 0.50	1.85 $\pm$ 0.48	1.95 $\pm$ 0.44	n.s.	1.40 $\pm$ 0.34	1.46 $\pm$ 0.38	1.85 $\pm$ 0.49	n.s.
Adductors	0.02 $\pm$ 0.02	0.01 $\pm$ 0.02	0.01 $\pm$ 0.02	n.s.	<0.01 $\pm$ <0.01	<0.01 $\pm$ <0.01	<0.01 $\pm$ <0.01	n.s.
Internal rotators	0.84 $\pm$ 0.32	0.87 $\pm$ 0.31	* $\dagger$ 1.18 $\pm$ 0.39	*=.001 $\dagger$ =.006	0.83 $\pm$ 0.26	0.87 $\pm$ 0.28	* $\dagger$ 1.25 $\pm$ 0.44	*=.025 $\dagger$ =.026
External rotators	0.39 $\pm$ 0.14	0.40 $\pm$ 0.15	* $\dagger$ 0.23 $\pm$ 0.08	*=.010 $\dagger$ =.010	0.22 $\pm$ 0.17	0.21 $\pm$ 0.14	0.14 $\pm$ 0.12	n.s.

Notes: Symbols indicating statistical significance:

* CT-Geometry vs. Generic;

$\dagger$ CT-geometry vs. Nonuniform. n.s.: not significant.

Table 3.

Hip JRFs and muscle forces (mean $\pm$ 1 standard deviation) at JRF1 and JRF2 for CONT subjects, normalized by body weight (xBW).

	At JRF1				At JRF2			
	Generic	Nonuniform	CT-Geometry	p-value	Generic	Nonuniform	CT-Geometry	p-value
JRF: resultant	3.44 $\pm$ 0.58	3.47 $\pm$ 0.59	* ∇ 3.30 $\pm$ 0.55	*=.024 ∇ =.009	4.32 $\pm$ 0.81	4.32 $\pm$ 0.77	4.71 $\pm$ 0.67	n.s.
JRF: (+) AP (-)	-0.28 $\pm$ 0.48	-0.28 $\pm$ 0.49	∇ -0.35 $\pm$ 0.46	∇ =.032	1.46 $\pm$ 0.68	1.44 $\pm$ 0.67	1.33 $\pm$ 0.55	n.s.
JRF: (+) SI (-)	3.34 $\pm$ 0.53	3.36 $\pm$ 0.54	∇ 3.24 $\pm$ 0.51	∇ =.023	4.01 $\pm$ 0.63	4.02 $\pm$ 0.61	4.49 $\pm$ 0.62	n.s.
JRF: (+) ML (-)	0.62 $\pm$ 0.15	0.66 $\pm$ 0.15	* ∇ 0.37 $\pm$ 0.09	*=.023 ∇ =.023	0.52 $\pm$ 0.14	0.53 $\pm$ 0.10	* ∇ 0.28 $\pm$ 0.11	*=.009 ∇ =.002
Flexors	0.06 $\pm$ 0.10	0.06 $\pm$ 0.07	* ∇ 0.16 $\pm$ 0.16	*=.033 ∇ =.033	2.37 $\pm$ 0.79	2.32 $\pm$ 0.77	* ∇ 2.82 $\pm$ 0.66	*=.019 ∇ =.035
Extensors	1.31 $\pm$ 0.48	1.33 $\pm$ 0.48	* ∇ 0.98 $\pm$ 0.43	*=.033 ∇ =.023	0.18 $\pm$ 0.09	∇ 0.21 $\pm$ 0.09	* ∇ 0.07 $\pm$ 0.06	*=.004 ∇ =.002 ∇ =.039
Abductors	2.17 $\pm$ 0.46	2.22 $\pm$ 0.46	* ∇ 2.04 $\pm$ 0.43	*=.020 ∇ =.003	1.49 $\pm$ 0.22	1.55 $\pm$ 0.20	1.34 $\pm$ 0.36	n.s.
Adductors	0.02 $\pm$ 0.02	0.01 $\pm$ 0.02	* ∇ 0.04 $\pm$ 0.03	*=.023 ∇ =.023	<0.01 $\pm$ <0.01	<0.01 $\pm$ <0.01	<0.01 $\pm$ <0.01	n.s.
Internal rotators	0.97 $\pm$ 0.26	0.98 $\pm$ 0.22	* ∇ 1.05 $\pm$ 0.23	*=.038 ∇ =.009	0.95 $\pm$ 0.18	0.96 $\pm$ 0.16	1.03 $\pm$ 0.39	n.s.
External rotators	0.50 $\pm$ 0.15	0.51 $\pm$ 0.16	* ∇ 0.23 $\pm$ 0.09	*=.001 ∇ =.001	0.16 $\pm$ 0.07	∇ 0.18 $\pm$ 0.07	* ∇ 0.07 $\pm$ 0.06	*=.008 ∇ =.002 ∇ =.033

Notes: Symbols indicating statistical significance:

* CT-Geometry vs. Generic;

∇ CT-geometry vs. Nonuniform;

∇ Nonuniform vs. Generic. n.s.: not significant.

Table 4.HJC locations (cm) relative to pelvis origin for DDH and CONT groups (mean $\pm$ 1 standard deviation).

		Generic	Nonuniform	CT-Geometry	p-value
DDH	(+) AP (–)	-6.7 ± 0.4	-7.3 ± 0.7	$\dot{\dagger} -5.9 \pm 0.9$	$\dot{\dagger} <.001$
	(+) SI (–)	-6.3 ± 0.3	-6.5 ± 0.4	-6.9 ± 0.6	n.s.
	(–) ML (+)	8.0 ± 0.4	8.2 ± 0.6	$^{*\dot{\dagger}} 9.1 \pm 0.4$	$^{*} <.001$ $\dot{\dagger} =.008$
CONT	(+) AP (–)	-7.3 ± 0.5	-7.3 ± 0.4	$^{*\dot{\dagger}} -5.9 \pm 0.6$	$^{*} =.001$ $\dot{\dagger} <.001$
	(+) SI (–)	-6.8 ± 0.5	-6.6 ± 0.5	$^{*\dot{\dagger}} -7.3 \pm 0.6$	$^{*} =.028$ $\dot{\dagger} =.001$
	(–) ML (+)	8.6 ± 0.6	8.7 ± 0.5	8.9 ± 0.5	n.s.

Notes: Symbols indicating statistical significance:

*
CT-Geometry vs. Generic; $\dot{\dagger}$
CT-geometry vs. Nonuniform. n.s.: not significant.