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INFLUENCE OF PATELLAR POSITION ON THE KNEE EXTENSOR MECHANISM IN NORMAL AND CROUCHED WALKING

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Abstract

Patella alta is common in cerebral palsy, especially those with crouch gait. Correction of patella alta has been advocated in the treatment of crouch, however the appropriate degree of correction and the implications for knee extensor function remain unclear. Therefore, the goal of this study was to assess the impact of patellar position on quadriceps and patellar tendon forces during normal and crouch gait. To this end, a lower extremity musculoskeletal model with a novel 12 degree of freedom knee joint was used to simulate normal gait in a healthy child, as well as mild (23 deg min knee flexion in stance), moderate (41 deg), and severe (67 deg) crouch gait in three children with cerebral palsy. The simulations reveal that quadriceps and patellar tendon forces increase dramatically with crouch, but are modulated by patellar position. For example with a normal patellar tendon position, peak patellar tendon forces were 0.7 times body weight in normal walking, but reached 2.2, 3.2 and 5.4 times body weight in mild, moderate and severe crouch. Moderate patella alta acted to reduce quadriceps and patellar tendon loads in crouch gait, due to an enhancement of the patellar tendon moment arms with alta in a flexed knee. In contrast, patella baja reduced the patellar tendon moment arm in a flexed knee and thus induced an increase in the patellar tendon loads needed to walk in crouch. Functionally, these results suggest that patella baja could also compromise knee extensor function for other flexed knee activities such as chair rise and stair climbing. The findings are important to consider when using surgical approaches for correcting patella alta in children who exhibit crouch gait patterns.

Introduction

Patella alta, or a superiorly displaced patella, is a common abnormality in cerebral palsy, especially in children who walk with a crouched gait pattern (Gage et al., 2009; Lotman, 1976; Topoleski et al., 2000). Patella alta has long been associated with knee extensor lag, which reflects an inability to actively extend the knee to its passive limit (Flynn and Weisel, 2010; Lotman, 1976). Alta could, in part, contribute to knee extensor lag by diminishing the patellar tendon moment arm (Lotman, 1976). However, some groups have nonintuitively measured an increase in the patellar tendon moment arm with alta (Sheehan et al., 2008; Ward et al., 2005). These discrepancies could arise in part from the strong dependence of the knee extensor moment arms on both knee flexion (Krevolin et al., 2004; Spoor and van Leeuwen, 1992) and patella position (Ward et al., 2005; Yamaguchi and Zajac, 1989). Such dependencies are very relevant in planning treatments for pathological gait, e.g. crouch, where the knee extensor mechanism functions at loads and postures remarkably distinct from normal gait.

Patellar tendon advancement (PTA) is a procedure used to help correct quadriceps insufficiency in those with crouch gait (Novacheck et al., 2009). The procedure distalizes the patellar tendon insertion, moving the patellar position inferiorly (Novacheck et al., 2009). It is often combined with a procedure to compensate for knee flexion contractures, such as the distal femoral extension osteotomy (DFEO) (Novacheck et al., 2009). When included, patellar tendon advancement has shown to improve gait outcomes more than performing DFEO in isolation (Stout et al., 2008). However, questions remain. PTA is often performed in a way that children tend to have patella baja, or an inferior patellar position, after surgery (Stout et al., 2008). It is not clear what the implications of patella baja are on locomotor function. Clinical outcomes data show that while knee extension during gait is often improved, there may be losses in other knee functional ability (in activities like stepping over an object or kicking an object) after the combined DFEO and PTA (Stout et al., 2008). Therefore, the effects of this corrective surgery and the trade-offs that may be occurring with the alteration in patellar position need to be better understood.

The purpose of this study was to investigate how patellar position (normal, alta, and baja) influences patellar tendon and quadriceps forces during walking in normal and crouch gait patterns. To do this, we used a novel musculoskeletal modeling platform to simulate tibiofemoral and patellofemoral mechanics during whole body movement. We hypothesized that patella alta would diminish the knee extensor moment arms in extended postures, but enhance the moment arms in flexed postures, and thus reduce the patellar tendon and quadriceps loads needed to walk in crouch.

Materials and Methods

Experimental Methods:

Retrospective gait analysis data were extracted from the database at Gillette Children's Specialty Healthcare. Use of these data for research has been approved by the University of Minnesota Institutional Review Board. Marker and force plate data during overground walking at a self-selected speed were chosen for 1 typically-developing healthy child (minimum knee flexion angle during stance = 9 degrees, age = 8.3 years, mass = 36.4 kg, height = 1.40 m, gait speed = 1.1 m/s), and 3 children with cerebral palsy who had received clinical gait analysis as part of standard of care (Davis et al., 1991). The children with cerebral palsy were chosen based on their display of crouch gait and their ability to complete gait testing without the use of walking aids. One child each displayed mild crouch (minimum knee flexion angle during stance = 23 degrees, age = 9.4 years, mass = 28.2 kg, height = 1.31 m, gait speed = 0.48 m/s), moderate crouch (minimum = 41 degrees, age = 11.0 years, mass = 28.7 kg, height = 1.43 m, gait speed = 1.12 m/s), and severe crouch (minimum = 67 degrees, age = 13.2 years, mass = 35.9 kg, height = 1.44 m, gait speed = 0.82 m/s).

Computational Modeling:

To be able to assess soft tissue loading about the knee throughout the gait cycle, we employed a 12 degree-of-freedom (DOF) knee model that has been previously described and validated in the literature (Lenhart et al., 2015a). Briefly, knee skeletal, cartilage, and ligament geometries were extracted from

MRI of a healthy adult female (23 years, 61 kg, 1.65 m). From the images, a three-body model of the knee was created with 6 DOF tibiofemoral and 6 DOF patellofemoral joints (Fig. 1). Fourteen ligament bundles, made up of sets of non-linear spring elements (Blankevoort and Huiskes, 1991) from the anatomical origins to insertions, connected the segments. Contact pressures during simulation were determined by penetration of the high-resolution meshes created from the cartilage geometries and an elastic foundation model (Bei and Fregly, 2004) (Elastic Modulus = 5MPa (Blankevoort and Huiskes, 1991; Caruntu and Hefzy, 2004), Poisson's ratio = 0.45 (Askew and Mow, 1978; Blankevoort and Huiskes, 1991; Caruntu and Hefzy, 2004; Shin et al., 2007), combined cartilage thickness of 6 mm (i.e. 3 mm on each surface (Segal et al., 2009). For this application, the femoral cartilage geometry was extended to include the anterior distal femoral shaft for simulation of patella alta (Fig. 1).

The knee model was incorporated into a lower extremity model from the literature (Arnold et al., 2010), which included 44 muscle-tendon units acting about the hip, knee, and ankle (Fig. 1). The pelvis was used as the base segment with 3 translational and 3 rotational DOF. The hip was modeled as a ball-in-socket joint with 3 rotational DOF, while the ankle was one DOF allowing for dorsi/plantarflexion. The model was implemented in SIMM (Delp and Loan, 2000) with multibody dynamics and equations of motion determined by Dynamics Pipeline (Musculographics Inc., Santa Rosa, CA) and SD/Fast (Parametric Technology Corp. Needham MA).

The model was first scaled to each subject using segment lengths during a static trial. A global optimization inverse kinematic routine was used to compute the pelvis, hip, knee and ankle kinematics that minimized the discrepancy between the measured and simulated marker positions (Lu and O'connor, 1999). In the inverse kinematics stage, the patellofemoral kinematics and 5 secondary tibiofemoral kinematic quantities were defined as functions of knee flexion determined by the healthy adult female model's behavior during a passive flexion-extension simulation (Lenhart et al., 2015a). Kinematic trajectories were subsequently low-pass filtered at 6 Hz (Van Hamme et al., 2015). Measured ground reaction force data were low-pass filtered at 30 Hz (Muniz and Nadal, 2009).

We performed concurrent optimization of muscle activations and secondary kinematics (COMAK) to characterize lower extremity muscle forces and knee mechanics over each gait cycle. In COMAK, the pelvis translation and orientation, hip angles, knee flexion and ankle dorsiflexion are set to measured values, and the ground reactions are applied to the feet at each frame. The COMAK routine is then used to simultaneously solve for muscle activations, patellofemoral kinematics and secondary tibiofemoral kinematics that are consistent with measured gait dynamics and minimize a cost function J

$$J = \sum_{i=1}^{m} V_i a_i^2 + W \cdot U_{contact}$$

consisting of the weighted sum of squared muscle activations and the net contact elastic energy (Lenhart et al., 2015b). In this equation, m is the number of muscles (=44), V is a muscle's volume (expressed in units of mm^3), a is a muscle's activation level (between 0 and 1), U is the contact elastic energy at the tibiofemoral and patellofemoral joints (expressed in joules), and W is the weighting factor on the contact energy (W = 500). In this formulation, muscles were defined to have force equal to their activation level (a_i) times their maximum isometric force (F_{0i}).

$$F_i = a_i F_{0i}$$

Consistency with gait dynamics was achieved by satisfying the constraints that the simulated accelerations of the primary degrees of freedom (hip flexion, adduction, and internal rotation, knee flexion, and ankle dorsiflexion) had to match experimental values ($\ddot{\mathbf{q}}_j^{\exp}$)

$$\ddot{\mathbf{q}}_j^{\exp} = \sum_{i=1}^{m} a_i F_{0i} \hat{\ddot{\mathbf{q}}}_{ji}^{\text{muscle}}(\mathbf{q}) + \ddot{\mathbf{q}}_j^{\text{other}}(\dot{\mathbf{q}}, \mathbf{q})$$

while the accelerations in the secondary degrees of freedom were required to be zero.

$$\mathbf{0} = \sum_{i=1}^{m} a_i F_{0i} \hat{\ddot{\mathbf{q}}}_{si}^{\text{muscle}}(\mathbf{q}) + \ddot{\mathbf{q}}_{s}^{\text{other}}(\dot{\mathbf{q}}, \mathbf{q})$$

In these equations, accelerations ($\ddot{\mathbf{q}}$) were generated by the muscles, in the primary ($\hat{\ddot{\mathbf{q}}}_{ji}^{\text{muscle}}$) and secondary ($\hat{\ddot{\mathbf{q}}}_{si}^{\text{muscle}}$) degrees of freedom, as well as by other forces ($\ddot{\mathbf{q}}^{\text{other}}$), which included external ground reactions, contact, ligament, gravitational, centripetal, and Coriolis forces. The $\ddot{\mathbf{q}}^{\text{other}}$ term also included accelerations due to the addition of a passive joint torque (M_{PJT}), which helped to modulate the changes in the secondary degrees of freedom from frame to frame.

$$M_{i_s} = -\dot{q}_i \eta_i$$

Here, $\dot{q}_i$ is the generalized speed of a secondary coordinate and η_i is a damping parameter set uniquely for each degree of freedom. See the Supplement for the values used in these simulations. It should be explicitly noted that tibiofemoral flexion was the only degree of freedom at the knee that was prescribed; the five other tibiofemoral and all patellofemoral degrees of freedom were allowed to evolve naturally in response to the muscle, ligament, contact, external, and inertial forces acting on the limb.

The process of scaling, inverse kinematics, and COMAK was repeated for the 4 gait cycles (normal and 3 crouch conditions) in each of 10 different patellar positions, for a total of 40 gait simulations. Patellar position was varied prior to scaling, such that the relative position and corresponding Insall-Salvati (IS) index (Insall and Salvati, 1971) would be similar across the different sized subjects. Variations from -1.3 cm (patella baja, Insall-Salvati Ratio = 0.53) to 3.4 cm (alta, Insall-Salvati Ratio = 2.11) were considered (Fig. 2). For this paper, positive displacement is indicative of patella alta, while negative displacement is indicative of baja, and all absolute displacements (cm) are scaled to the size of the typically-developing healthy child (height = 1.40 m). Each patella configuration was defined in the unscaled, nominal model by shifting the neutral position of the patella inferiorly (baja) or superiorly (alta) relative to the normal patellar location. Then, holding the knee flexion angle at zero degrees, passive forward simulations were performed with minimal quadriceps activation (1%) to pull the patella into contact with the femoral cartilage and to define reference patellar tendon and ligament lengths.

From the simulations, quadriceps (sum of vastus medius, vastus intermedius, vastus lateralis, and rectus femoris) and patellar tendon forces (sum of individual patellar tendon strands) were extracted over the stance phase of the gait cycle. The patellar tendon moment arm was calculated as the three dimensional perpendicular distance between the transepicondylar axis and the patellar tendon line of action (Krevolin et al., 2004). Peak moment arms were extracted at the instant of maximal knee flexion during stance (Fig. 3).

Results

Quadriceps forces varied nonlinearly with patellar positions and increased with crouch gait severity (Fig. 4). For the normal patellar position, walking with normal gait required up to 0.8 times body weight (BW) of quadriceps force, while walking with moderate crouch required 4.8 BW. Walking in severe crouch increased forces further, to just over 8 BW with a normal patellar position. The patellar position that minimized quadriceps loading was dependent on walking posture. For normal walking, having 1.1 cm of patella alta (IS = 1.31) reduced the quadriceps forces needed to 0.7 BW. With increasing crouch severity, progressively greater patella alta minimized quadriceps forces needed during gait. For example, 1.7 cm of alta (IS = 1.52) minimized quadriceps forces (2.0 BW) during mild gait, while 2.3 cm of alta (IS = 1.73) minimized quadriceps forces during moderate crouch gait (2.8 BW), and severe crouch gait (5.8 BW).

Patellar tendon (PT) forces also varied between patellar positions and increased with crouch gait severity (Fig. 5). The patellar positions that minimized patellar tendon forces were slightly more distal compared to the positions that minimized quadriceps forces. For example, 0.5 cm of baja (IS = 0.83) reduced the peak patella tendon force to 0.7 BW during normal walking. Increasingly proximal patella positions minimized patellar tendon forces with crouch, with 1.1cm of alta (IS = 1.31) reducing the peak patellar tendon force to 5.1 BW in severe crouch. The variations in the patellar tendon forces were largely attributable to the dependence of the patellar tendon moment arm on patellar position (Fig. 5, bottom). Specifically, with increasing crouch gait severity, more superior patellar positions led to enhanced patellar tendon moment arms. For normal gait, the largest moment arms for the patellar tendon occurred at a patella position of -0.5 cm which falls within the normal range (IS = 0.83). For mild crouch gait, the maximal patellar moment arm occurred with the patella shifted superiorly, but still within the normal range (0 cm, IS = 0.98). However, for moderate and severe crouch, the patellar tendon moment arm was greatest when the patella was in alta (1.1 cm, IS = 1.31).

In normal gait, quadriceps forces were greater than patellar tendon forces for baja patellar positions, while patellar tendon forces were greater than quadriceps forces for alta positions. Hence, patellar and quadriceps forces were most similar at normal patella positions (IS = 1.11) in normal gait. However increasing degrees of patella alta were needed to equalize quadriceps and patellar tendon forces in mild crouch (+1.1 cm, IS = 1.31), moderate crouch (+1.7 cm, IS = 1.52), and severe crouch (+2.3 cm, IS = 1.73) gait. Thus, the patella position for which quadriceps and patellar tendon forces were most balanced shifted progressively into alta with increasing crouch severity (Fig. 6).

Discussion

The purpose of this study was to investigate the biomechanical implications of patellar position on knee mechanics when walking in normal and crouch gait postures. Our work indicates that patellar position has substantial consequences for extensor function. The most salient finding was that patellar tendon forces were minimized, and balanced with quadriceps loads in crouch gait, when moderate patella alta was present. The reduction in patellar tendon force with alta was the result of an enhancement of the patellar tendon moment arm at increasing flexed postures. While crouch gait severely increased loading of the extensor mechanism, it does seem that moderate patella alta can act to reduce extensor loading when walking in crouch.

Patella alta has long been acknowledged in those with crouch gait and thought to contribute to knee extensor lag (the difference between passive and active knee extension capabilities) (Flynn and Weisel, 2010; Gage et al., 2009; Miller, 2005; Novacheck et al., 2009). Some have even theorized that alta contributes to crouch recurrence through a reduced moment arm in extended postures (Lotman, 1976). However, imaging studies have not consistently found a reduction in the patellar tendon moment arm with alta. For example using dynamic imaging, Sheehan found that CP children with patella alta tended to have slightly larger moment arms than healthy controls (Sheehan et al., 2008). However, that study had relatively small subject numbers and only was able to analyze moment arms over a limited range of knee flexion. Ward and others also studied those with patella alta, and found that they had reduced patellar tendon moment arms (Ward et al., 2005). However, these studies were in healthy subjects with only a mild to moderate amount of patella alta and only a 2 dimensional technique was used to determine moment arms during a static, supine, low-load task. Unlike these studies, our work indicates a more nuanced description of how patellar position affects moment arms during gait. We have found that both knee flexion angle and patellar position are needed to describe patellar tendon moment arms (Fig. 5), quadriceps forces (Fig. 4), and tendon forces (Fig. 5), and patellofemoral contact loads (Fig. 7)

Our quadriceps load estimates are generally consistent with Steele et al. (Steele et al., 2012a; Steele et al., 2012b) who documented the tremendous strength requirements necessary to walk in crouch. For example, approximately 5 times more quadriceps force is need to walk in severe crouch as compared to normal gait, with moderate crouch requiring 2-3x the extensor loads seen in normal walking (Steele et al., 2012b). These high quadriceps loads with crouch likely contribute to the tremendous metabolic cost of locomotion seen in CP patients (Johnston et al., 2004). Our study extends the analysis of Steele et al. by considering the influence of patella position on knee extensor loading. Substantial reductions (~2 BW) in quadriceps loading were feasible in moderate and severe crouch by placing the patella in alta (Fig. 4). Interestingly, patella baja also could slightly reduce quadriceps loading in moderate and severe crouch.

However these baja positions resulted in the quadriceps tendon wrapping around the distal femur (Fig. 7) and contributed to greatly reduced patellar tendon moment arms (Fig. 5).

Our results have implications for the treatment of quadriceps insufficiency and patella alta with PTA in those with crouch gait. First it is noted that correcting crouch should be the first priority. Quadriceps and patellar tendon forces increased dramatically with crouch severity, regardless of patellar position; this likely contributes to large patellofemoral contact forces and anterior knee pain in crouch patients (Sheehan 2012, Senaran 2007). The current PTA surgical approach is to place the patella in baja (Novacheck et al., 2009). Retrospective analysis has shown that outcomes are generally good with children exhibiting improved knee extension during gait after the procedure (Stout et al., 2008). However, our results suggest that placing children in baja with PTA may diminish the patellar tendon moment arm in flexed knee postures. Hence, deleterious outcomes could arise when correcting alta but not crouch, with more inferior positions leading to high strength requirements and more superiorly oriented patellofemoral contact loads (Fig. 7). Further, placing the patella in a baja position may hinder other activities that require strength in flexed postures, such as getting out of a chair or climbing stairs. Further experimental work should be done to assess whether children exhibiting baja report challenge or pain with flexed-knee tasks.

It is important to recognize the limitations of this modeling study. The generic knee model (Lenhart et al., 2015a) used in this study was scaled to individual subjects, but was originally based on healthy adult geometries. It is known that children with cerebral palsy tend to have other abnormalities, such as femoral anteversion and abnormal tibial slopes, which can affect moment arm estimates (Arnold et al., 2001) and should be further explored in the future. Our gait simulations did not account for spasticity, contractures, or other muscular abnormalities seen in CP. We used a standard optimization-based approach to estimate muscle forces that tends to minimize the use of co-contraction. Thus, our muscle-tendon force estimates are likely at the low-end, with contractures and spasticity likely amplifying the loads needed. The small subject numbers also limited our ability to understand how inter-individual variances in observed crouch gait patterns may influence force estimates. Finally, we simulated patella alta and baja by changing the length of the patellar tendon. This approach likely represents differences between normal controls and CP patients with alta (Lotman, 1976), and simulates patellar tendon shortening procedures that shift the patella into baja (Beals, 2001). However, our model does not directly emulate the PTA procedure in which patella position is modulated by distally moving the tendon insertion down the tibia (Novacheck et al., 2009). Modeling the PTA would slightly alter the line of action of the patellar tendon, but would not likely affect the patellofemoral interactions simulated in this study.

In conclusion, our work suggests that patellar position has substantial implications for quadriceps and patellar tendon forces needed to walk in crouch gait postures. In particular, a moderate degree of patella alta seems biomechanically favorable while walking in crouch, enhancing extensor patellar tendon moment arms and thereby reducing patellar tendon loads. In contrast, patella baja, such as after patellar tendon advancement, dramatically decreases the patellar tendon moment arm in flexed postures that could arise with residual crouch, or in chair-rise and stair climbing activities. Hence, surgeons must carefully consider biomechanical consequences when altering patella position in procedures used to treat crouch gait.

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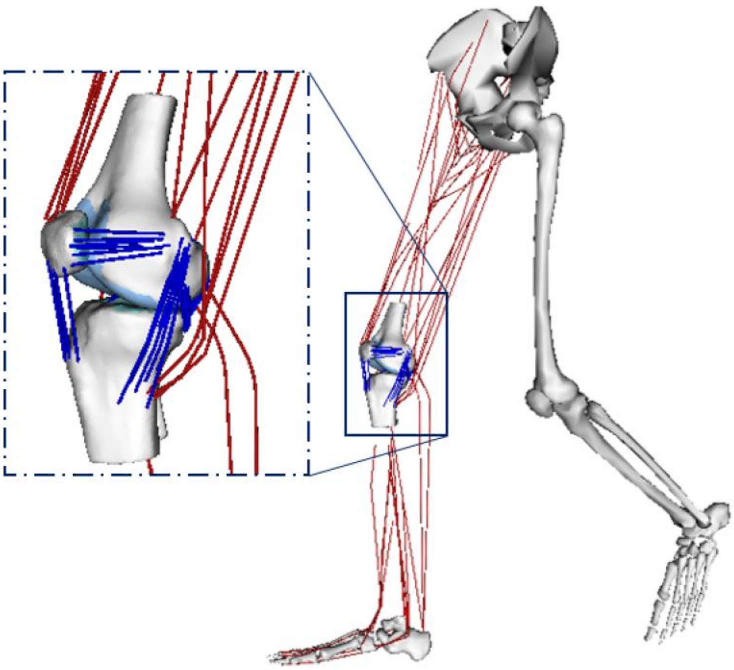
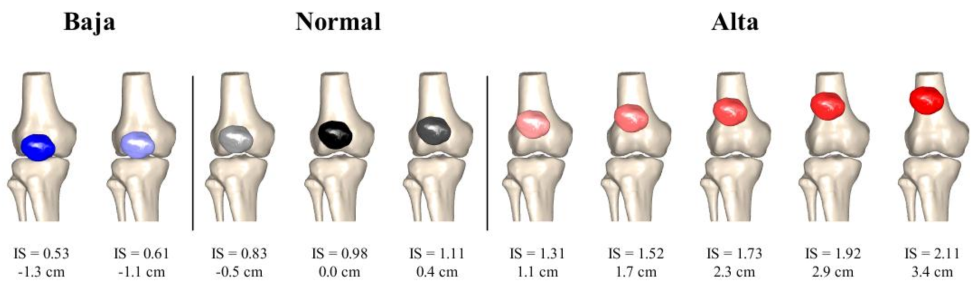
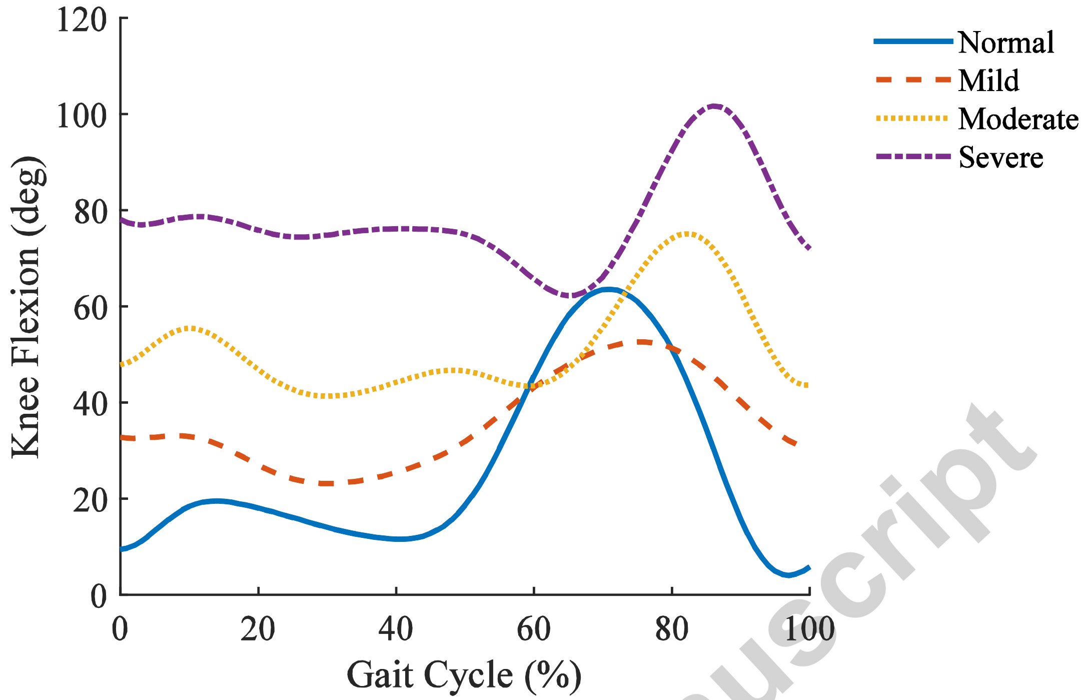
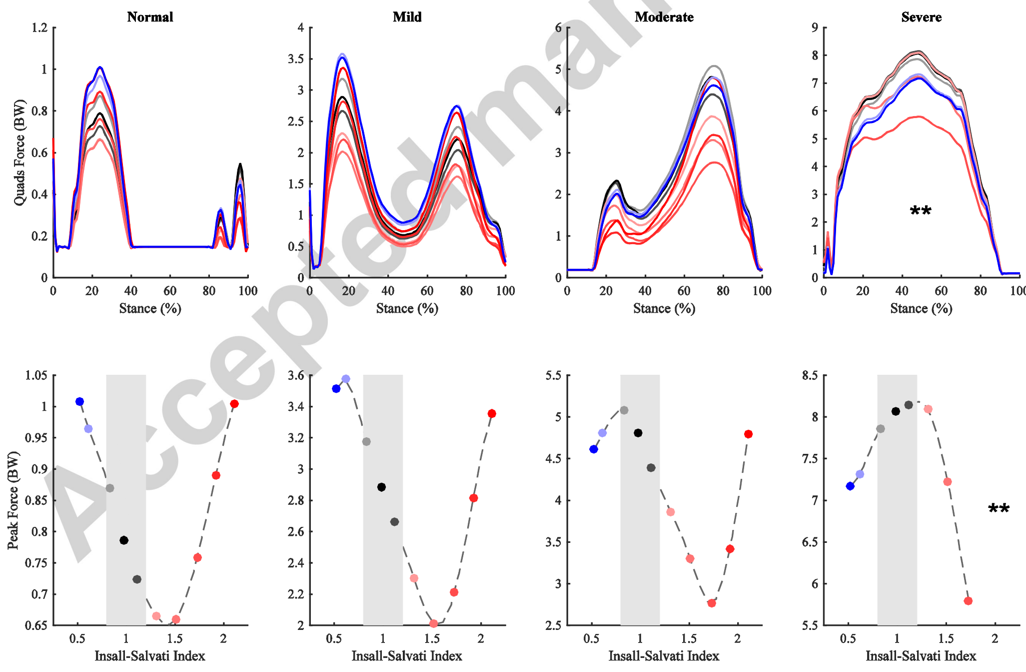
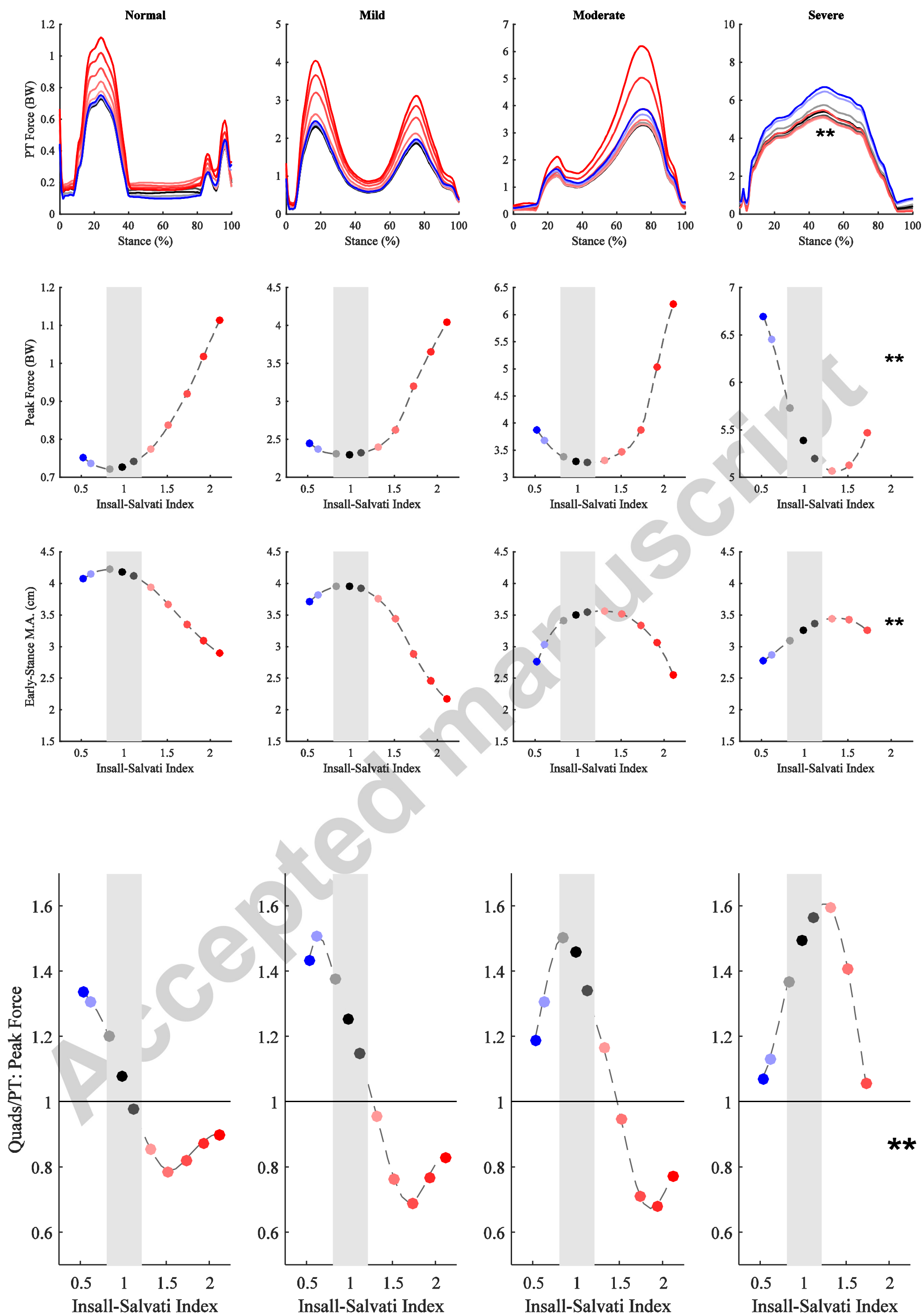


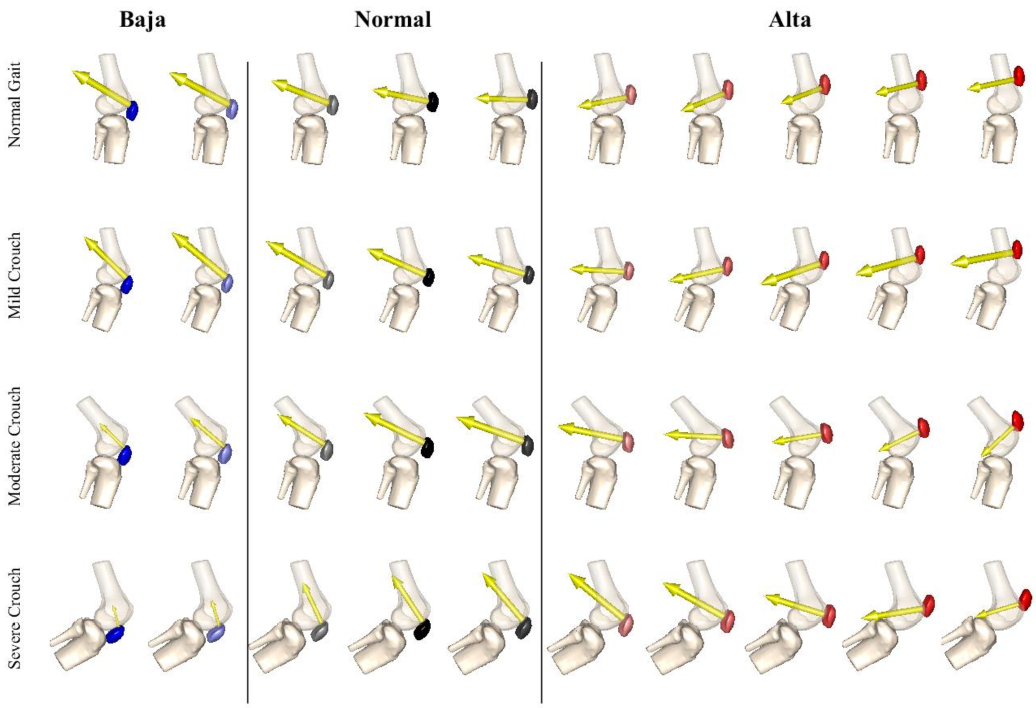

Baja
Normal
Alta
IS = 0.53
-1.3 cm
IS = 0.61
-1.1 cm
IS = 0.83
-0.5 cm
IS = 0.98
0.0 cm
IS = 1.11
0.4 cm
IS = 1.31
1.1 cm
IS = 1.52
1.7 cm
IS = 1.73
2.3 cm
IS = 1.92
2.9 cm
IS = 2.11
3.4 cm


Knee Flexion (deg)
Gait Cycle (%)
Normal
Mild
Moderate
Severe


Normal
Mild
Moderate
Severe
Quads Force (BW)
Stance (%)
**
Peak Force (BW)
Insall-Salvati Index
**



Normal
Mild
Moderate
Severe
PT Force (BW)
Stance (%)
Peak Force (BW)
Insall-Salvati Index
Early-Stance M.A. (cm)
Quads/PT: Peak Force
**



Figures

Figure Captions

Figure 1: Lower extremity musculoskeletal model with novel knee joint with 6 degrees of freedom at both the tibiofemoral and patellofemoral joints. Ligament, cartilage, and bony geometries were taken from MR images, and the behavior was validated against dynamic MRI (Lenhart et al., 2015a).

Figure 2: Demonstration of the range of patellar positions used during gait simulations. Patella positions were classified as baja, normal, or alta based on the Insall-Salvati (IS) ratio (Insall and Salvati, 1971), (Normal: $0.8 < IS < 1.2$). For ease of comparison, the inferior-superior patella position is also presented in centimeters (cm), with respect to the mean normal simulation (0.0 cm). These absolute positions are scaled to the size of the typically-developing child (height = 1.40 m) used for the normal gait simulations.

Figure 3: Knee flexion angle over the gait cycle for each of the subjects used in this analysis.

Figure 4: (Top) Quadriceps forces normalized to body weight (BW) over the stance phase of the gait cycle. Different colored lines indicate the different patellar positions simulated. Baja = Blue, Normal = Black, Alta = Red. (Bottom) Maximum quadriceps forces normalized by body weight for each patellar position. The shaded region is the normal range for patellar position. ** Simulations of severe crouch gait with the two extreme alta (IS = 1.92 and IS = 2.11) were not valid due to the lack of a wrapping constraint between the patellar tendon and distal femur. For large knee flexion angles, extreme alta caused unrealistic penetration of the patellar tendon within the distal femur.

Figure 5: (Top) Patellar tendon forces normalized to body weight (BW) over the stance phase of the gait cycle. Different colored lines indicate the different patellar positions simulated. Baja = Blue, Normal = Black, Alta = Red. (Middle) Maximum patellar tendon forces normalized by body weight for each patellar position. The shaded region is the normal range for patellar position. (Bottom) Patellar tendon moment arms (M.A) at the instant of maximal knee flexion during stance, for each patella position. Moment arms are scaled to the size of the typically-developing child (height = 1.40 m) used for the Normal Gait simulations. ** Simulations of severe crouch gait with the two extreme alta (IS = 1.92 and IS = 2.11) were not valid due to the lack of a wrapping constraint between the patellar tendon and distal femur. For large knee flexion angles, extreme alta caused unrealistic penetration of the patellar tendon within the distal femur.

Figure 6: Ratio of quadriceps to patellar tendon peak force during stance versus patellar position. Baja = Blue, Normal = Black, Alta = Red. The shaded region is the normal range for patellar position. ** Simulations of severe crouch gait with the two extreme alta (IS = 1.92 and IS = 2.11) were not valid due to the lack of a wrapping constraint between the patellar tendon and distal femur. For large knee flexion angles, extreme alta caused unrealistic penetration of the patellar tendon within the distal femur.

Figure 7: Demonstration of the interaction between patellar position and knee flexion angle on the magnitude and orientation of patellar contact force. Yellow arrow indicates the magnitude and direction of the patellar contact force acting on the femur. Force vectors are scaled for visualization; the scale factor is constant within each row, but differs between rows (crouch severity). Note that the quadriceps tendon wraps over the distal femur when baja is introduced in moderate and severe crouch, which contributes to the apparent decrease in the patello-femoral contact load in those cases.