Gait & Posture 60 (2018) 181–187

Contents lists available at ScienceDirect

Gait & Posture

journal homepage: www.elsevier.com/locate/gaitpost

Full length article

The coupled effects of crouch gait and patella alta on tibiofemoral and patellofemoral cartilage loading in children

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ARTICLE INFO

Keywords:
Cerebral palsy
Knee mechanics
Patellofemoral joint
Patella baja
Cartilage contact pressure

ABSTRACT

Background: Elevated tibiofemoral and patellofemoral loading in children who exhibit crouch gait may contribute to skeletal deformities, pain, and cessation of walking ability. Surgical procedures used to treat crouch frequently correct knee extensor insufficiency by advancing the patella. However, there is little quantitative understanding of how the magnitudes of crouch and patellofemoral correction affect cartilage loading in gait.
Methods: We used a computational musculoskeletal model to simulate the gait of twenty typically developing children and fifteen cerebral palsy patients who exhibited mild, moderate, and severe crouch. For each walking posture, we assessed the influence of patella alta and baja on tibiofemoral and patellofemoral cartilage contact.
Results: Tibiofemoral and patellofemoral contact pressures during the stance phase of normal gait averaged 2.2 and 1.0 MPa. Crouch gait increased pressure in both the tibofemoral (2.6–4.3 MPa) and patellofemoral (1.8–3.3 MPa) joints, while also shifting tibiofemoral contact to the posterior tibial plateau. For extended-knee postures, normal patellar positions (Insall-Salvatti ratio 0.8–1.2) concentrated contact on the middle third of the patellar cartilage. However, in flexed knee postures, both normal and baja patellar positions shifted pressure toward the superior edge of the patella. Moving the patella into alta restored pressure to the middle region of the patellar cartilage as crouch increased.
Conclusions: This work illustrates the potential to dramatically reduce tibiofemoral and patellofemoral cartilage loading by surgically correcting crouch gait, and highlights the interaction between patella position and knee posture in modulating the location of patellar contact during functional activities.

1. Introduction

Crouch gait is a common abnormality in cerebral palsy, with the primary feature being excessive knee flexion during the stance phase of gait. This form of walking is physically demanding, and is often associated with patellofemoral pain [1–4]. While it is generally thought that the pain is due to large mechanical demands placed on the patellofemoral joint in crouch [2], there are no prior investigations of the patellofemoral contact pressures during crouched walking. Further, patella alta (i.e. a superiorly displaced patella) is nearly universal in those with crouch [4,5], and an imaging study has suggested a relationship between the degree of patella alta and patellofemoral pain [6]. Hence, the relationship between alta, crouch, and patellofemoral pressures is important to explore.

Tibiofemoral cartilage loading is also important to consider in children with crouch gait. Prior studies have examined tibiofemoral loading in flexed postures [7,8] and crouch gait [9], with the latter study suggesting that tibiofemoral contact forces are as much as ∼2.2 times higher in severe crouch than normal walking. It is speculated that crouch may also excessively load the posterior tibia, contributing to abnormal posterior slope of the tibial plateau [10]. However, traditional gait models lack the anatomical fidelity needed to assess the influence of crouch on knee cartilage pressure magnitudes and locations.

Patellofemoral and tibiofemoral pressures are relevant to the surgical treatment of crouch gait and patella alta. Patellar tendon

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https://doi.org/10.1016/j.gaitpost.2017.12.005
Received 12 July 2017; Received in revised form 3 October 2017; Accepted 3 December 2017

advancement (PTA) is increasingly used in the treatment of crouch gait in children with cerebral palsy, and has generally been effective at reducing the degree of crouch [11]. The PTA transfers the distal insertion of the patella inferiorly on the tibia to address quadriceps insufficiency [12], which often is present in children with crouch [5]. The procedure is typically performed in a manner that places the patella in baja [11,12]. However, it is unclear how patella baja alters patellar contact pressure magnitude or location during gait. Furthermore, treatment of crouch with PTA results in reduced posterior tibal slope, especially in younger children [10]. Yet, it remains unknown whether this might be caused by alterations in tibial cartilage loading patterns.

To address these questions, we have developed a musculoskeletal model with six degrees of freedom at both the tibiofemoral and patellofemoral joints [13] to allow prediction of cartilage pressures in normal and crouch gait. This model enables patellar position to be modified and simulates the coupled influence of muscle forces, ligamentous restraints, and cartilage contact on internal knee behavior during dynamic movements such as gait. We have previously validated this model against dynamic MRI [13] and used it to predict quadriceps and patellar tendon forces during walking in crouch with varying patellar positions [14].

Hence, the objective of this study was to use our modeling framework to investigate the coupled effects of walking posture and patellar position on tibiofemoral and patellofemoral pressures. To do this, we used retrospective gait analysis data to simulate the gait of twenty healthy children with normal gait and fifteen children with varying degrees of crouch. Our goal was to provide a better understanding of the effects of patellar tendon advancement procedures on internal knee mechanics during functional activities.

2. Materials and methods

Whole body kinematics and ground reaction forces were obtained from the database at Gillette Children's Specialty Healthcare for twenty typically-developing children (mean ± SD: height 1.49 ± 0.2 m, mass 44.2 ± 16.0 kg, speed 1.1 ± 0.2 m/s) and fifteen children with cerebral palsy who exhibited mild (N = 6, height 1.41 ± 0.11 m, mass 42.5 ± 11.9 kg, speed 0.9 ± 0.2 m/s), moderate (N = 5, height 1.49 ± 0.14 m, mass 45.2 ± 13.0 kg, speed 1.0 ± 0.2 m/s), and severe (N = 4, height 1.37 ± 0.25 m, mass 35.5 ± 14.7 kg, speed 0.8 ± 0.1 m/s) crouch gait and had received clinical gait analysis as part of standard of care [15]. Crouch severity was classified by the minimum knee flexion angle during stance [16], with mean (SD) values of 5 (5), 30 (7), 45 (4), and 60 (6) degrees for normal, mild, moderate, and severe subject groups, respectively (Fig. 2A). Use of these data for research has been approved by the University of Minnesota Institutional Review Board. A 12-camera motion capture system (Vicon, Denver, CO) and four overground forceplates (AMTI, Watertown, MA) recorded marker trajectories and ground reaction forces, respectively. Kinematic trajectories and ground reaction forces were low-pass filtered at 6 Hz and 30 Hz, respectively. All children walked at their self-selected speeds and without assistive aids.

A three-body knee model, as described and validated in Lenhart et al. [13], was incorporated into a published lower extremity musculoskeletal model [17] and used to simulate knee mechanics during gait (Fig. 1A). The knee model included 6 degrees of freedom (DOF) and cartilage contact at both patellofemoral and tibiofemoral joints, and multi-strand bundles for fourteen ligaments [18,19]. For this study, the femoral cartilage geometry was extended to include the anterior femur such that, when placed in alta, the patella articulated with the distal femur (Fig. 1B).

The model was scaled for each subject using anatomical markers, then joint angles were computed at each frame of the gait cycle using inverse kinematics [20]. During inverse kinematics, the primary knee coordinate (flexion angle) was solved from motion capture data while all remaining secondary knee coordinates (non-sagittal tibiofemoral rotations, all tibiofemoral translations, and all patellofemoral DOF), which are difficult to measure using motion capture, were prescribed as functions of the flexion angle based on the passive behavior of the 12-DOF knee model [13].

We performed concurrent optimization of muscle activations and kinematics (COMAK) [21–23] to simultaneously solve for ligament forces, muscle forces, and cartilage contact pressures at every frame of the gait cycle. Briefly, COMAK estimates muscle activations and secondary knee kinematics that minimize the muscle volume-weighted sum of squared activations and the net cartilage contact elastic energy [22]. COMAK enforces dynamic constraints requiring that the simulated muscle forces, together with the internal knee contact and ligament forces exactly produced the measured hip flexion, hip adduction, hip internal rotation, knee flexion, and ankle dorsiflexion accelerations while generating zero accelerations in the secondary tibiofemoral and all patellofemoral degrees of freedom. Pelvic kinematics were prescribed to experimental values, such that whole body dynamics were implicitly resolved. COMAK solves a full gait cycle in ~20 min on a standard desktop computer.

For each subject, eight simulations were performed to investigate the effect of changing the patellar position from extreme alta (Insall-Salvati ratio [24], IS = 1.7) to baja (IS = 0.5) (Fig. 1B). For each simulation, patellofemoral and tibiofemoral contact location and magnitude information were extracted throughout the gait cycle. To assess the shift in patellar contact location, the cartilage surface was divided into inferior, middle, or superior regions, each spanning one-third of the inferior-superior height. The percentage of the total contact area falling within each region was averaged throughout stance.

3. Results

3.1. Muscle and contact forces

At the normal patellar position, quadriceps forces (sum of vasti and rectus femoris, Fig. 2B) increased over six-fold in severe crouch versus normal walking (6.4 vs. 1.0 times body weight (BW), respectively). Similarly, patellofemoral contact forces (Fig. 2D) increased dramatically with crouch, with peak loads of 0.7, 1.8, 3.2, and 5.1 BW in normal, mild, moderate and severe crouch, respectively. While peak tibiofemoral forces were similar in normal gait (2.9 BW) and mild crouch (3.2 BW), peak contact forces in moderate (4.1 BW) and severe crouch (5.7 BW) were much greater (Fig. 2F).

3.2. Contact locations

Both crouch and patellar position influenced the location of patellofemoral contact pressure. In the normal patellar position (IS = 1.0) and with normal gait, the majority of the contact area (73%, Fig. 4B) fell within the middle region of the cartilage surface. However, the contact area shifted superiorly both as crouch severity increased and as the patellar position shifted from alta to baja (Fig. 3, Fig. 4). The majority of patellar contact was restored to the middle of the cartilage surface when the patella was shifted superiorly by approximately 0.5 cm (IS = 1.1), 1.3 cm (IS = 1.3), or 2.1 cm (IS = 1.5) for mild, moderate, and severe crouch, respectively (Fig. 4B).

The tibial contact location was highly influenced by crouch severity, and sensitive to patellar position in crouched postures (Fig. 5). Compared with normal gait, the mean location of the tibiofemoral contact center of pressure (COP) during stance shifted posteriorly by 3.0, 4.8, and 8.5 mm for mild, moderate, and severe crouch, respectively (Fig. 5, Insall-Salvati = 1.0). For severe crouch, baja patellar positions reduced this posterior shift in the center of pressure by approximately 5 mm (−3.5 vs −8.5 mm, Fig. 5). Subsequent investigation revealed that the combination of baja and large knee flexion angles causes the quadriceps to wrap around the distal femur. This wrapping aligns the quadriceps and patellar tendons, and thereby mitigates the anterior patellar tendon