Knee biomechanics during squat rising with patella alta and after distal transfer of tibial tubercular surgery

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ABSTRACT

Objective: Patella alta has commonly seen in prior adolescent cerebral palsy population with crouch gait pattern. Unfortunately, for the treatments of patella alta and crouch gait, the outcomes of physical therapy, orthotic and surgical treatments are highly questionable. The purpose of this study was to investigate the effects of patella alta on knee biomechanics and to analyze the benefits of distal transfer of tibial tubercle (DTTTS) during dual limb knee squat extension. Methods: A three-dimensional dynamic knee model comprising patellofemoral and tibiofemoral joints was developed. Patellar tendon length was increased 25% and 50% of its original length to simulate the pathology. Tibiofemoral and patellofemoral contact forces, loads on ligament bundles, tibial rotations and quadriceps efficiencies were compared for pathology, normal and after DTTTS. Results: The results showed that, as patellar tendon elongation, the medial tibiofemoral contact force between 10° to 20° knee extension and the lateral tibiofemoral contact force between 17° to 30° increased gradually. It altered the neutral tension of ligament bundles, reduces the medio-lateral knee stability between 0° to 17° and increases the patello-femoral contact force and the efficiency of the quadriceps muscle during squat rising. Even though DTTTS normalized some of the biomechanical alterations, internal rotation in higher flexion angles, valgus/varus rotation of the tibia and the tension of aACL and PCL remained different than normal. Conclusion: These alterations may play a contributory role to develop cartilage degeneration on tibiofemoral and patellofemoral joints. Inadequate quadriceps force and abnormal loading pattern on some of the ligaments may also contribute to the recurrence mechanism of crouch gait after DTTTS surgery.

Keywords: Ligament, knee, squat, patella alta, patellar tendon, three-dimensional model

ÖZET


Anahtar kelimeler: Ligament, diz, çömelme, patella alta, patellar tendon, üç boyutlu model
INTRODUCTION

Patella alta and increased patellar tendon length (IPTL) are common in children with cerebral palsy who walk in crouch gait pattern. Patella alta is described in clinic as longer patellar tendon than the length of the patella.\(^1,2\)

Crouch gait pattern is a complex, multi-level disorder, characterized by exaggerated dorsiflexion of the ankle with flexion of the knee and hip during stance phase. Weak muscle groups that maintain erect posture such as hip extensors, vasti, soleus muscles and spasticity of the hamstring and iliopsoas, play primary role on developing crouch gait posture in early years in cerebral palsy.\(^3,4\) Over time, hip and knee extensors become inadequate to prevent crouch posture and consequently patella alta, excessive energy consumption, failure of patellar mechanism, excessive stress and strain on the joints, knee pain, fragmentations and stress fractures on patella and tibial tubercle develops, which make the walk impossible in adolescence period of growth.\(^1\)

Although some patients demonstrate noticeable improvement with the surgery, yet some others have little or no improvement, if not get worse.\(^1,3\) The treatment is even more difficult when crouch gait is firmly established in the midst of the adolescent growth spurt.\(^1\) The biomechanical reasons of continuing excessive hip and knee flexion after the treatment are unclear.\(^3\)

Over the year’s patella alta alters the normal biomechanical response of the ligaments and contact surfaces, creating excessive abnormal load on some areas on joint surfaces, and changing the normal kinematic behavior of the knee joint which may even worsen the existing abnormalities. Recurrence mechanism of crouch gait pattern can also be a result of biomechanical alterations.\(^1,2\)

Shortening of the quadriceps mechanism is one of the common surgical approach for treating crouch posture, which is performed surgically by transferring the tibial tubercle distally (DTTTS) or patellar tendon placation procedures.\(^5\) The aim is to match inferior pole of the patella with the joint line.\(^6\) Neither the affects of patella alta on the three dimensional knee extensor mechanics, nor the biomechanical changes after quadriceps shortening procedure has not yet been studied.\(^2\)

In this study we investigated the biomechanical consequences of elongated patellar tendon length during squat extension. To compare the alterations of the tibiofemoral contact forces, the tensions on the ligament bundles and tibial rotations for the knee with normal, IPTL and surgically treated patellar tendon length (PTL), a three dimensional tibiofemoral-patellofemoral (TPF) model was simulated.

METHODS

We studied the tibiofemoral model and tested by comparing the tibial rotation, translation and contact forces during passive knee flexion earlier (Figure 1).\(^7\) A three-dimensional patellofemoral model was simulated and integrated to the tibiofemoral model. First the kinematic behavior of the knee model was examined during the knee extension, for normal patellar tendon length (PTL). Next squat extension is simulated for knee with normal; patella alta; and after DTTTS. As a final step, the knee biomechanics with IPTL, DTTTS and normal PTL were compared.

![Figure 1. Reconstruction of tibiofemoral model: The back-oblique (a) and front-oblique (b) views of tibiofemoral model](image-url)

Reconstruction of Tibiofemoral Model

Thirteen ligament bundles were defined between femur and tibia are; anterior and posterior portions of anterior cruciate ligaments (aACL, pACL) and posterior cruciate ligament (aPCL, pPCL), anterior, deep and oblique bundles of medial collateral ligament (aMCL, dMCL, oMCL), lateral collateral ligament (LCL), medial and lateral fibers of posterior capsule (MCP, LCP), and oblique and arcuate popliteal ligaments (OPL, APL). The attachments and the physical properties of the ligaments were adopted from elsewhere. Three dimensional deformable contacts and friction force coefficients (0.04) were described between the articulating geometries.

Constructing Patellofemoral Model

Distal femoral geometric shape was constructed according to a cadaver study. Because the individual changes of the mean sulcus angle in different femur cadavers are small, it was assumed to be constant, 146° in the present study.

Patella has two axes; (3.94 cm length, 1.63 cm depth), and a plane which is bounded around the intercondylar groove (Figure 3). This plane is extended along the sulcus for resulting in a patellar length of 3.94 cm. The inertial patellar parameters were taken from elsewhere. Patellar tendon (PT) was represented as three linear springs as illustrated in Figure 4. The force on PT is:

\[
F_p = k_p (L_{tp} - L_{op})
\]

where \(F_p\) is the patellar tendon force, and \(k_p\), \(L_{tp}\), \(L_{op}\) are the stiffness (200 N/mm), the current length and the slack length of the patellar ligament respectively. The slack length \(L_{op}\) was specified to allow a ratio of 0.6 between the patellar ligament force and the quadriceps force at 90° knee flexion.
biomechanics in this range that was previously validated in elsewhere.\textsuperscript{10}

The TPF model was tested by simulating the knee extension exercise. The initial positions were defined according to the experimental data available in the literature.\textsuperscript{6} Accordingly quadriceps generate 400 N mean force during knee extension.

Once the patellar rotations, translations and tibiofemoral contact forces on the medial and lateral compartments during knee extension exercise were found satisfactory,\textsuperscript{6,9,13,15} the TPF model was utilized for the simulation of dual limb squat extension exercise as in Figure 5.\textsuperscript{10}

\begin{figure}[h]
\centering
\includegraphics[width=\textwidth]{figure5.png}
\caption{Oblique (a) and sagittal (b) views of the knee model during the simulation of squat extension}
\end{figure}

The hip joint was represented with a combination of revolute and translational joint which slides vertically during the simulation of squat rising. The ankle joint model allows free medio-lateral translation, valgus-varus, internal-external rotations and flexion-extension. However, the antero-posterior and superior-inferior movements of the ankle joint were restricted.\textsuperscript{10} Simulation of dual limb squat extension with normal PTL, IPTL and surgically treated IPTL.

Knee was described in full extension when the hip was at 0°. Both knee and the hip joints are 90° when the femur is parallel to the ground. A constant vertical load (200N) was applied to the hip during the entire simulation. The single force representing the quadriceps muscle in the model was used to extend the knee joint. During the squat extension, quadriceps acted parallel to the femoral axis and its force- knee angle relationship was defined according to the literature.\textsuperscript{10}

The biomechanical changes during squat extension were analyzed by utilizing the above described TPF model for four different conditions; i) with normal (3.94 cm) PTL, ii) elongated PTL with 25\% and iii) elongated PTL with 50\%, iv) for a simulation of DTTT surgery on the 50\% extended PTL. The surgical simulation is the displacement of the distal end of the PT along the long axes of the tibia, distally. Displacement aims the alignment of the inferior pole of the patella close to the joint line. The simulation of squat extension was repeated for all the four scenarios comparing the results with the findings of normal PTL, IPTL and DTTTS.

Since, the present study aimed to investigate the biomechanical consequences of patella alta theoretically by a computerized simulation three dimensional model, no actual patient was participated and, therefore, no approve from the ethical committee was necessary.

\textbf{RESULTS}

\textbf{Validation of the Patellofemoral Model}

Patellar rotations, patellar translations and tibiofemoral contact forces during the simulation of knee extension exercise were compared with the literature for the normal PT. In the literature patellar flexion is described as an almost linear function of the knee flexion reaching 60° at 90° knee flexion, as simulated (Figure 6). Patellar tilt and patellar rotation are reported to be in between 0-5° as in literature.\textsuperscript{6,9,15}

\begin{figure}[h]
\centering
\includegraphics[width=\textwidth]{figure6.png}
\caption{Patellar rotations; patellar flexion, extension and tilt during the simulation of knee extension}
\end{figure}

Patellar displacements were also in agreement with the literature\textsuperscript{9} as shown in Figure 7. The tibiofemoral contact force on medial compartment was found as always larger than the lateral compartment as expected (Figure 8).\textsuperscript{6} The force on the medial component is roughly twice as large of the lateral compartment also as anticipated, during the last 30° knee extension.\textsuperscript{6}

\begin{figure}[h]
\centering
\includegraphics[width=\textwidth]{figure7.png}
\caption{Posterior (dashed line), proximal and lateral (solid lines) displacement of the patella}
\end{figure}
The Outcomes of the Squat Rising

Tibiofemoral contact forces, ligament bundle loads and tibial rotations were analyzed during simulation of squat extension of the knee with normal, 25% and 50% elongated PTL, and surgically treated IPTL.

Between 10° to 23° knee flexion, the medial tibiofemoral contact force increased as a result of patellar tendon elongation. Lateral tibiofemoral contact force increased gradually by extending PTL between 17° to 30° as shown in Figure 9.

By elevating the patella, the tension on ACL had a tendency of increasing for a specific range of knee extension. Figure 10 illustrates the tension on aACL increased at higher flexion angles due to the elongation. The rise of tension on pACL, however was seen for the lower flexion angles.

For the knee with the 50% longer PTL the loads on aPCL was higher than the normal for lower knee flexion angles (between 20-25°), however, pPCL force increased at higher flexion angles and the peak force about 3 fold higher than normal (from 155 to 149 N) (Figure 11).

As the patellar tendon length elongates the peak tension of aMCL increased (Figure 12). The peak force of aMCL enlarged from 154 to 300 N between 25 to 30° knee extension. For the 50% extended PTL the load on the dMCL is lower than the normal between 10° to 23° knee extension, but gets larger at higher flexion angles (Figure 13).

As PT length was extended, the force on the LCL not only peaked at lower knee flexion angles but also started loading at higher extension angles compared to
normal PTL (Figure 14). For 50% of PTL elongation, the LCL tension was three-fold higher than normal length. The load changes due to tendon length were negligible for the rest of the ligament bundles.

Even though the internal tibial rotation was higher than the normal for the elongated PTL for the lower knee flexion angles, the tibial rotation decreased at higher flexion angles (Figure 15). Similarly, peak varus rotation of the tibia was higher in between 0° to 17° (from around 1.5 to 5.5° knee extension), although it diminished and turned into peak valgus rotation (1 to -5.5°) at higher flexion angles (Figure 16).

The patellofemoral compression force rises as the flexion angle increases. But that increase is progressively higher as PT is elongated. For 50% PTL elongation the tension rises from 2250N to 2750N. The increase is more dominant at the higher knee flexion angles (Figure 17) and the maximum quadriceps force decreased from 2035N to 2000N and prompted by elongating PTL (Figure 18).

As a result of DTTTS most of the ligament tension for the squat rising could approach to the normal. However, the loads on aACL and on the bundles of PCL remained slightly different than the normal. Although was closer to the normal compared to the extended PTL, DTTTS produced different varus/valgus and internal/external rotations (Figures 15 and 16). After DTTTS, the peak varus and valgus angle was still higher than normal (2.5 and 0.5° for normal and 3.8°, -1° after DTTTS respectively). Between 30 to 15° knee flexion, after the DTTTS, the internal rotation was about lower than normal.

DISCUSSION

Validation of the Tibio-patello-femoral Model

The results of normal knee extension exercise simulation are agreed with the literature.6,9,13,15 (Figure...
Although the existence of some discrepancies with the reference model, however do not exist when compared with the natural kinematics of the tibiofemoral and the patellofemoral joint contact forces for both compartments are match with the literature.

**Simulation of Squat Extension with Different PTLs**

The contact force on medial compartment is higher than the lateral for normal positioned patella which is in agreement with the literature.

We can conclude that as a result of IPTL the medio-lateral stability of the knee is distorted noticeably (Figure 9). The asymmetric loading between medial and the lateral compartments of the knee can be associated with the rotational instability in frontal plane (Figure 16). The elevation of varus rotation corresponded with the escalation of medial contact force, and the increase of valgus rotation is related with the rise of lateral contact force. These tibiofemoral alterations have been hypothesized to contribute the cartilage degeneration and knee osteoarthritis. When the other daily activities such as walking, squatting or even jumping are considered, it may cause a destructive effect on tibiofemoral joint cartilage for patients with patella alta.

On the other hand, the reason of coronal plane instability may due to the extra degree of freedom of ankle joint on medio-lateral direction. This characteristic of the simulation is inherited from the model. However, this behavior of the ankle provided normal squat knee extension performance for normal positioned patella. Therefore, we can assume that patella alta alters the medio-lateral behavior of the knee as well as the ankle.

According to DTTTS simulations tibiofemoral compressive forces can be corrected. Therefore, DTTTS can be decided for crouch gait patients with patella alta at an earlier stage of the growth, before articular degenerations and altered ligament tensions emerge.

Figure 18 illustrates that, as PTL increased, quadriceps can complete the same extension range with lower amount of quadriceps peak force in shorter time compared to a normal extension. This indicates that quadriceps are used more efficiently for IPTL patients as described in literature. The force-time relationship of quadriceps force returned to normal as a result of DTTTS (Figure 18). This requires a stronger quadriceps muscle contraction yielding a knee extension closer to the normal. Although children with cerebral palsy could stand more effectively with less quadriceps force, they may not be able to produce the higher quadriceps force required after DTTTS. As a result, the expected knee extension can never be achieved due to lack of quadriceps force. These inadequate quadriceps force may contribute the recurrence mechanism of crouch gait after the orthopedic surgeries. In such case the incomplete knee extension should not be attributed to a failed DTTTS but rather to weakness in quadriceps muscle.

When we analyze the results, although closer to a normal knee extension, DTTTS produced slightly different motion in general. Our findings support the theory that the surgery attempted to equalize the quadriceps moment arm and patellar tendon moment arm as it is seen in the intact knee, but the quadriceps leverage remained dominant even after the surgery. The increased anterior directed force is resisted by enlarged ACL tension at higher flexion angles as it was illustrated in Figure 10. As ACL load increases, external rotation of the tibia increases (Figure 15). By the surgery, patellofemoral compressive force is slightly under the normal. The DTTTS may decrease the patellar tendon moment arm (SM₄) as illustrated simply in Figure 19. As the tubercle transferred distally, in sagittal plane, the angle of the patellar tendon relative to the tibia decreases from (β) to (α), therefore patellar moment arm and patellofemoral contact force is reduced as stated elsewhere. For higher knee flexion angles, reduced patellofemoral contact force results in higher force transmission from quadriceps to patellar tendon.

Although the tibial rotations became closer to normal as a result of DTTTS, the internal rotation was found as lower and the excursion of the valgus-varus rotations were still slightly higher than normal for high knee flexion angles (Figure 15, 17). The new distal attachment of the patellar tendon may be the reason for the rotational instability (internal-external and valgus-varus) of the knee because it alters the normal loading
pattern of the ligaments. By the surgery, increased force transmission from quadriceps to tibia, as we mentioned earlier, may tighten the aACL which causes diminished internal rotation and contribute the frontal plane instability.

The ACL loading pattern during squatting is questionable. In Ecamiña’s work (2000), it was argued no ACL force during squatting, however several studies stated the existence of tensile force on ACL during squatting.16,19,21 The reason of the existence of ACL load in our study might be the absence of the primary muscles that produce posterior directed force during squatting such as hamstrings and gastrocnemius.21 On the other hand, most of these muscles are tight for children with cerebral palsy and can generate excessive posterior directed force on tibia. To analyze the biomechanical alterations of patella alta and DTTTS in detail, a knee model which includes other relevant muscles should be included.

The PCL force progressively increased for higher flexion angles for normal knee as reviled in literature.19,21 (Figure 11). Increased tensional load on pPCL in lower flexion angles and earlier initiation of loading pattern on aPCL could be due to induced valgus rotation which is primarily resisted by medial collateral ligaments and PCL.22 Figure 16 illustrates that as elongation of patellar tendon, the knee flexion angles where the valgus rotation increase coincided with the augmentation of tensional force on dMCL, aMCL, and PCL (Figure 11, 12, 13). To resist the medial directed force (varus rotation in frontal plane), the tension on LCL and ACL increases22 as also illustrated in Figure 10, 14 and 16.

As agreed with the literature, the patellar contact force increased as the patellar tendon elongation which may cause cartilage deformation between trochlea and the patella and generate anterior knee pain.23,24

The meniscus and the other ligaments that attach meniscus to tibial plateau and patella to knee capsule were also not included because of their complexity of the structures.7,16 Besides these limitations, the model represented the natural biomechanical behavior of the normal knee during the simulations of knee extension exercise and squat rising. Future cadaver or MRI validation studies to verify our DTTTS findings would be helpful to justify our statements.

CONCLUSION

These findings suggest that patella alta should be corrected by the surgical interventions such as distal transfer of tibial tubercle, before the irreversible tibiofemoral and patellofemoral cartilage degenerations settle which can cause the osteoarthritis for children with cerebral palsy in adult spurt of growth. Furthermore, postoperative quadriceps muscle weakness and abnormal loading patterns on ligaments after DTTTS may play a contributory role of recurrence of crouch gait pattern.

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