Analysis and Simulation of the Effect of Knee Structure on the Condylar Forces

Farshid Sadeghian [*a], **Mehrdad Davoudi** [b], **Narges Meftahi** [c,d], **Mohamad Parnianpour** [b], **Morad Karimpour** [a]

a School of Mechanical Engineering, College of Engineering, University of Tehran, Tehran, Iran; b Laboratory of Wearable Technologies and Neuromusculoskeletal Research, Department of Mechanical Engineering, Sharif University of Technology, Tehran, Iran; c Rehabilitation Sciences Research Center, Shiraz University of Medical Sciences, Shiraz, Iran; d Physical Therapy Department, School of Rehabilitation Sciences, Shiraz University of Medical Sciences, Shiraz, Iran.

***Corresponding Author:** Farshid Sadeghian, School of Mechanical Engineering, College of Engineering, University of Tehran, Tehran, Iran.
E-mail: sadeghian.f92@gmail.com

Submitted: 2020-05-29; **Accepted**: 2020-08-14; ***DOI:*** 10.22037/jcpr.v4i3.30697

Abstract

The varus deformity of the femur and the tibia leads to an increase in the medial condylar force, which may result in osteoarthritis in the medial condyle, if left untreated. Through Medial Opening Wedge Osteotomy for these patients, it is possible to realign the load-bearing surfaces of the knee joint which can lead to the tightening some of the ligaments between the femur and the tibia on the medial side, and it can generate an excessive force on the medial condyle. The correction angle of osteotomy determines how much force is removed from the medial condyle, due to the realignment of the joint, and how much force is exerted due to the tightening of the Medial Collateral Ligament (MCL). If the post-surgical force exerted on the medial condyle is greater than the force before surgery, it can cause even further damage on the medial condyle. The OpenSim has been utilized to simulate patients with varus deformity as well as calculating the condylar forces during a gait cycle. The objective of this research was to find the critical correction angle in the varus-aligned knee joint for a simulated patient in which the medial condylar force increases after surgery, due to ligament tightening. Medial condylar force of the knee and tension in the Lateral Collateral Ligament (LCL) and Popliteofibular Ligament (PFL) in a simulated subject showed that a scenario with a 76 degree Medial Proximal Tibial Angle (MPTA) corresponding to 14 degrees of deformity is a critical point beyond that the surgeon must perform an MCL release. Without releasing of MCL during High Tibial Osteotomy (HTO), an increased tension in this ligament makes a greater force in medial condyle after the surgery.

Keywords: High Tibial Osteotomy, Gait Analysis, Condylar Force, Knee, Osteoarthritis, OpenSim

Please cite this paper as: Sadeghian F, Davoudi M, Meftahi N, Parnianpour M, Karimpour M. Analysis and Simulation of the Effect of Knee Structure on the Condylar Forces. J Clin Physio Res. 2020; 5(4): e20. *DOI*: 10.22037/jcpr.v4i3.30697

Introduction

In the case of a varus knee, the medial condyle tolerates greater amounts of force compared with the lateral condyle (1-3). In addition, in the single-support phase of a gait cycle, the load is concentrated on the medial condyle in a varus-aligned knee. This additional pressure can cause an articular cartilage damage and eventually osteoarthritis (OA) (4). It is shown that rate of articular cartilage loss and subsequent arthritis, are directly related to the peak adduction moment of the knee throughout of the stance phase of a gait cycle (5, 6). Moreover, it has been demonstrated that people with a varus knee have a higher degree of flexion during walking, resulting in greater tension in the quadriceps (for resistance against knee moment flexion), and subsequent higher joint force (7). Knee OA is often unicompartmental and affects the medial compartment in greater numbers (4) which leads to greater varus deformity, resulting in a greater force exerted on the medial compartment and accelerating its degeneration (7). In order to determine forces exerted on the different structures of the knee joint, the complex finite element models have been used in the previous studies (8-4). However, these models have some limitations for example the exact modeling depends on the knee geometry inside the body. Due to the variations of force caused by unique individual geometries of patient's bodies, medical imaging techniques such as MRI and CT have been used to incorporate patient-specific geometry into studies (10, 12-19).

Yang *et al.* (20) have created a three-dimensional finite element model of the knee joint based on MRI to investigate the effect of the frontal plane tibiofemoral angle on the stress and strain distribution at the knee cartilage. Their findings showed that in the presence of the varus deformity, the compression in the medial compartment of the knee was greater compared to those with normal and valgus alignment. They suggested that subjects with varus deformity may be more susceptible to medial compartment OA. Agneskirchner *et al.* measured the impact of valgus opening wedge high tibial osteotomy (HTO) and different loading alignment on joint contact pressures in vitro (21). The procedure was carried out with and without the dissection of medial collateral ligament (MCL). When the MCL was released, the pressure distribution was 1.4 MPa and 2.2 MPa in the medial and lateral compartments, respectively. They also reported the alignment which applies the greatest load to the medial compartment was the varus knee and showed that decompressive effect of HTO only occurred by MCL release. Gausbeek *et al.* (22) concluded that HTO not only leads to tightening in the MCL, but also has some influences on patellar height. Schmitz and Piovesan developed a knee model consisted of a six degrees of freedom (DOFs) of tibiofemoral joint, and a one DOF of patellofemoral joint used in their study. This model was created in the OpenSim (Schmitz and Piovesan's model) (23) and is freely available online at OpenSim webpage (24). Schmitz and Piovesan's model included eighteen ligament bundles as well as tibiofemoral contacts. It is an open source, discrete element model of the knee where the rigid body dynamics and soft tissue loads can be solved at the same time in a co-simulation framework in order to couple the musculoskeletal dynamics and tissue mechanics. This model shows that computational method selection and the accurate acquisition of all knee DOFs are important in evaluating the loads of the soft tissue.

The aim of the present study was to predict the results of a HTO surgery on the medial condylar forces of a simulated patient with varus knee, to determine whether the osteotomy improves the conditions. It is possible to increase the pressure on the medial condyle after the osteotomy as a result of the superficial MCL stretching causing an increased rate of arthritis progression. We tried to find a threshold as a "critical correction angle" in which deformity correction after HTO surgery, without MCL release, leads to an increased medial compartment pressure due to ligament tightening. This problem was studied through modeling the gait of a single subject by using the OpenSim software.

It should be pointed out that, although in the methodology of this study, patient's modeling based on changing the muscles insertion point of a healthy subject using the coding environment of OpenSim was considered as an approximation, the results have not been validated as of now. In most of the previous studies, at first, approaches like statistical shape modeling (25,26) or tools like Mimics (27) have been utilized to build the 3-dimensional bone geometry of a specific subject based on the MRI of his/her lower body (28). Then, the bones have been grouped and muscles have been modeled using Meshlab (29) and NMS-builder (30,31), respectively. NMS-builder is an open-source software application to process biomedical data and create subject-specific musculoskeletal models for OpenSim. The subject-specific geometry OpenSim model can be used to analyze motion capture data of a patient by simulating the gait and testing the consequences of altered geometry by the surgical intervention. The effectiveness of the surgery on the patient's gait functionality and knee condylar force can be assessed using forward dynamics provided in the OpenSim. Therefore, further studies are needed to validate the outcomes of the current approximated methodology. Hence, the reader should exercise caution and not generalize these results till subject-specific geometry and larger sample size study are used.

Methods and Materials

A healthy 25-year-old male (175 cm height and 60 kg weight) without a varus or valgus knee deformity (thoroughly was inspected by a knee specialist) participated in this study. He signed the consent form before attending the test session. This study has been officially approved by the ethics committee of Tehran University. The following steps have been conducted to generate an OpenSim model based on Schmitz and Piovesan's model used in this study:

1. The kinematic data during gait was collected by placement of 41 retroreflective markers on the participant's body landmarks. Two markers were placed on the right and left anterior superior iliac spines (ASIS) and one marker on the sacrum (L4-L5). Four other markers were placed on the following locations of the particular limb under consideration: greater trochanter, directly lateral to the estimated average axis of rotation of the knee joint, lateral malleolus, space between the second and third metatarsal heads, mid-thigh, and another on the mid-shank) as same as used by Schmitz and Piovesan (24) at sample rates of 120 Hz to find joint angles. Then, markers data was filtered after gap-filling (estimating of missed markers trajectories) using the Butterworth low pass filter with cut off frequency of 6 Hz. Ground reaction force data was also collected during gait by using two Kistler force plates of type 9287 synchronized with the motion capture system (sample rates of 120 Hz).
2. These kinematics (markers trajectories) and kinetics (ground reaction forces) data were used for running the OpenSim model. At first, in order to record the static calibration data of the person, he is asked to stand on the force plate motionless for 30

Figure 1.The photograph of walk way layout showing marker

seconds. The gait test was consisted of 5 trials with 10 seconds walking. The subject's weight and height were recorded during the static calibration and marker data were applied on the OpenSim model using the scaling approach provided in the OpenSim. Marker-based scaling in the OpenSim included a linear resizing of the bones of a generic model (in the Schmitz and Piovesan's model) based on the weight, height, and coordination of markers in a specific subject. The shape of bones did not change during the scaling procedure.

3. Figure 1 shows the subject with markers placed on his body.
4. Furthermore, MotoNMS software was used to transform the recorded Vicon format of marker position data "csv" to a "trc" format file and force plate data to a "mot" format file to be prepared for the OpenSim modeling. Using inverse kinematics, the joints angles related to the subject were extracted. Then, by using the Reduce Residuals Algorithm (RRA), kinetic and kinematic data was prepared as an input for the next step in which the muscle forces were calculated by applying the Computed Muscle Control (CMC) approach. Using CMC, muscle activation data and ligaments forces on condyles during gait were provided. CMC performed this (activation estimation) by using a combination of proportional-derivative (PD) control and static optimization (32, 33). The aforementioned steps were conducted on data collected from a healthy participant. The next step was calculating the kinematics and kinetics of a patient who did HTO. In the present study, instead of using imaging data of a real HTO patient, we numerically altered insertion points to simulate the HTO surgical intervention with different degree of varus deformity in the Schmitz and Piovesan's model (23, 24).

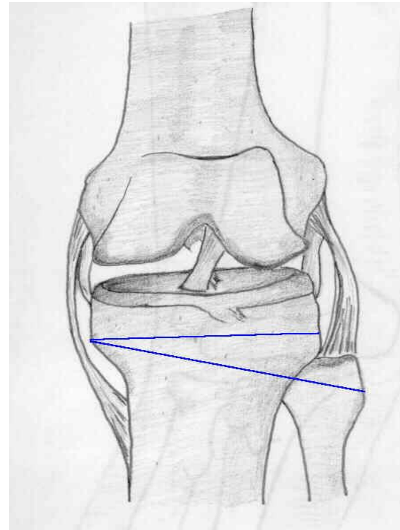

Figure 2. Medial opening wedge HTO of the knee joint for a varus patient (34)

The model developed by Schmitz and Piovesan, using the discrete element method embedded in the OpenSim model, provides the user with the ability to obtain force at the level of the condyles. After analyzing the gait of the healthy person mentioned in the model, because we could not change the geometric structure of the bone in this model, we changed the insertion point of the muscles so that the effect of the operation was simulated with the different medial proximal tibial angles and under each defined scenario for each angle, the gait test was run again. In this way, the results of model with the various surgical conditions were compared with the healthy subject.

In order to analyze the effect of HTO on the condylar forces of the knee joint, it is necessary to study how the HTO affects the tibia, muscles, ligaments, and tendons of the knee. In the healthy participant, none of the ligaments or tendons were removed, and only the geometry of the tibia was changed. As the focus of this research was on contact and ligament forces, the only geometric parameters were the origins and insertion points of the muscles, tendons, and ligaments on the tibia. Some of the insertions may change after the HTO surgery, therefore, the new coordinates of these insertions must be found. Figure 2 shows medial opening wedge HTO surgery of the knee joint model of a varus patient.

The cutting plane is defined according to the widely accepted surgical technique and in order to achieve correction, the proximal section of the tibia was fixed, and the distal section was rotated about the lateral cortex (24). Insertion points which are inferior to the cutting plane were rotated within the Schmitz and Piovesan's model; therefore, the corresponding model of each deformity was produced in

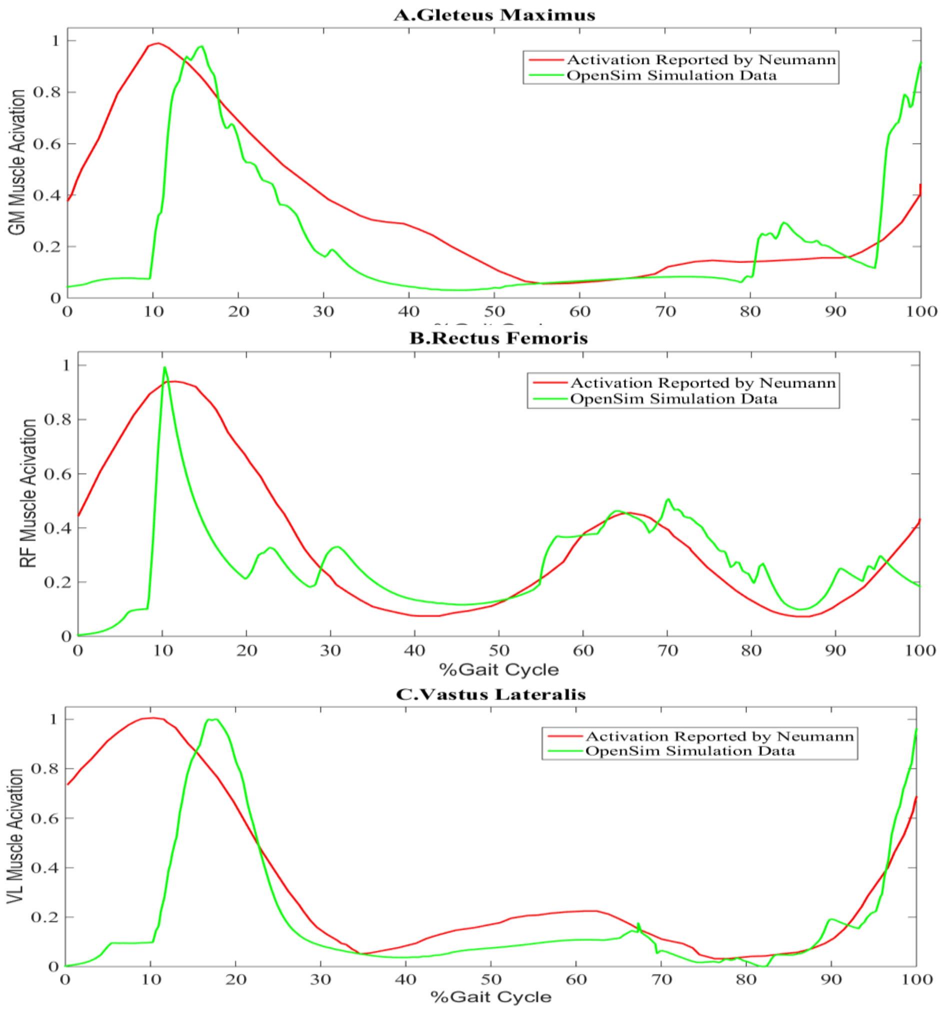

Figure 3. Comparing the OpenSim simulation results (green line) to EMG data reported by literature (red line) for A) Gluteus Maximus (GM), B) Rectus Femoris (RF) and C) Vastus Lateralis (VL) muscles, for a gait cycle

OpenSim. By modifying the connecting points of the muscle, tendon, and ligament affected by the osteotomy, different scenarios with varus deformities were simulated. These scenarios included changes of Medial Proximal Tibial Angle (MPTA) from 74 to 80 degrees corresponding to deformities of 16 to 10 degrees, respectively. These scenarios have been considered in advice of an orthopedic surgeon from Imam Hossein Hospital at Tehran, Iran. In the model, new connective points have been defined for each scenario, and each scenario model with different degrees of varus deformity was produced in the OpenSim. Additionally, due to the model deformity, the length of the ligaments whose insertion points lie on either side of the osteotomy plane changed. The extent of deformity correction determines the change in length of these ligaments post operatively. The change in the length of these ligaments is equal to the width of the tibia plateau in the Anterior-Posterior (AP) view at the deformity correction angle. The variation in the length obtained in varus patients has been subtracted from the initial length of the MCL ligaments to reproduce the initial state of patients.

Previous studies have utilized a two-dimensional inverse kinematic and kinetic approach from gait data obtained from patient populations to determine the internal kinetics structures of the knee joint soft tissue (35-37). Changing the position of muscle insertion points to simulate varus scenarios with various degrees of deformities can change the force in the muscles connected to the tibia. This can result in the change of the contact pressure exerted on the condyles which itself is comprised of forces exerted by the muscles, ligaments, and tendons as well as the force transfer through the joint. Using distinct degrees of deformity and post-operative conditions to simulate scenarios, the force applied to the medial condyle and the force created in the LCL can be compared in pre and post-operative conditions. In the stance phase, due to the

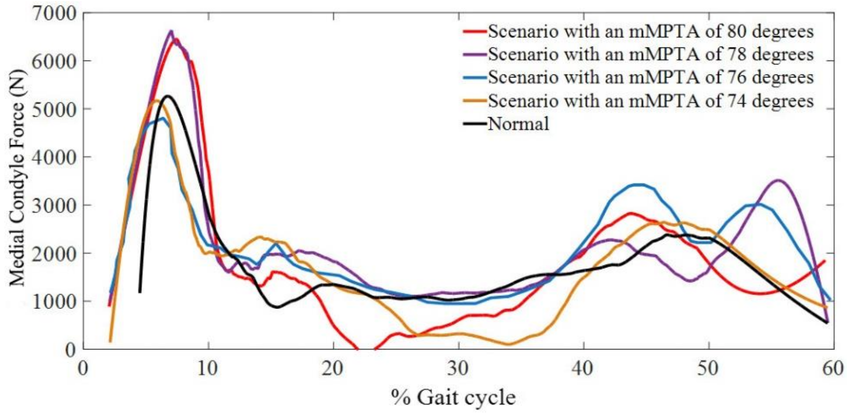

Figure 4. Forces exerted in the medial condyle for different varus scenarios with various deformities

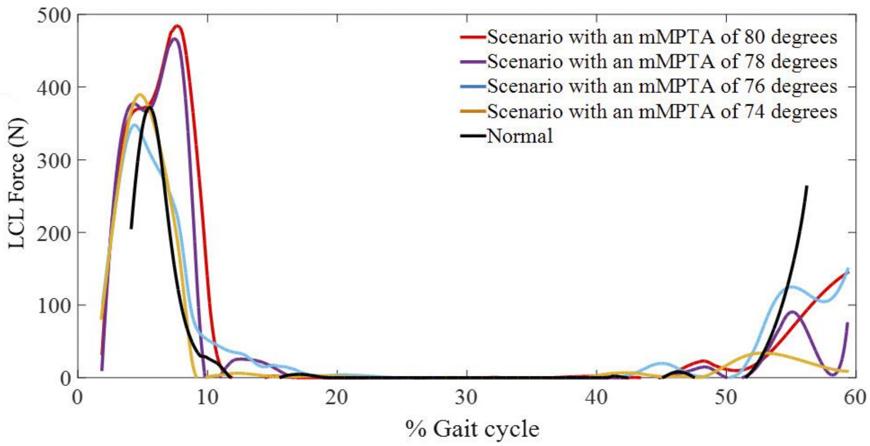

Figure 5. Lateral collateral ligament (LCL) force for different varus scenarios with various deformities

application of ground force, femoral and tibial condyles are expected to bear a greater amount of load and that is why the focus of this paper was on the stance phase. In this study, additionally, the activity of the Gluteus Maximus, Rectus Femoris, and Vastus Lateralis muscles, obtained from EMG data reported in the literature, were compared with muscles activation results obtained from simulation in the OpenSim software, for verification purposes.

The Rectus Femoris muscle originates on the anterior inferior iliac spine and is inserted on the tibial tuberosity via the quadriceps tendon, patella, and patellar ligament which produces the extension of the knee (38). Since HTO causes a rotation in the tibia distal region, the Rectus Femoris muscle remains constant relative to the femur, and the only part which changes is the distance between the patella and the tibial tuberosity (39). In some types of HTO like Z-osteotomy, tibial tuberosity is more proximal than the cutting plane, since the forces exerted by the quadriceps muscle and the direction of action are constant in the patella ligaments; so, the knee flexion rate for patients before and after surgery is constant (40). In other types of HTO, the cutting plane is more proximal than the tibial tuberosity and causes its displacement. In these surgeries, the maximum opening of the cortex is about 15 mm (40).

To study the exerted force on the medial condyles during the gait cycle, the force tolerated by the collateral ligaments must also be examined. In a varus deformity, the majority of the force exerted on the knee joint is tolerated by the medial condyle of the knee and the LCL. In a normal gait cycle, the compressive force exerted on the medial or lateral condyles of the knee and the tensile force present in the collateral ligaments are in equilibrium. In this research, the medial condyle and LCL forces are investigated in the presence of the varus deformity and based on the geometry of the knee; these are expected to increase at greater degrees of deformity.

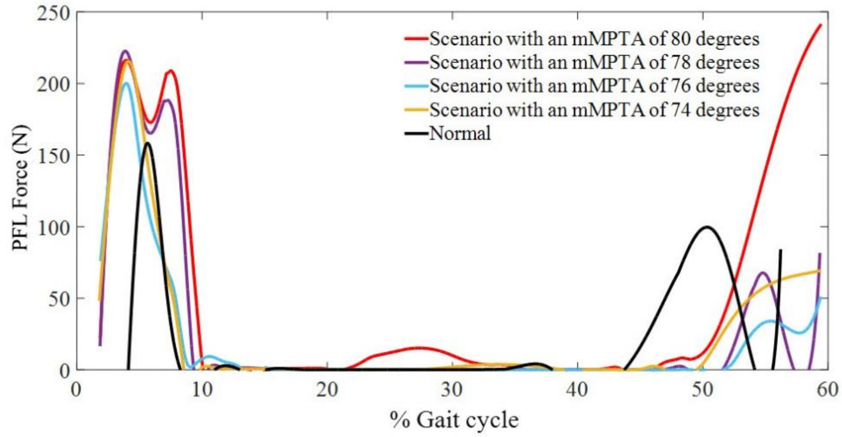

Figure 6. Popliteofibular ligament (PFL) force for different varus scenarios with various deformities

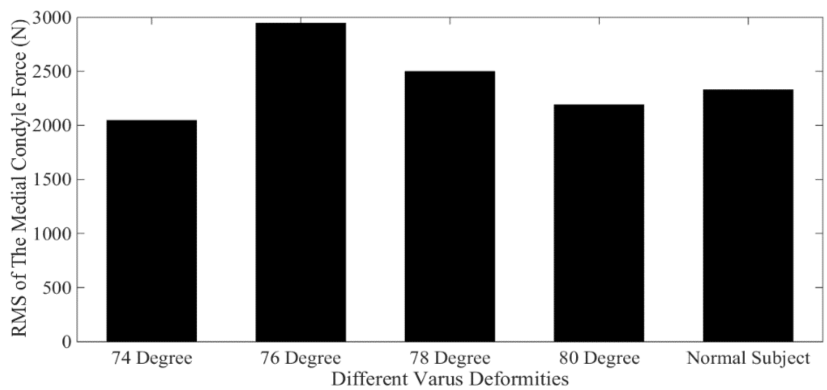

Figure 7. Root-mean-square of the forces exerted in the medial condyle for different varus scenarios with various deformities

Results and Discussion

The results of comparison of Gluteus Maximus, Rectus Femoris, and Vastus Lateralis activity obtained from simulation (OpenSim) with these muscles activity obtained from EMG data, which were reported in the literature (41-45) indicated a good agreement as shown in Figure 3.

Forces exerted in the medial condyle (Figure 4) and force of LCL (Figure 5) and PFL (Figure 6) were predicted during a gait cycle for different varus scenarios with various deformities. It is important to note that all 3 forces (ligaments and condylar forces) which lead to equilibrium in the knee for the varus patient must be consider together, in order to be able to successfully analyze the knee. As shown in Figure 4, the maximum force in the medial condyle corresponding to heel contact and foot flat part of the gait cycle (10%) for patients with a MPTA of 74 and 76 degrees (corresponding to deformities of 16 and 14 degrees, respectively) is lower than a healthy person. After surgery, this force was increased. The reason was the change in the length of the MCL while simulating varus patients with various deformities. Vertical force in the medial condyle after correction decreased for a scenario with an MPTA of 78 degrees, but increased for the scenario with an MPTA of 76 degrees. The minimum amount of force tolerated in the medial condyle and LCL occurred at an MPTA between 78 and 76 degrees, that after corrective osteotomy, the force on the medial condyle increased due to the stretching of MCL. The difference in the exerted pressure on the medial condyle in the scenario with a 76 degree deformity was less than that of a deformity with 74 degrees as compared with the healthy subject. In order to remove the noise from the OpenSim output, the “sgolyfilt” function in MATLAB (Mathworks, Inc., Natick, MA, USA) was utilized.

The results showed that the pressure on the medial condyle and the tension in the LCL was decreased in the 78 and 80 degrees of deformity. It is reported that beyond the aforementioned degrees of deformity, the femoral condyle comes into contact with the eminence of the tibia. This may explain the reduction of vertical force in the medial condyle in these degrees. In addition, this finding was further supported by an increase in the shear force reported by OpenSim. This may be caused due to not modeling the joint capsule ligament in this simulation. As shown in Figure 4, an MPTA of 76 degrees of deformity had the lowest peak force in medial compartment as well as tension in the LCL and PFL. The greatest amount of increase in these forces occurred when it is brought to normal conditions (post osteotomy). On the other hand, by calculating the root-mean-square (RMS) of the force exerted onto the medial condyle in all deformities compared to this value in a normal person, a scenario with an MPTA of 78 and 76 degrees had the highest levels compared with a normal subject, as shown in Figure 7. According to the Figure 4, the force exerted on the medial condyle in a scenario with a deformity of 78 degrees had the highest levels compared to this value in the normal subject and when this scenario is brought to normal conditions (post osteotomy), a lower force was exerted on the medial compartment. As a result, the most critical situation occurs in a scenario with an MPTA of 76 degrees and the surgeon for osteotomy should release the MCL in a scenario with an MPTA greater than 76 degrees for the subject studied in the present research.

Conclusion

In this study, after thorough investigation of the pressure in the medial side of the knee and tension in the LCL and PFL in a simulated subject, a scenario with a 76 degree MPTA corresponding to 14 degrees of deformity was found to be a critical point beyond that the surgeon must perform an MCL release. It was shown if a surgeon chooses not to perform the MCL release, an increased tension in the MCL may lead to an even greater pressure in the medial condyle after the medial high tibial osteotomy.

Limitations

The movements and degrees of freedom of the knee joint are a function of ligaments, tendons, and joint capsules which help to control the medio-lateral motion of the knee joint. In reality, therefore, the contact of the femoral condyles with the eminence of the tibia rarely happens. However, in this simulation, due to not modelling the articular capsule, we expected that vertical force was dropped in the medial condyle for some deformities; which is due to the collision of the femoral condyles with the eminence of the tibia. Additional studies are required to confirm this hypothesis. It should be pointed out that in order to correct the force direction of the knee joint, it was assumed that after the osteotomy, the force passes through the center of the knee joint. In order to exert the tightening of the MCL after osteotomy, its initial length was simulated in different scenarios with various deformities. In the other word, in order to compare the force on the medial condyle for different deformities, the tightening of the MCL after osteotomy in scenarios with various deformities were simulated in the form of the correction of the length of the MCL before osteotomy. Since the purpose of this study was to find the limit of increasing the force on the medial condyle in correcting various deformities, comparison of force in different deformities was important. Change in the initial length of the MCL before surgery may change the force of the medial condyle compared with the actual patient. However, when comparing the force applied to the medial condyle, it does not make a mistake in various deformities.

An important limitation of the present study was using the kinematic data of a healthy participant to simulate patient behavior by changing the muscles insertion point. It is necessary to do this process for real patients individually and experimentally, in order to calculate the force on the medial condyle, based on their specific geometry, the amount of muscle strength, and other influential factors. Therefore, it is suggested that future studies analyze experimentally captured patients gait data to obtain more accurate results.

Acknowledgments

None

Conflict of interest:

None

Funding support:

None

Authors' contributions:

All authors made substantial contributions to conception, design, acquisition, analysis and interpretation of data.

References

1. Hsu, R.W., et al., Normal axial alignment of the lower extremity and load-bearing distribution at the knee. Clinical orthopaedics and related research, 1990. 255: p. 215-227.
2. Paley, D., Principles of Deformity Correction. 1 ed.: Springer-Verlag Berlin Heidelberg. XXV, 806.
3. Andriacchi, T.P., Dynamics of knee malalignment. The Orthopedic clinics of North America, 1994. 25(3): p. 395-403.
4. Baliunas, A., et al., Increased knee joint loads during walking are present in subjects with knee osteoarthritis. Osteoarthritis and cartilage, 2002. 10(7): p. 573-579.
5. Teichtahl, A., et al., Change in knee angle influences the rate of medial tibial cartilage volume loss in knee osteoarthritis. Osteoarthritis and Cartilage, 2009. 17(1): p. 8-11.

6. Foroughi, N., R. Smith, and B. Vanwanseele, The association of external knee adduction moment with biomechanical variables in osteoarthritis: a systematic review. The Knee, 2009. 16(5): p. 303-309.
7. Barrios, J.A., et al., Lower extremity walking mechanics of young individuals with asymptomatic varus knee alignment. Journal of Orthopaedic Research, 2009. 27(11): p. 1414-1419.
8. Bergmann, G., et al., Hip contact forces and gait patterns from routine activities. Journal of biomechanics, 2001. 34(7): p. 859-871.
9. Papaioannou, G., et al., Patient-specific knee joint finite element model validation with high-accuracy kinematics from biplane dynamic Roentgen stereogrammetric analysis. Journal of Biomechanics, 2008. 41(12): p. 2633-2638.
10. Yao, J., et al., Sensitivity of tibio-menisco-femoral joint contact behavior to variations in knee kinematics. Journal of Biomechanics, 2008. 41(2): p. 390-398.
11. Hull, M., et al., A finite element model of the human knee joint for the study of tibio-femoral contact. Journal of biomechanical engineering, 2002. 124(3): p. 273-280.
12. Penrose, J., et al., Development of an accurate three-dimensional finite element knee model. Computer Methods in Biomechanics & Biomedical Engineering, 2002. 5(4): p. 291-300.
13. Li, G., et al., A validated three-dimensional computational model of a human knee joint. Transactions-American Society of Mechanical Engineers Journal of Biomechanical Engineering, 1999. 121(6): p. 657-662.
14. Perie, D. and M. Hobatho, In vivo determination of contact areas and pressure of the femorotibial joint using non-linear finite element analysis. Clinical Biomechanics, 1998. 13(6): p. 394-402.
15. Bei, Y. and B.J. Fregly, Multibody dynamic simulation of knee contact mechanics. Medical engineering & physics, 2004. 26(9): p. 777-789.
16. Arnold, A.S., et al., Accuracy of muscle moment arms estimated from MRI-based musculoskeletal models of the lower extremity. Computer Aided Surgery, 2000. 5(2): p. 108-119.
17. DeFrate, L.E., et al., In vivo tibiofemoral contact analysis using 3D MRI-based knee models. Journal of biomechanics, 2004. 37(10): p. 1499-1504.
18. Heller, M., et al., A new model to predict in vivo human knee kinematics under physiological-like muscle activation. Journal of biomechanics, 2007. 40: p. S45-S53.
19. Hashemi, J., et al., The geometry of the tibial plateau and its influence on the biomechanics of the tibiofemoral joint. The Journal of Bone and Joint Surgery. American volume., 2008. 90(12): p. 2724.
20. Yang, N.H., et al., Effect of frontal plane tibiofemoral angle on the stress and strain at the knee cartilage during the stance phase of gait. Journal of Orthopaedic Research, 2010. 28(12): p. 1539-1547.
21. Agneskirchner, J.D., et al., The effects of valgus medial opening wedge high tibial osteotomy on articular cartilage pressure of the knee: a biomechanical study. Arthroscopy: The Journal of Arthroscopic & Related Surgery, 2007. 23(8): p. 852-861.
22. Gaasbeek, R.D., et al., Correction accuracy and collateral laxity in open versus closed wedge high tibial osteotomy. A one-year randomised controlled study. International orthopaedics, 2010. 34(2): p. 201-207.
23. Schmitz, A. and D. Piovesan, Development of an open-source, discrete element knee model. IEEE Transactions on Biomedical Engineering, 2016. 63(10): p. 2056-2067.
24. https://simtk.org/projects/kneemodel
25. Modenese, L., Montefiori, E., Wang, A., Wesarg, S., Viceconti, M. and Mazzà, C., 2018. Investigation of the dependence of joint contact forces on musculotendon parameters using a codified workflow for image-based modelling. *Journal of biomechanics*, *73*, pp.108-118.
26. Salhi, A., Burdin, V., Boutillon, A., Brochard, S., Mutsvangwa, T. and Borotikar, B., 2020. Statistical Shape Modeling Approach to Predict Missing Scapular Bone. *Annals of Biomedical Engineering*, *48*(1), pp.367-379.
27. Karimpour, M. and Rahimian, R., 2019. Biomechanical Analysis of Effect of Inclination Angle on Stress Distribution in the Acetabulum and Acetabularcup after Total Hip Arthroplasty. *Modares Mechanical Engineering*, *19*(3), pp.559-567.
28. Prinold, J.A., Mazzà, C., Di Marco, R., Hannah, I., Malattia, C., Magni-Manzoni, S., Petrarca, M., Ronchetti, A.B., De Horatio, L.T., van Dijkhuizen, E.P. and Wesarg, S., 2016. A patient-specific foot model for the estimate of ankle joint forces in patients with juvenile idiopathic arthritis. *Annals of biomedical engineering*, *44*(1), pp.247-257.
29. https://www.meshlab.net
30. http://www.nmsbuilder.org
31. Valente, G., Pitto, L., Testi, D., Seth, A., Delp, S.L., Stagni, R., Viceconti, M. and Taddei, F., 2014. Are subject-specific musculoskeletal models robust to the uncertainties in parameter identification?. *PLoS One*, *9*(11), p.e112625.
32. Delp, Scott L., et al. "OpenSim: open-source software to create and analyze dynamic simulations of movement." IEEE transactions on biomedical engineering 54.11 (2007): 1940-1950 .
33. Thelen, Darryl G., and Frank C. Anderson. "Using computed muscle control to generate forward dynamic simulations of human walking from experimental data." Journal of biomechanics 39.6 (2006): 1107-1115.
34. http://www.wheelessonline.com/ortho/lateral_plateau_frx_approach_fixation.
35. Paley, D., Principles of Correction for Monocompartmental Arthritis of the Knee. Brown TE, Cui.
36. Shelburne, K.B., M.R. Torry, and M.G. Pandy, Contributions of muscles, ligaments, and the ground-reaction force to tibiofemoral joint loading during normal gait. Journal of orthopaedic research, 2006. 24(10): p. 1983-1990.
37. Kakihana, W., et al., Effects of laterally wedged insoles on knee and subtalar joint moments. Archives of physical medicine and rehabilitation, 2005. 86(7): p. 1465-1471.
38. Glitsch, U. and W. Baumann, The three-dimensional determination of internal loads in the lower extremity. Journal of biomechanics, 1997. 30(11): p. 1123-1131.
39. Martini, F., M.J. Timmons, and R.B. Tallitsch, Human Anatomy. 2012: Pearson Benjamin Cummings.
40. Lee, Y.S., et al., Achievement of targeted posterior slope in the medial opening wedge high tibial osteotomy: a mathematical approach. Annals of biomedical engineering, 2010. 38(3): p. 583-593.
41. Neumann, D.A., Kinesiology of the Musculoskeletal System-E-Book: Foundations for Rehabilitation. 2013: Elsevier Health Sciences.
42. Perry, J. and Davids, J.R., 1992. Gait analysis: normal and pathological function. *Journal of Pediatric Orthopaedics*, *12*(6), p.815.
43. Winter, D.A., 1991. *Biomechanics and motor control of human gait: normal, elderly and pathological.*
44. Winter, D.A., 1995. Anatomy, biomechanics and control of balance during standing and walking. Waterloo, Canada: Graphic Services.
45. Wootten, M.E., Kadaba, M.P. and Cochran, G.V.B., 1990. Dynamic electromyography. II. Normal patterns during gait. Journal of Orthopaedic Research, 8(2), pp.259-265.