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The patellofemoral joint and the total knee replacement

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Abstract

The biomechanical investigation of the entire knee joint is essential in understanding the joint function and interaction between various components in both intact and perturbed condition. In the most clinically successful cases many patients treated by total knee replacement (TKR) cannot achieve normal joint function over time.

The knee joint is the most complicated joint in the human body. The model of the knee joint with total knee replacement is based on the finite element method. The attention of present study was focused on the patellofemoral joint and the patella tracking with different Q-angle and condylar twist angle. The patellofemoral joint is frequently the source of the knee anterior pain related to disturbances in normal tracking.

Keywords: knee, patellofemoral joint, FEM analysis

1. Introduction

Patello-femoral complications have been described following total knee replacement (TKR) surgery.

The aim of this study is to find the reason of the patella subluxation and pain in the knee after TKR. The attention has been focused on the different Q-angle and condylar twist angle.

2. Knee anatomy

To better understand how knee problems occur, it is important to understand some of the anatomy of the knee joint and how the parts of the knee work together to maintain normal function.

The knee is the most complex joint in the body and is made up of a combination of bones, muscles, tendons, ligaments, and other soft tissue. Knees are the largest, heaviest, and strongest joints in the body, providing mobility and support to carry almost half your body's weight.

The knee joint can be divided into the femoro-patellar joint and femoro-tibial joint.

2.1. The knee bones

The knee joint is made up of four bones, which are connected by muscles, ligaments, and tendons. The femur is the large bone in the thigh. The tibia is the large shin bone. The fibula is the smaller shin bone, located next to the tibia. The patella, otherwise known as the knee cap, is the small bone in the front of the knee (see fig. 1). It slides up and down in a groove in the femur (the femoral groove) as the knee bends and straightens.

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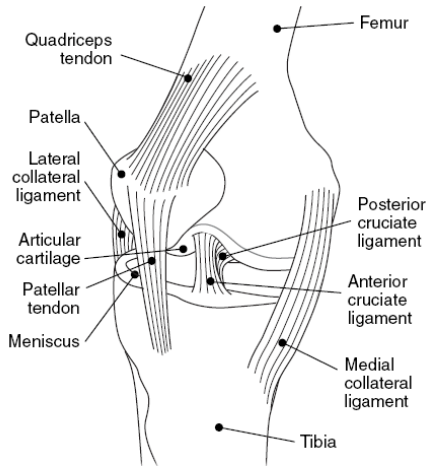

Fig. 1. Lateral view of the knee.

2.2. Muscle, ligaments and cartilage

Ligaments are like strong ropes that help connect bones and provide stability to joints. In the knee, there are four main ligaments. On the inner (medial) aspect of the knee is the medial collateral ligament (MCL) and on the outer (lateral) aspect of the knee is the lateral collateral ligament (LCL). The other two main ligaments are found in the center of the knee. These paired ligaments are called the anterior cruciate ligament (ACL) and the posterior cruciate ligament (PCL). They are called cruciate ligaments because the ACL ”crosses” in front of the PCL (see fig. 1).

These are cartilaginous elements within the knee joint which serve to protect the ends of the bones from rubbing on each other and to effectively deepen the tibial sockets into which the femur attaches. They also play a role in shock absorption. There are two menisci in each knee, the medial meniscus and the lateral meniscus. Either or both may be cracked, or torn, when the knee is forcefully rotated and/or bent.

Tendons connect muscles to bone. The strong quadriceps muscles on the front of the thigh attach to the top of the patella via the quadriceps tendon. This tendon covers the patella and continues down to form the patellar tendon. The patellar tendon in turn, attaches to the front of the tibia. The hamstring muscles on the back of the thigh attach to the tibia at the back of the knee. The quadriceps muscles are the main muscles that straighten the knee. The hamstring muscles are the main muscles that bend the knee (see more in [4], [3]).

2.3. Kinematics of the knee joint

Three basic plane (sagittal, frontal, transversal) placed in the middle of the knee create three basic axes. Along these axes, the knee can perform six different movement - the knee joint has 6 degrees of freedom, but predominant movement occurs in the sagittal plane. For the sake of simplicity, the joint can be thought of as a hinge.

Normal functional range of knee movement is $20^\circ - 120^\circ$. Flexion beyond 120° is entirely passive. Active flexion can be increased beyond 120° by hip flexion as the hamstrings are relaxed and become more efficient knee flexors.

The knee is stable in extension and unstable in flexion. Knee can rotate on flexion, this is maximal at 90° of flexion. Knee flexion, while allowing knee rotation, prevents hip rotation.

The relative movement of the condyles and the tibio-femoral contact surfaces are rather complex. Femur both slides and rolls on tibia. If it only rolled then femoral condyle would have completely dislocated from the tibial platform (it does to some extent) as the length of circumference of femoral condyle is twice the length of the tibial condyle. If it only slide, then flexion would have been prematurely stopped by impact of femur on the posterior border of the tibial condyle. It was thought that both condyles rollback and slide. Recent MRI evidence would suggest that the medial femoro-tibial articular surfaces is somewhat congruent to allow any effective sliding and as such the medial condyle can only rotate like the ball in a socket and produce flexion and longitudinal rotation. On the other hand, as the lateral surface is comparatively flat, it allows the lateral condyle to roll and slide. It is now felt that the medial condyle flexes and rotates longitudinally, whereas the lateral condyle undergoes both roll and glide.

2.3.1. Patello-femoral joint and Q-angle

The biomechanics and stability of the patello-femoral joint is affected by patellar tendon length and the Q-angle. The Q-angle is the angle formed by a line drawn from the spine iliac anterior superior to the central patella and a second line drawn from the central patella to the tibial tuberosity. An increased Q-angle is a risk factor for patellar subluxation. Normally Q-angle is 10 - 12 deg for males and 14 - 16 deg for females (see fig. 2). The Q-angle is increased by genu valgum, external tibial torsion, laterally positioned tibial tuberosity etc.

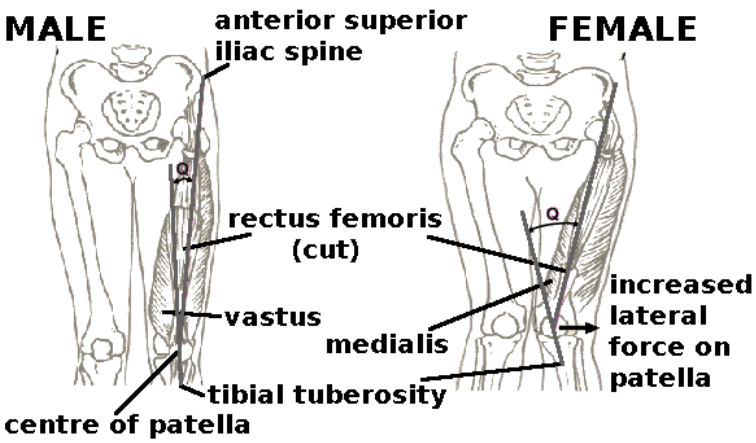

Fig. 2. Q-angle measurements.

2.4. The total knee replacement

Attempts to replace the knee joint with an arthroplasty have been made for at least 140 years. A breakthrough appeared about 50 years ago when the hinged knee replacement were introduced followed by numerous designs. Important factors to speed up this development were the concomitant introduction of polyethylene and bone cement originally employed in arthroplasties of the hip.

Today, total knee replacement (TKR) is a successful surgical procedure. The older patients can expect that the implant will last for the rest of their lives. However, both in vitro and in vivo studies have shown that knees with implanted prostheses have an abnormal kinematic behaviour. Theoretically there are different ways to alter the kinematics of artificial knees by

changing the configuration of the surfaces of the joint, the addition of stabilising devices such as a central spine or by resection of the PCL or not. These modifications of a TKR can also have other effects. The wear of the polyethylene might change due to changes in the pattern of motion and load. The forces transmitted to the implant/bone or cement/bone interfaces can be expected to mirror the inherent stability of the design and will thus have an influence on the fixation of the components. Finally, but not least, the design of the joint will most probably also influence the clinical function and the subjective opinion of the patient about the outcome of the operation (fig 3).

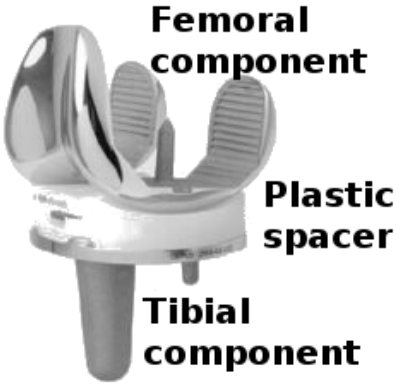

Fig. 3. A total knee replacement prosthesis.

2.4.1. Condylar twist angel

The position of the femoral component can influence the patello-femoral joint and the biomechanics of this joint. One possibility to avoid the patello-femoral problems is to rotate the fomoral component. The condylar twist angle (CTA) has been defined as a angle between the trans-epicondylar line and the posterior condylar axis. Attention of this study is concentrated on the position of the patella for different condylar twist angle and different Q-angle.

3. Description of the mechanical model

3.1. Creation of the mechanical model

The biomechanical model of the virtual lower extremity, developed by [6], is based on the CT and RGB photographs from Visible Human project. The base stone for this model of the lower extremity are the CT and RGB photographs, which were processed by special software to generate volume of individual bone parts (see fig. 4a). These grids were adjusted to create acceptable grid for biomechanical simulation. Consequently, the model of TKR has been embedded into the model of leg (fig. 4b).

Consequently, the material properties (Young's modul, Poisson number etc.) have to be defined for different biological tissues (bones, ligaments, tendon - [2], [7]).

3.2. Contact

The parts of the model move under the influence of a loading. The penetration of structures may be a result of this motion. To avoid these penetrations it is necessary to define mutual contacts among the individual structures in the model. The contact interfaces require the definitions of the "master" and "slave" segments, that may get into contact.

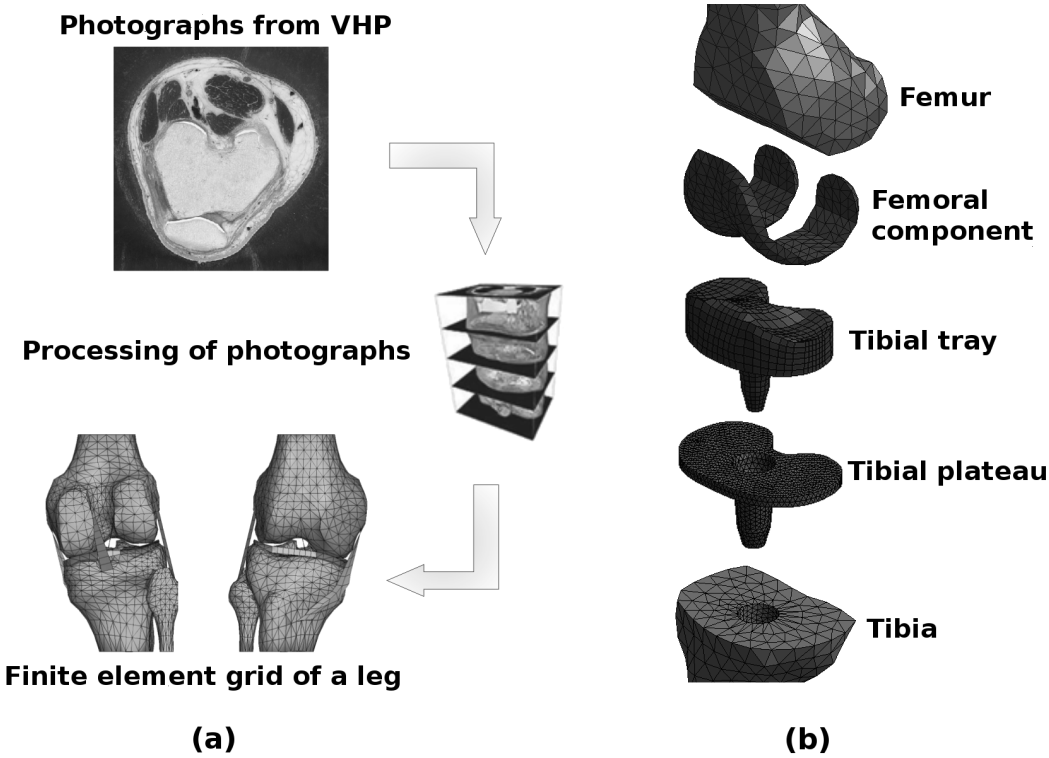

Fig. 4. Creation of biomechanical model of the lower extremity.

During the modelling the following type of the contacts have been chosen

- Node-Surface Tied Interface - for this interface the slave nodes can be tied at a certain distance from the master surface - tab. 1.
- Sliding interface - model of the interaction between structures or part of structures that are not permanently connected - tab. 2

Tied Interface name
femur vs. femoral_component
tibial_tray vs. tibia

Tab. 1. Tied interfaces in the model of the lower extremity.

Sliding Interface name
patella vs. femoral_component
femoral_component vs. tibial_plateau
tibial_plateau vs. tibial_tray
MCL vs. bone
PCL vs. bone

Tab. 2. Sliding interfaces in the model of the lower extremity.

3.3. Loading

Vertical and horizontal forces take effect on the patello-femoral joint. These forces produce the joint pressure and guide the patella within femoral groove. The vertical component of these forces is composed of the quadriceps femoris muscle and the ligamentum patellae. The active part of the horizontal forces consists of the medial and lateral vastus muscles [1].

This computational model of the lower extremity is loaded via body weight and muscle force. The body weight is replaced by ground reaction force and is acting on the COG of the tibia. For simplification, the muscle force is represent only by acting of the musculus quadriceps femoris. The force activity of this muscle has been replaced by concentrated loading. The direction of concentrated loading is direction in which the individual parts of the muscle quadriceps femoris (musculus rectus femoris, musculus vastus lateralis and musculus vastus medialis) act.

The horizontal and vertical part of the ground reaction force are taken into account (see fig. 5 and fig. 6).

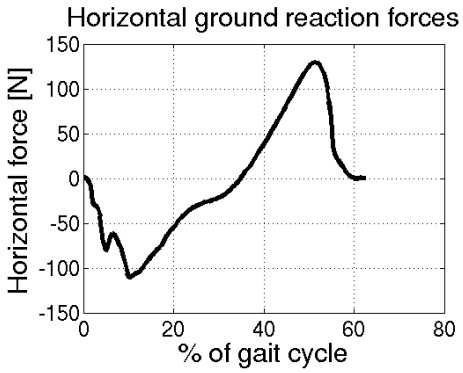

Fig. 5. Horizontal part of ground reaction force (Taken from [5]).

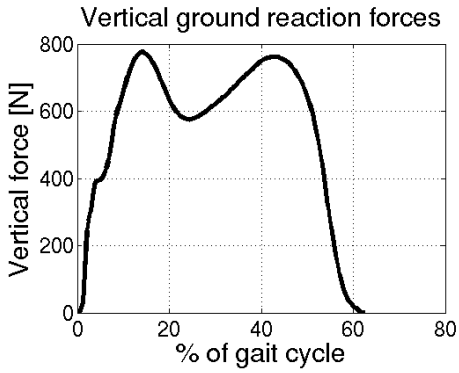

Fig. 6. Vertical part of ground reaction force (Taken from [5]).

3.4. Results

In this study the attention is focused on the patella tracking in the knee with total knee replacement. Hence, it was observed the position of the patella after loading, contour of the static pressure and the contact force among patella and femoral component of the knee replacement.

It has been performed several simulations for flexion angle $0°$, $30°$ and $60°$. For each flexion angle it has been prepared static model for different Q-angle ($10°$, $14°$) and for CTA - condylar twist angle ($0°$ and $5°$).

In the contour of the static pressure it can be seen the position of the patella and also the contact area between the femoral component and tibial component. In the figures 7, 8 and 9 is possible to see the change of the contact area between patella and femoral component of the TKR. For CTA equal to $5°$ the contact area is placed more laterally. The static pressure for CTA $5°$ is higher. This could mean bigger pain for the patients.

In the figure 10, 11, 12 and 13 is possible to see the contact force magnitude between patella and femoral component of the TKR. The force is higher than body weight. This is caused by muscle force and higher flexion angle.

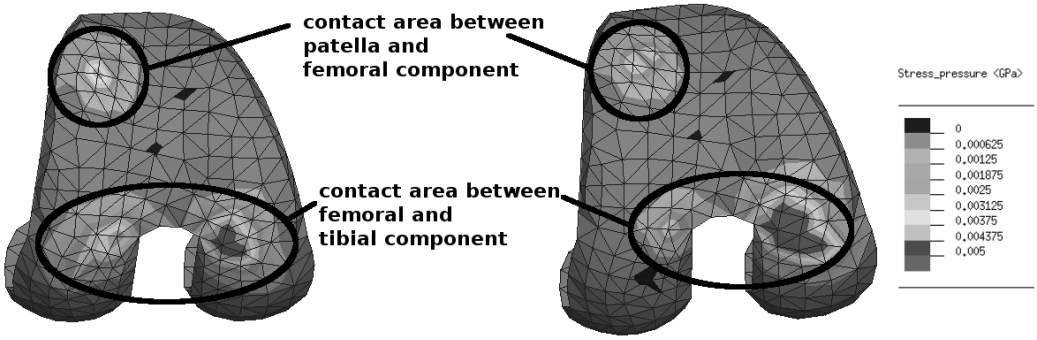

Fig. 7. Contour of static pressure for flexion angle 0°, Q-angle 10°, CTA 0° (left), and for CTA 5° (right).

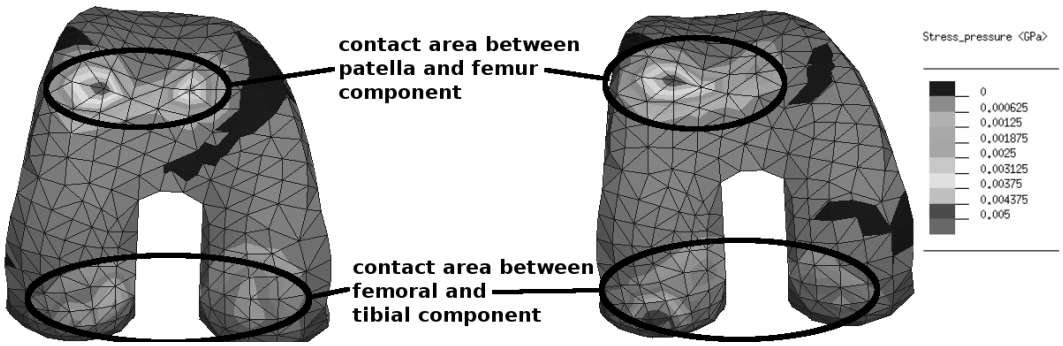

Fig. 8. Contour of static pressure for flexion angle 30°, Q-angle 10°, CTA 0° (left), and for CTA 5° (right).

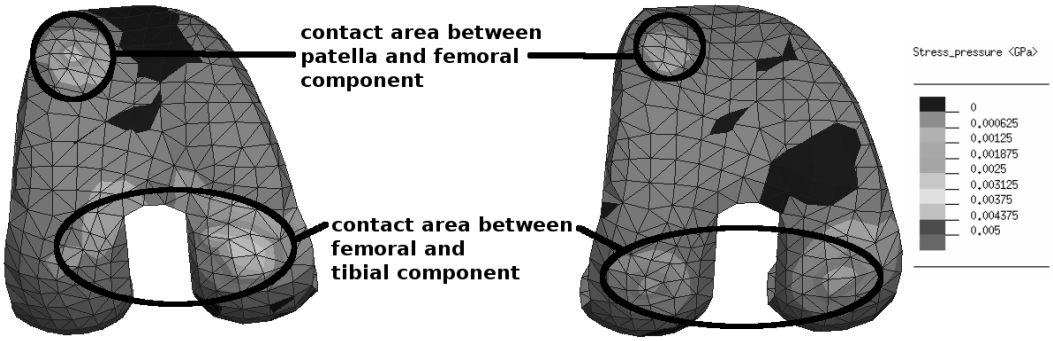

Fig. 9. Contour of static pressure for flexion angle 0°, Q-angle 14°, CTA 0° (left), and for CTA 5° (right).

4. Conclusion

It has been prepared model of the lower extremity for simulation of the patella behavior for different flexion, Q-angle and CTA. The results show the location of the maximal pressure stress and the magnitude of contact force between femoral component of the TKR and patella. From the results, the contact area between femoral component and tibial component of the TKR is also visible.

In the future it will be performed the simulation for different type of the total knee replacement. Also it will be performed the simulations for other flexion, Q-angle and CTA so that can be found the relations between these angles.

In the next phase, the attention it will be focus on the failure of the total knee replacement for genu valgum and varus.

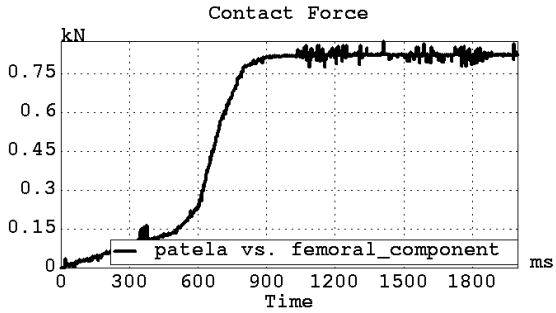

Fig. 10. The contact force magnitude for flexion angle 0°, Q-angle 10°, CTA 0°.

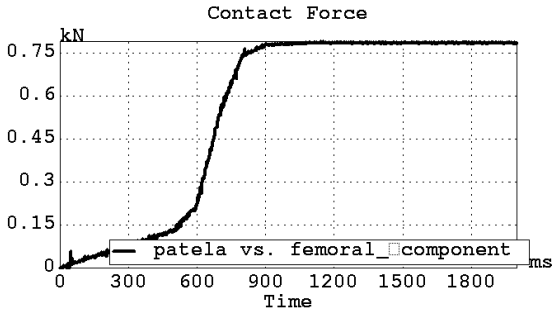

Fig. 11. The contact force magnitude for flexion angle 0°, Q-angle 10°, CTA 5°.

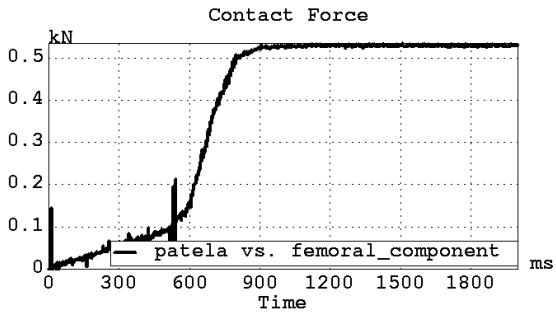

Fig. 12. The contact force magnitude for flexion angle 0°, Q-angle 14°, CTA 0°.

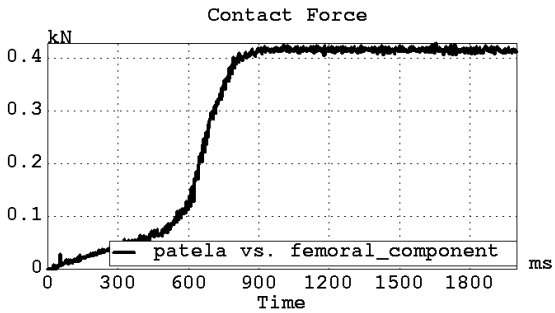

Fig. 13. The contact force magnitude for flexion angle 0°, Q-angle 14°, CTA 5°.

Acknowledgements

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