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The effect of different quadriceps loading patterns on tibiofemoral joint kinematics and patellofemoral contact pressure during simulated partial weight-bearing knee flexion

Markus Wünschel · Ulf Leichtle · Christian Obloh · Nikolaus Wülker · Otto Müller

Abstract

Purpose The purpose of this in vitro study was to investigate the influence of different quadriceps loading patterns on tibiofemoral joint kinematics and patellofemoral pressure.

Methods A dynamic muscle-loaded knee squat was simulated on eight knee specimens with an upright knee simulator while measuring tibiofemoral joint kinematics and patellofemoral pressure distribution. The quadriceps muscle was attached to three actuators simulating the three main extensor muscles, and five different quadriceps loading patterns were tested.

Results Tibial axial and varus-valgus-rotation are affected most while changing quadriceps loading patterns from lateral to medial. Higher internal tibial rotation is associated with higher medial muscle load compared to the symmetrical loading condition. Contact force, contact area and maximum peak pressure rise with increasing flexion angles. Accentuating the vastus lateralis muscle induces a significant reduction in patellofemoral contact force and a 30% diminished contact area at 90° of flexion.

Conclusion Strengthening the vastus medialis muscle leads to increased internal tibial rotation, thus optimizing patella tracking by lowering the Q-angle. In contrast, weakness of the vastus medialis muscle causes decreased tibial internal rotation and is associated with lower patellofemoral contact pressure and contact area. Vastus medialis exercise is advisable to improve patella tracking but may not be recommended in patients with disorders due to increased patellofemoral contact pressure.

Keywords Patella contact pressure · In vitro · Knee kinemator · Vastus medialis · Knee kinematics

Introduction

Disorders of the patellofemoral joint are widespread and have significant influence on patients' ability to perform activities of daily living like walking, kneeling or climbing stairs [20, 29]. The origin of the pain may be easy to find like fractures or osteoarthritic changes whereas in many cases remains open [2–5, 25, 26, 29].

It is well-established knowledge through in vivo and in vitro studies that patellofemoral kinematics and thus contact pressure are influenced by many different aspects [1, 12, 14, 27, 28]. The leg axis and determined by it the Q-angle are important anatomical determinates of patella kinematics [23]. Axis-deviations in varus or, clinically more relevant valgus (thus increasing the Q-angle), can cause major problems with recurrent instability and luxation of the patella [7].

Another very important factor of patellofemoral contact pressure and kinematics is the quadriceps muscle which can be separated into 3 functional units: vastus lateralis, vastus intermedius + rectus femoris, vastus medialis [18, 30]. Many studies have focused on the vastus medialis obliquus muscle [5, 8, 11, 30], and it was stated that patellofemoral malalignment is one of the main reasons for patellofemoral pain leading to a vast variety of different surgical procedures to correct this ostensibly cause of the pain [14, 15, 19, 21, 25].

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The purpose of this in vitro study was to investigate the influence of different quadriceps loading patterns on tibio-femoral joint kinematics and patellofemoral pressure in order to recommend treatment options concerning quadriceps strengthening. Our null-hypothesis was different loading patterns of the quadriceps muscle have no influence on tibiofemoral kinematics and patellofemoral contact pressure.

Materials and methods

A dynamic muscle-loaded knee squat was simulated on eight fresh-frozen knee specimens with an upright knee simulator (Fig. 1) in a closed kinetic chain experiment. During the squat, the patellofemoral pressure distribution was measured with a flexible resistive force sensor.

Specimens

Eight fresh-frozen human cadaveric knee specimens (4 male/4 female) with an average age at the time of death of
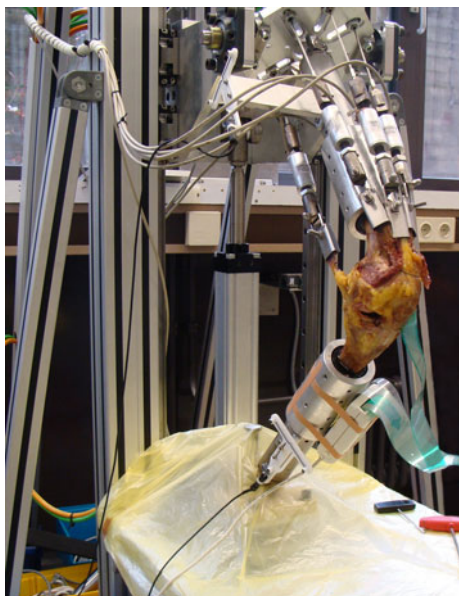

Fig. 1 Knee simulator with a mounted knee specimen in an intermediate position while simulating a muscle-loaded knee flexion. In order to allow unconstrained movement in six degrees of freedom (DOF), the hip assembly allows knee flexion and abduction/adduction and is vertically driven by an electronic servo motor. The ankle bearing allows for tibia axial rotation, varus/valgus rotation as well as tibia flexion/extension. The ankle assembly is fixed to the bottom frame of the simulator including a vertical force transducer with a medial offset to account for natural valgus orientation of the knee joint of about 7°. Muscle forces are applied by steel wires attached to the muscle tendons. Ultrasonic sensor triads aligned to the segment shaft axes record the motion of the femur and tibia segments. A flexible pressure sensitive sensor foil is inserted between femur and patella and fixed by three small sutural incisions

75.4 ± 10.4 years (mean $\pm$ standard deviation) were studied. All knees were examined macroscopically and radiographically to exclude specimens with degenerative or traumatic changes. The femur and tibia were cut 150 mm from the joint line, and the five muscle tendons (i.e. musculus vastus lateralis (VL), vastus medialis (VM) and rectus femoris (RF) as well as the medial and lateral hamstring muscles namely biceps femoris and semitendinosus) have been exposed while keeping the cruciate and collateral ligaments intact. All other skin and soft tissues were removed. The fibula was secured to the tibia with cortical screws to prevent its motion during the test. The femur and tibia were mounted onto a thick-walled steel cylinder using a bone cement compound (PMMA: Technovit 2060, Heraeus Kulzer, Hanau, Germany) and multiple accurately positioned set screws (Fig. 1).

Knee simulator

While moving the knee from near full extension to 90° of flexion, the knee simulator is able to apply variable muscle forces at the five aforementioned major muscles around the knee joint in order to maintain a preset value for the vertical force at its ankle bearing (ankle force, AF). AF can be regarded as the amount of simulated body weight (BW) or ground reaction force (GRF), respectively. The technical details and the validation of the simulator have been described previously [22]. To preserve the specimens for the different measuring conditions, we used a reduced value of simulated BW (AF = 50 N) for all trials which required a total quadriceps force of at least 600 N near 90° of flexion. The hamstrings forces have been kept constant at 20 N just to tighten them slightly. To prevent the specimens from possible hyperextension, we started the knee movement at 10° of flexion. The simulator then moves the knee with a nearly constant flexion speed of 1° per second to the final position of 90°, preset by the custom made application software (developed in LabVIEW, National Instruments, Austin, TE) which also continuously controls the muscle forces in order to reach the preset amount of AF. The actual force values of the muscles are recorded with the sampling rate of the control loop of the application (i.e. approximately 300 ms).

Motion capturing system: calculation of knee kinematics

While flexing, the knee ultrasonic sensor triads of a motion capturing system (CMS-H, Zebris, Isny, Germany) record the motion of the femur and tibia segments with a sampling rate of 1 Hz. With the help of an initial reference measurement of the knee joint at full extension, a segment-based moving coordinate system (CS) was calculated from

the recorded tracking data for the femur and tibia respectively. In full extension, both CS are located in the midpoint of two reference points which are palpated with a stylus pointer as the most prominent medial and lateral points on the tibia plateau. After the squat was conducted, six measurement values for the kinematic description of the knee were calculated for each time step in terms of three translation values between the two moving CS (anterior/posterior (AP), medial/lateral (ML) and cranial/caudal) and three Euler angles representing the orientation of the tibia CS with respect to the femur CS (flexion, varus/valgus rotation, and axial internal/external rotation). The flexion axis points along the medial/lateral direction which is defined by the two reference points. The flexion axis and the segment shaft axis define a plane perpendicular to the varus/valgus (VV) rotational axis (AP-direction). The axial rotation axis is mutually orthogonal to the flexion axis and the AP-axis. All calculations were performed with custom made software written in MatLab (The MathWorks, Natick, MA).

Patellofemoral pressure measurement

A flexible pressure sensitive sensor foil (K-Scan #4000, TekScan, Boston, MA) with a measuring range up to 62 MPa was inserted between femur and patella and fixed by three small sutures. Therefore, the knee joint was opened through a lateral parapatellar incision of about 35 mm along the joint space (Figs. 1, 2). The pressure foils were calibrated with a material testing machine (858 MiniBionix II, MTS, Eden Prairie, MN) before the specimens were measured. While moving the knee, one pressure frame per second (i.e. a sensor matrix of 572 sensels per
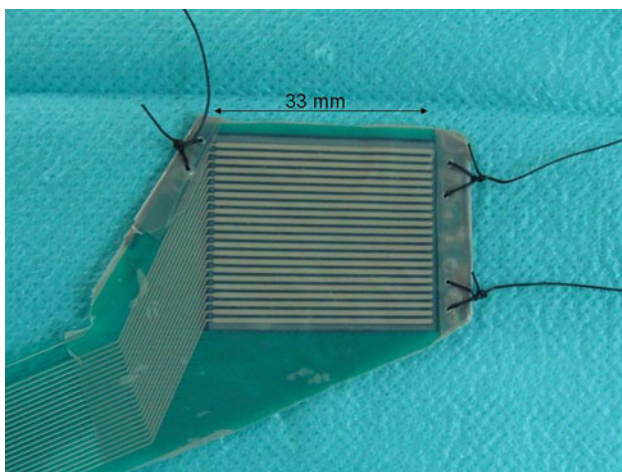

Fig. 2 Pressure sensitive foil with sutures. The matrix based foil (thickness of film 0.1 mm) consists of 22 × 26 sensor elements (sensels) with a rectangular spacing of 1.3 mm leading to a size of the foil of 28 mm times 33 mm and a pressure range from 0 to 62 MPa

second covering an active sensing area of 9.24 cm^2) was captured by the corresponding application software (IScan 5.83). With software routines written in MatLab, the pressure data have been further analysed. Values for the patellofemoral contact force, and area of loaded sensors, maximal peak pressure (MPP) and centre of pressure (COP) were determined for each frame. Figure 3 displays data for each muscle loading condition for one typical specimen. For every loading condition, an overall MPP image including the course of the COP for all flexion angles is shown. The diagram illustrates representative for all our measurements that the COP moves from distal to proximal along the retropatellar surface during knee flexion. The highest contact pressure values are observed at the highest flexion angles. The MPP shifts according to the different loading conditions.

Experimental protocol

For each specimen, a reference measurement was done at full extension without any muscle load in order to get the above-mentioned two reference points of the knee joint. When starting the muscle-loaded motion from full extension, the specimen could move into hyperextension and be destroyed. Therefore, the specimens were driven to a save starting position of about 10° of flexion. Each specimen was then moved to 90° of flexion in an unloaded muscle condition. The first muscle loading condition we applied were equal muscle forces for all three extensor muscles (sym). In the second loading condition, we applied a slightly lateral loading distribution weighting the muscle force values for VM, RF and VL with 17, 33 and 50%, respectively (lateral 1). Extreme lateral muscle load was realized by the third condition with the weighting factors 0/33/67 (lateral 2). Corresponding values have been chosen in order to simulate asymmetrical muscle loading conditions medially (50/33/17 and 67/33/0; medial 1 and medial 2). Test–retest repeatability of the kinematic variables was <0.5°/1 mm for rotation and translation, respectively [22]. Therefore, each loading condition was run twice. Doing so, repeatability of COP-motion was <0.5 mm in either direction. Reproducibility was <0.1 MPa for MPP-values, 5 N for the patellofemoral force and 5 mm^2 for the patellofemoral contact area.

Statistical analysis

From all calculated measuring values, the knee flexion angle was selected to be the independent variable. Before analysing any data, the flexion values and all other data were interpolated in order to get equidistant flexion steps of exactly 1°. At each flexion step, we obtained mean values for two successive flexion runs for each specimen and each
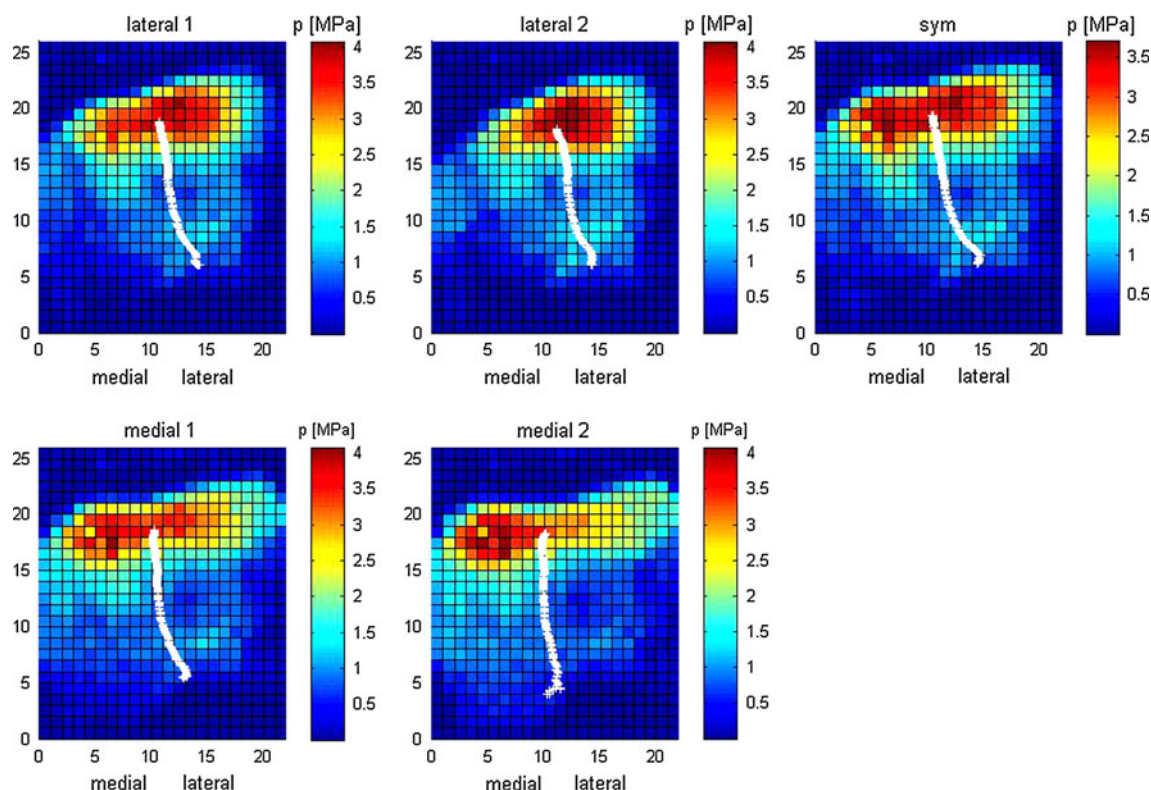
Fig. 3 Summarized pressure image of the sensor foil for one representative specimen and 5 loading conditions (lateral 1 and 2; medial 1 and 2; sym). Colour indicates maximum peak pressure (*MPP*) value for each sensel during one flexion run. Pressure values are scaled in the adjacent *colour bars*. The image corresponds to a frontal view onto the patella. Furthermore, the course of centre of pressure is shown in steps of 1°. The point farthest down marks the start of the knee flexion. The course continues from bottom up until 90° of flexion is achieved

loading condition. In order to compare the influence of different muscle conditions during knee flexion, we calculated sample mean values plus the standard error of the mean (STE) for all dependent measuring values at each flexion step. We tested our hypothesis that different loading patterns of the quadriceps muscle have no influence on tibiofemoral kinematics and patellofemoral contact pressure by repeated measurement ANOVA (anova-rm, MatLab File Exchange) at flexion steps of 5° (significance level: $P < 0.05$).

Results

Knee kinematics

Tibia axial rotation and varus-valgus rotation are affected most while changing quadriceps muscle loading patterns from lateral to medial. In any loading condition, the tibia internally rotates about 6–8° during flexion reaching plateau values beyond 70° (Fig. 4a–d). Higher medial muscle load (medial 1 + 2) is associated with higher internal tibial rotation compared to the symmetrical loading condition, whereas varus rotation is reduced slightly but significantly

(Fig. 4a, b). Conversely, lateral muscle loading (lateral 1 + 2) produces less tibia internal rotation beyond 70° of flexion and furthermore augments tibia varus rotation beyond knee flexion values of 35° (Fig. 4c, d).

Patellofemoral joint

Contact force, contact area and MPP continuously rise with increasing flexion angles. The COP in medial/lateral direction varies approximately 2–6 mm. In axial direction, the COP moves from the lower part of the patella towards the upper part during flexion (Figs. 5, 6).

Patellofemoral contact force and contact area:

In small flexion angles, an increased vastus medialis force does not alter the patellofemoral contact force or loaded area (Fig. 5). From 65 up to 90° of flexion, however, patellofemoral force as well as patellofemoral contact area (except at 90°) increased significantly with respect to the balanced loading situation.

In contrast, strengthening vastus lateralis induces a significant reduction in patellofemoral contact force

Fig. 4 Course of tibia internal/external rotation and varus/valgus rotation during knee flexion for lateral or medial muscle loading condition respectively, each compared with the symmetrical case. *Upper two diagrams* Tibia internal rotation (**a**) and tibia varus/valgus rotation (**b**) for symmetrical and medial muscle loading (sym, medial 1 and medial 2). *Lower two diagrams* Tibia internal rotation (**c**) and tibia varus/valgus rotation (**d**) for symmetrical and lateral muscle loading (sym, lateral 1 and lateral 2). The *asterisks* indicate statistical significant differences of the sample means according to repeated measurement ANOVA ($P < 0.05$), whereas the *error bars* indicate half of the standard error of the mean (*STE*) of each sample mean
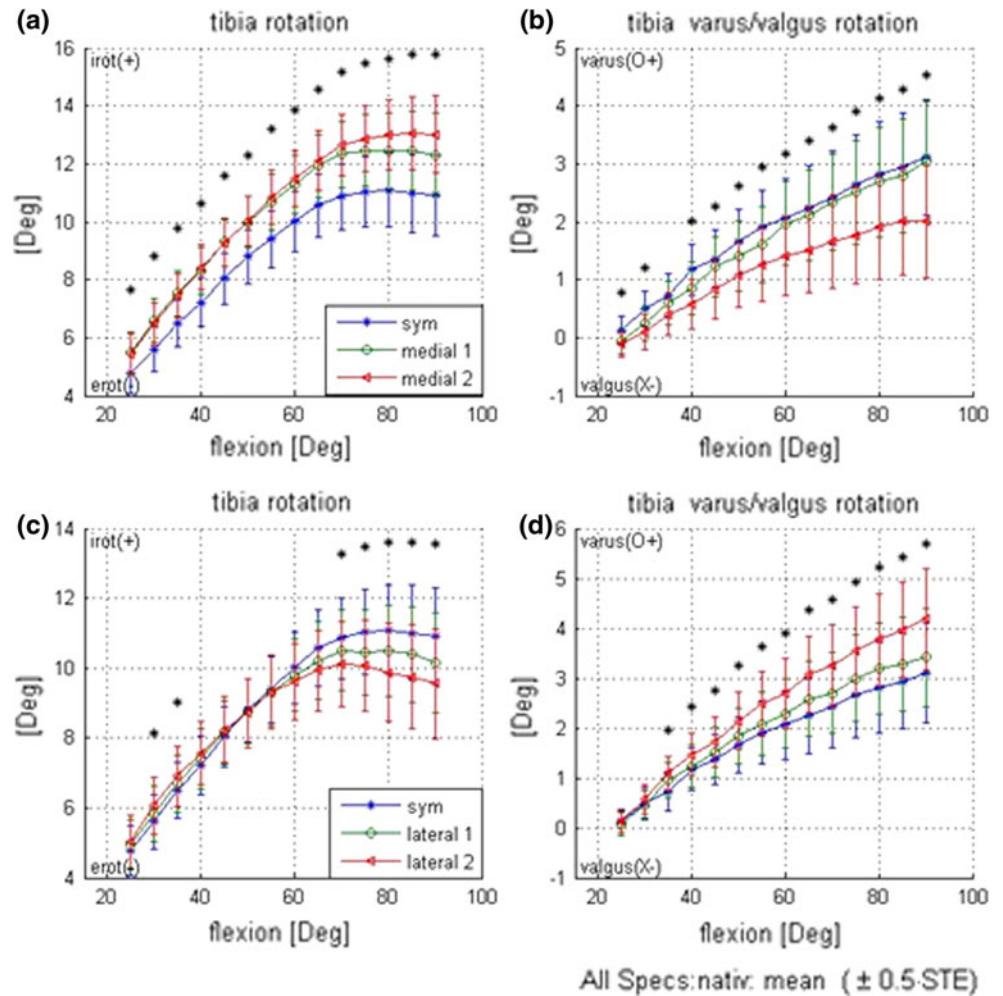
(Fig. 6) and a 30% diminished contact area, at 90° of flexion.

Maximum peak pressure and centre of pressure

There is only little influence on MPP by variation of loading among the attached muscles (Figs. 5, 6). The COP does not change significantly concerning medial/lateral position of the sensor foil for the medial 1 + 2 trials (Fig. 5), whereas more vastus lateralis force is able to move the COP significantly lateral from 25 to 90° of flexion (Fig. 6).

Discussion

The most important finding of the present study was that the vastus medialis muscle influences patella alignment by two different ways. Weakening the vastus lateralis muscle (thereby emphasizing vastus medialis) leads to a markedly increased internal tibial rotation. We figure that this is due to the different anatomical attachments at the patella of

these two muscles and accentuates the well-known importance of the vastus medialis muscle not only to patellofemoral kinematics but also to tibiofemoral kinematics [27]. Moreover, with the double sided effect of directly medialising the patella plus medialising the tuberositas tibiae by increasing tibial internal rotation thus lowering the distance between the tuberositas tibiae and the trochlea groove (TT-TG distance), this muscle proves to be the important stabilizer of the patella [19].

The medial/lateral position of the COP is significantly influenced, whereas the cranial/caudal position was hardly affected by different muscle loading patterns. This corroborates other studies [14]. We contribute this to the anatomy of the patellofemoral joint. When emphasizing the vastus medialis muscle (medial 1 + 2, Fig. 5), the COP is not changed ($P > 0,05$), whereas an increased lateral pull changed the COP significantly almost throughout the whole flexion range (lateral 1 + 2, Fig. 6), a result which confirms the computer simulation of Flatow et al. [10] who had similar results. Since the medial condyle usually is lower and shorter than the lateral condyle [27], the patella may be able to move medially [1] without changing the COP. In
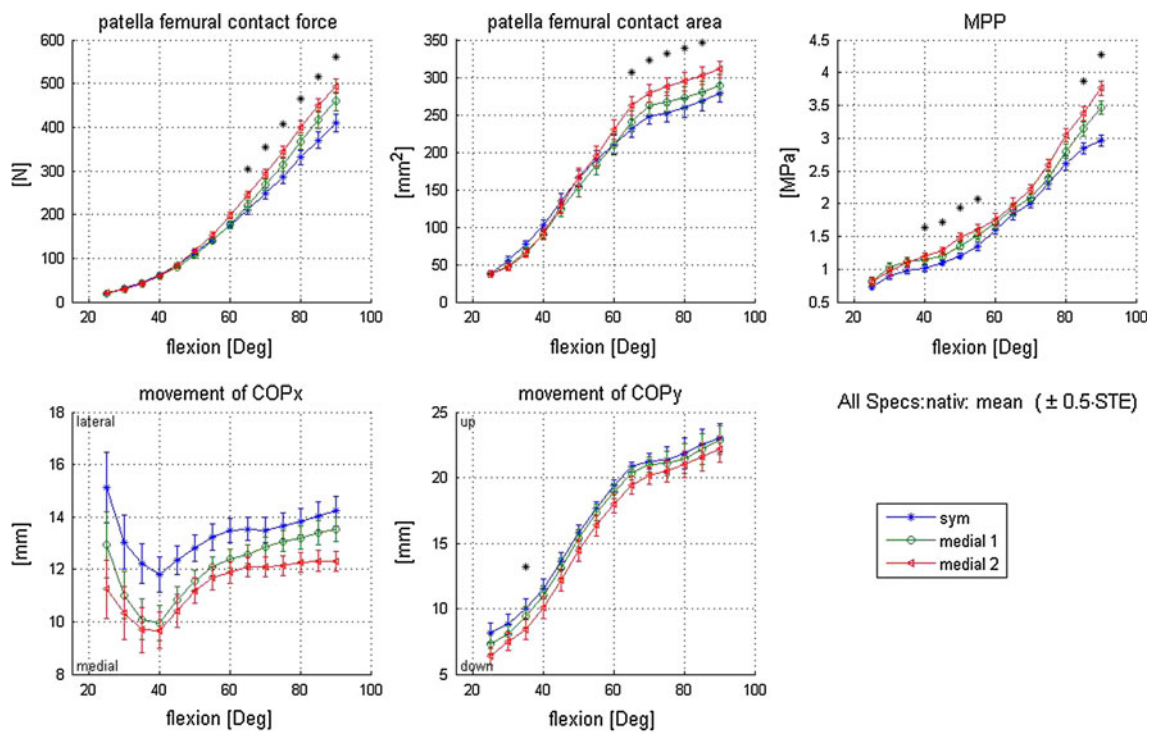
Fig. 5 Alteration of pressure measurement values due to medial muscle loading (sym, medial 1, medial 2). The *asterisks* indicate statistical significant differences of the sample means according to repeated measurement ANOVA ($P < 0.05$), whereas the *error bars* indicate half of the standard error of the mean (*STE*) of each sample mean

Fig. 6 Alteration of pressure measurement values due to lateral muscle loading (sym, lateral 1, lateral 2). The *asterisks* indicate statistical significant differences of the sample means according to repeated measurement ANOVA ($P < 0.05$), whereas the error bars indicate half of the standard error of the mean (*STE*) of each sample mean
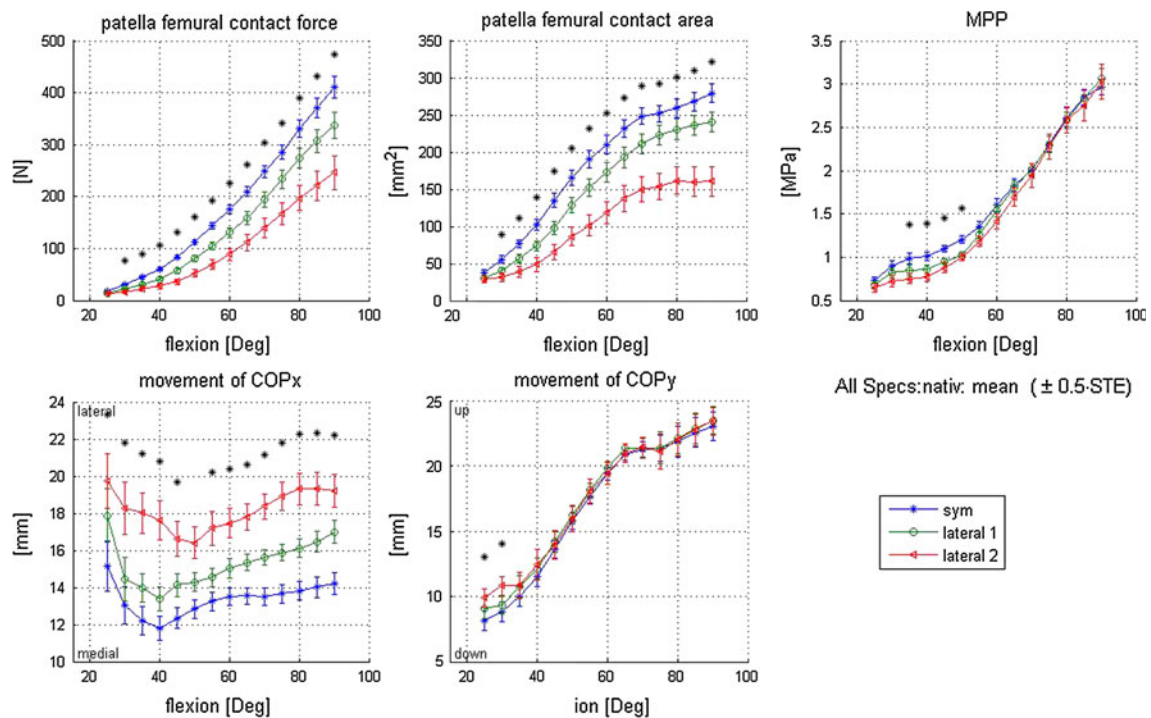
the other state, the lateral condyle hinders patella movement more, but due to the lateralized force vectors the COP noticeably changes.

The findings concerning patellofemoral contact pressure are in concordance with several other studies [6, 12, 24] showing that the patellofemoral contact force and the patellofemoral contact area are rising with flexion angles due to changes in the lever arm during flexion. Interestingly our data suggest that weakness of the vastus medialis muscle leads to lower contact forces with a decreased contact area at the same time. This seems to be a contradiction at first and is in conflict to other studies [16, 17, 21] since one would expect higher contact forces if the contact area lessens and the simulated bodyweight remains constant. We contribute this to the different anatomy of the medial and the lateral condyle. Since the lateral condyle is slightly higher and reaches more anterior, the lateral pull of the quadriceps might lead to a slight anterior shift of the patella, leading to a more efficient lever arm and thus making it possible to simulate the same bodyweight (AF) with lower patellofemoral contact force and contact area. MacIntyre confirmed an association between higher anterior patella translation and lateral translation of the patella in an in vivo MRI-study [20]. We conclude that the adapted anatomy of the distal femur and the condyles compensate the physiological valgus of the leg axis. Kuroda et al. [14] found increased patellofemoral contact pressure after a medializing tibial tuberosity transfer, a procedure leading to a better bony alignment of the patella while on the same time weakening the effect the vastus medialis muscle thus confirming our results.

One of the limitations in the current study is that only a portion of the body weight and 10 N hamstring force were simulated. This set-up certainly is not comparable to real physiological mechanical conditions in vivo yet was inevitable because the quadriceps tendons in some of the specimens could not sustain the corresponding muscle forces resulting from higher body weight simulation. Previous investigation [22] has shown that the change in the knee kinematics profile is not sensitive to the increasing simulated body weight. Therefore, we believe that qualitative clinical insights can still be elucidated with the partially loaded knee. However, the results cannot definitely be extrapolated to predict the knee kinematics and kinetics under full body weight, especially taking into account that in a clinical situation one will not start from the ideal position but a lateralized or tilted patella. The current set-up does not reflect the pathological condition in the patellofemoral joint in vivo. A second limitation in this study is that the knee flexion only started from 10°, because the joint could not be flexed from fully extended position by only applying a PBW and quadriceps force. Since we drive the kinemator fully force-controlled and full

extension is an unstable state for the force-controlled movement, the specimen might hyperextend and thus be destroyed if the extension angles reach below 10°. The information within first 10° of flexion is thus missing. A third limitation is the hamstring forces were kept constant during the simulated flexion. This is mainly because the multiple agonist and antagonist muscle forces composed a mechanically indeterminant system, and it is not likely to determine their individual contributions. Nonetheless, in EMG-studies, it has been shown that estimated hamstring forces remain relatively constant and low in a squatting exercise [9, 13].

Another shortcoming is the attachment of the quadriceps-actuators. Since we have only three actuators, we could not attach the 6 different muscles of the quadriceps but had to simplify and reduce it to the three main muscle groups mentioned earlier.

Conclusion

Strengthening the vastus medialis muscle leads to a medialisation of the patella as well as to an increased internal tibial rotation, thus optimizing patella tracking by lowering the Q-angle. In contrast, weakness of the vastus medialis muscle causes decreased tibial internal rotation and is associated with lower patellofemoral contact pressure and contact area. It is concluded that vastus medialis exercise is advisable to improve patella tracking but may not be recommended in patients with disorders due to increased patellofemoral contact pressure. The results of the present study may help to improve rehabilitation of patients suffering from disorders of the patellofemoral joint like anterior knee pain.

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