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Increased internal femoral torsion can be regarded as a risk factor for patellar instability — A biomechanical study

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ABSTRACT

Background: Increased internal femoral torsion is regarded as a risk factor for patellar instability. Biomechanical investigations confirming this hypothesis are missing.
Methods: Eight fresh-frozen cadaver knees were tested on a specially designed simulator. Patellar motion and patellofemoral pressure were evaluated for 0°, 10°, and 20° of increased internal and external femoral torsion with native and with transected medial patellofemoral ligaments used to simulate patellar instability. A regression analysis was used for statistical analysis.
Findings: In native medial patellofemoral ligaments, there were no significant changes in mean or peak pressures for any torsional states ($P \geq 0.07$). At 20° increased internal femoral torsion, there was a significant center of force shift towards the lateral side ($P = 0.01$). Patellar shift was directed laterally at low knee flexion angles up to 30°. Lateral patellar tilt increased significantly at 10° and 20° of increased internal femoral torsion ($P \leq 0.004$). In transected medial patellofemoral ligaments, mean pressure ($P \leq 0.005$) and peak pressure ($P \leq 0.02$) decreased significantly for all torsional states. There was a significantly greater lateral center of force shift with increased internal femoral torsion ($P \leq 0.04$). Lateral patellar tilt increased significantly ($P < 0.001$). Patellar shift did not change significantly with increased internal femoral torsion ($P \geq 0.30$).
Interpretation: In a native medial patellofemoral ligament, 20° of increased internal femoral torsion can be regarded as a significant risk factor for patellar instability. With an insufficient medial patellofemoral ligament, 10° of increased internal femoral torsion already represents a significant risk factor.

1. Introduction

Patellar instability is a common problem of the patellofemoral joint. One form of patellar instability among others applies to patients suffering from patellar dislocation. Other forms include patellofemoral pain and patellofemoral pain with concomitant anatomical anomalies. The recurrence rate for dislocations ranges from 2.4% to 49%, depending on the treatment procedure used, the number of previous dislocations, and predisposing factors such as soft-tissue and bone abnormalities (Fithian et al., 2004; Maenpaa and Lehto, 1995, 1997; Maenpaa et al., 1997).

Soft-tissue risk factors that may be involved include ligamentous laxity or an insufficient function of the medial patellofemoral ligament (MPFL). The MPFL is thought to be the major restraining ligament that counteracts the lateralizing vector of the quadriceps and thus provides protection against lateral patellar instability (Conlan et al., 1993; Hautamaa et al., 1998). Patients with recurrent dislocation thus have an inadequate medial restraint (Nomura, 1999; Schmeling, 2010), as the MPFL ruptures in up to 94% of all patients with first-time patellar dislocations (Sallay et al., 1996).

Osseous abnormalities include trochlear dysplasia, valgus deformity, and torsional deformities such as increased external tibial or increased internal femoral torsion — all related to an increased tibial tuberosity–trochlear groove (TT–TG) distance (Petersen et al., 2012; Redziniak et al., 2009; Schmeling, 2010; Strobl and Grill, 1998). Increased internal femoral torsion is therefore regarded as being a risk factor for patellar instability (Biedert, 2008; Dejour and Le Coultre, 2007; Dickschas et al., 2014; Hinterwimmer et al., 2012; Post et al., 2002; Teitge, 2006). The mean value for physiological femoral torsion ranges from 7° to 24° of internal torsion, depending on the

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measurement technique used (Koerner et al., 2013; Kuo et al., 2003; Strecker et al., 1997). Some authors have proposed performing derotational osteotomy in patients with recurrent patellar dislocations in cases of concomitant absolute values of 15°–25° of internal femoral torsion in order to prevent instability events and recurrent dislocations (Ateschrang et al., 2014; Biedert, 2008; Hinterwimmer et al., 2012; Teitge, 2006).

The influence of femoral torsion on patellofemoral pressures has been evaluated in the literature (Besier et al., 2008; Lee et al., 1994). However, to the best of the present authors' knowledge, no biomechanical studies have yet been conducted in order to investigate the effects of femoral torsion on the patellofemoral joint in relation to patellar instability. It remains unclear whether increased femoral torsion facilitates patellar instability. As it is unclear whether femoral torsion can be regarded as a risk factor for patellar instability, potentially resulting surgical treatment remains questionable.

Therefore, the purpose of this study was to assess the effects of femoral torsion on patellofemoral tracking and retropatellar pressure in cadaver knees either with native patellofemoral soft-tissue anatomy or with transected MPFL simulating conditions in a patient with recurrent patellar instability events. The results of the present biomechanical study contribute to the understanding of coherences between femoral torsion and patellar instability and dislocations. A more profound knowledge of these coherences might point towards new treatment options in the future. It was hypothesized that an increase in internal femoral torsion increases lateralizing effects on the patella such as tilt, shift, center of force shift and increased patellofemoral peak pressure.

2. Methods

Eight fresh-frozen knees (from donors aged 52–88 years; eight women) were used for testing. The bodies were donated by individuals who had provided informed consent before death to the use of their bodies for scientific and educational purposes (McHanwell et al., 2008; Riederer et al., 2012). Quantitative computed tomography scanning (qCT; Lightspeed VCT 64, General Electrics, Milwaukee, Wisconsin, USA) was carried out on each specimen in order to exclude knees with major malformations such as trochlear dysplasia (sulcus angle > 145°), bipartite patella, major arthrosis, or previous surgery.

The femur was cut 150 mm proximal to the knee joint line and the tibia was cut 115 mm distal to it. The fibula was removed underneath the fibular head. The quadriceps muscle and all ligaments and retinacula were left intact, and the capsule was incised proximally in order to apply the pressure measurement system.

The knees were mounted on a specially designed knee simulator (Fig. 1). The simulator consisted of a continuous passive motion (CPM) machine (Artromot K4, Ormed Ltd., Freiburg, Germany) with a self-constructed tibial and femoral mounting. The CPM machine performed knee flexions and extensions at between 0° and 90°. The tibia was embedded in polymethylmethacrylate (PMMA; Technovit 3040, Heraeus Kulzer Ltd., Wehrheim, Germany) to prevent translation and torsion inside the mounting. During flexion and extension, tibial mounting allowed free translation in the ventrodorsal and proximodistal directions. An intramedullary rod was inserted inside the femur and locked with three distal locking screws under fluoroscopic guidance. Femoral mounting allowed adjustable internal and external torsional states of the femur. By the use of a levelling tool it was ensured that the posterior condyles were horizontally embedded. The transepicondylar axis was regarded as the flexion–extension axis and was aligned with the pivot of the CPM machine in the unrotated state and remained similar for the simulation of femoral torsion.

Muscle loads were applied to the quadriceps muscle, which was categorized into three parts (the rectus femoris and vastus intermedius, vastus lateralis, and vastus medialis). Each muscle part was sutured with FiberWire (Arthrex Inc., Naples, Florida, USA) transverse to the muscle fibers in order to prevent muscle rupture through the clamp.

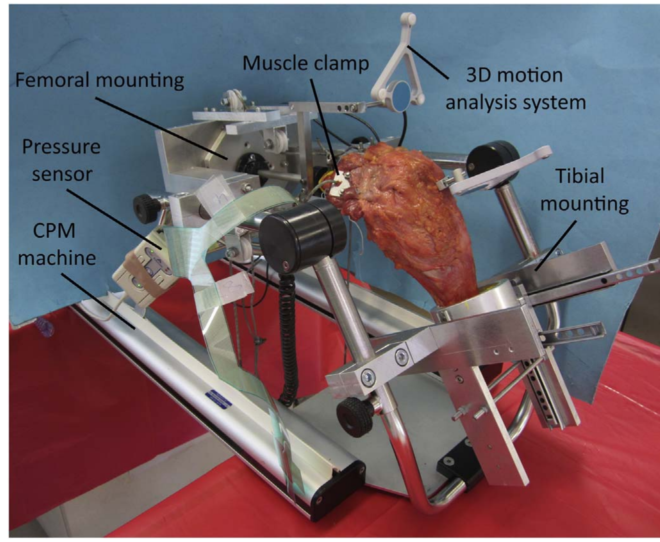

Fig. 1. The knee testing simulator. A knee specimen has been mounted here, with the motion and pressure measurement devices. CPM, continuous passive motion.

The direction of muscle force was calculated relative to the reported force vectors and cross-sectional areas, simplifying three out of five reported quadriceps parts (Farahmand et al., 1998). The rectus femoris and vastus intermedius muscle pulled in a 0° direction in the frontal plane and 0° direction in the sagittal plane. The vastus lateralis muscle force vector had a 19° lateral pull in the frontal plane and a 7° posterior pull in the sagittal plane. The vastus medialis muscle force vector had a 27° medial pull in the frontal plane and 17° posterior pull in the sagittal plane.

The muscle load was distributed equally to each of the three parts of the simplified quadriceps muscle system. Depending on the knee flexion angle, the total quadriceps load was 3 × 15 N (45 N) at 0° of knee flexion, 3 × 19 N (57 N) at 15°, 3 × 38 N (114 N) at 30°, 3 × 57 (171 N) at 45°, 3 × 87,5 N (262,5 N) at 60°, 3 × 125 N (375 N) at 75°, and 3 × 160 N (480 N) at 90° of knee flexion (Muller et al., 2009). The muscle load was applied manually using dead weights via muscle clamps. The knee specimen was flexed from 0° (knee extension) automatically by the CPM machine and stopped automatically by the CPM machine at each knee flexion angle of 15°, 30°, 45°, 60°, 75°, and 90°.

Patellar motion (patellar shift and patellar tilt) was recorded using an ultrasound-based three-dimensional motion analysis system (CMS70P8, Win-BioMechanics version 1.2; Zebris Medical Ltd., Isny, Germany). Patellar motion was recorded within a coordinate system simulated around the femur. The coordinate system therefore rotated simultaneously with femoral torsion. Patellar position was evaluated in regard to the femur. Comparison of patelar motion was conducted in regard to the neutral patellar position.

Retropatellar pressure (mean pressure, peak pressure and center of force shift [the shift of the position of the center of force]) was measured with a thin tactile dynamic pressure sensor (Pressure Mapping Sensor 4000, Tekscan Inc., South Boston, Massachusetts, USA), using a newly equilibrated and calibrated sensor for each specimen. Equilibration and calibration was conducted according to the manufacturer's handbook. The retropatellar sensor was inserted via a suprapatellar incision and fixed to the patellar capsule with stitches. This type of fixation has proved to be the most stable method of fixing the sensors in place, preventing movement during testing and prolonging durability (Wilharm et al., 2013).

For the experimental design, the posterior condyles were positioned horizontally, representing 0° of proximal component torsion, which was defined as the mean femoral torsion in the population. All torsional values have to be regarded as 10° and 20° of increased internal torsion (IT) and external torsion (ET) apart from the normal value. The knee