Tibiofemoral contact mechanics with a femoral resurfacing prosthesis and a non-functional meniscus

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Abstract

Background: Increased contact stress with a femoral resurfacing prosthesis implanted in the medial femoral condyle and a non-functional meniscus is of concern for potential deleterious effects on tibiofemoral contact mechanics.

Methods: Peak contact pressures were determined in seven fresh frozen human cadaveric specimens using a pressure sensitive sensor placed in the medial compartment above the menisci. A knee simulator was used to test each knee in static stance positions (5°, 15°, 30°, 45°) and through 10 dynamic knee flexion cycles (5°–45°) with single body weight ground reaction force which was adjusted to the living body weight of the cadaver donor. All specimens were tested in three different conditions: untreated knee (A); flush implantation of a 20 mm resurfacing prosthesis (HemiCAP®) in the weight bearing area of the medial femoral condyle (B); complete radial tear at the posterior horn of the medial meniscus with the femoral resurfacing device in place (C).

Findings: On average, flush device implantation resulted in no statistically significant differences when compared to the untreated normal knee. The meniscal tear resulted in a significant increase of the mean peak contact pressures by 63%, 57%, and 57% (all \( P < 0.05 \)) through the dynamic knee flexion cycle. No significant different maximum peak contact pressures were observed at 5° stance position.

Interpretation: Although the condition of a meniscal tear without the resurfacing device could not be compared, possible effects of reduced meniscal tissue and biomechanical integrity of the meniscus must be considered in an in vivo application.

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1. Introduction

Full thickness articular cartilage defects are frequently diagnosed and often associated with substantial morbidity and functional limitation (Curl et al., 1997). Hjelle et al. (2002) diagnosed focal chondral or osteochondral defects (of any type) in 19% of the patients evaluating one thousand consecutive knee arthroscopies. Location of the cartilage defects were predominantly on the medial femoral condyle and concomitant meniscal injury was found in 42% of these patients. Increased incidence of osteoarthritis is demonstrated with the presence of a significant chondral or osteochondral defect and/or with loss of meniscus tissue by partial or total meniscectomy in numerous studies (Fairbank, 1948; Hede et al., 1992; Higuchi et al., 2000; Linden, 1977; Maletius and Messner, 1996; Rangger et al., 1997; Shelbourne et al., 2003).

The patient over the age of 40 years with a focal full thickness chondral or osteochondral defect reflects a serious problem for the orthopedic surgeon. Conservative treatment at best ameliorates the symptoms. Biological repair techniques, such as autologous chondrocyte transplantation, osteochondral transplantation (OATS, Mosaicplasty) and marrow stimulation techniques have shown promising results in younger patients (Hangody and Fules, 2003; Mithoefer et al., 2005; Nehrer et al., 2006; Peterson et al., 2002) but appear to be increasingly ineffective with increasing age (Kreuz et al., 2006; Mithoefer et al., 2005). Unicompartimental or total knee arthroplasty represent procedures of final resort prompting the likelihood of revision surgery for the middle aged patient with associated morbidity during the patient’s lifetime.

A novel metallic resurfacing prosthesis (HemiCAP®, Arthrosurface Inc., Franklin, MA, USA) was developed as an interim or alternative treatment strategy in patients when only one compartment is affected by posttraumatic, degenerative disease or necrosis associated with large unstable articular defects with significant subchondral bone exposure. However, effects of a metallic implant...
articulating with intact opposing tibial articular cartilage and meniscus remain largely unanswered to date. In a previous biomechanical study, flush implantation did not appear to be a biomechanical disadvantage, whereas elevated implantation of the device has shown significantly increased tibiofemoral peak contact pressures compared to the untreated condition in human cadaver knees with intact chondral surfaces and meniscus (Becher et al., 2008).

The menisci of the knee aid in load bearing (Kurosawa et al., 1980; Seedhom et al., 1974) and joint stability (Sullivan et al., 2008). Load transfer characteristics are altered in injured menisci or after total or partial meniscectomy. An in vitro study in human cadaver knee specimens has demonstrated increased peak contact pressures proportionally to the amount of meniscus removed (Lee et al., 2006). Increased peak contact pressures up to 136% was shown after total medial meniscectomy (Lee et al., 2006). No information exists, however, if loss of meniscus integrity and meniscus function might lead to deleterious effects to the articular cartilage of the tibial plateau with the metallic device implanted in the weight bearing area of the medial femoral condyle.

This in vitro study aims to determine the effect of a complete radial tear on peak contact pressure in the tibiofemoral joint with the resurfacing prosthesis implanted in the medial femoral condyle.

2. Materials and methods

A total of seven fresh frozen knee cadaver specimens (3 pairs, 1 single) were used for data collection in this study. The specimens were obtained from donors, who consented in writing during their lifetime to the use of their body for research and education. The average age of the seven male specimen was 69 years (range: 61–78) with an average weight of 72 kg (range: 61–85 kg). Specimens were selected after inspection of the medial compartment according to the following criteria: intact tibiofemoral cartilage, intact meniscus, and intact collateral and cruciate ligaments. A total of 24 knees were screened resulting in seven knees being appropriate according to the inclusion criteria.

A specially designed knee simulator was used for this study (Fig. 1) (von Skrbensky and Huber, 2006). Similar to in vivo conditions, the main system composed of artificial muscle, force transducer sensor, the joint angle detection and the ground reacting force form a closed loop. The ground reaction force is adjustable according to the donor’s weight. The knee simulator consists of a loading frame (MTS® 858 Eden Prairie, St. Paul, USA) with a long stroke main actuator driven by a hydraulic pump unit (MTS 505.11 silent flow) to simulate body weight and the vertical hip displacement in the mechanical axis of the lower limb. Ankle joint simulation is performed with a hinge joint with one free motion axis. The possible rotation during the movement occurs in the artificial hip joint as if standing with fixed shoe contact. A load transducer is fixed between the mounting plate and ankle joint to detect the vertical ground reacting force (U3 load cell, Hottinger-Baldwin, Darmstadt, Germany). Two smaller actuators apply loads which simulate the quadriceps force. The tendons of the quadriceps muscle are attached to customized curved cryoclamps which avoid patella tilting. A cooling liquid is flowing through the cryoclamps (Haake Synth 60, Karlsruhe, Germany) at a steady temperature of minus 24°C (248 K) (Haake GH, temperature controller D8, Karlsruhe, Germany). These cryoclamps are connected to a waterproof force transducer (SSM-AJ 500, Interface™, Scottsdale, Arizona, USA) and connected to an artificial muscle (Fluidic muscle MAS, Festo®, Esslingen, Germany). The MAS consists of an inner chloroprene tube covered with an aramid fibre shell in a helical mesh. Pressurization of the chloroprene tube results in a longitudinal shortening of the device of up to 25% by increasing the diameter. The mathematical model shows, that the properties of this fluidic muscle is comparable with a skeletal muscle (Tondu and Lopez, 2000). Two MAS40 (40 mm diameter) fluidic muscles are simulating the quadriceps muscle. Two proportional valves (MPPS ¼ 6-010, Festo, Esslingen, Germany) control the air flow to the muscles, provided by a 0.6 MPa medical air pressure line. The proportional valve is integrated in a closed loop with the force transducers. The amplified signal from the ground reaction force transducer is the reference input for the MPT Controller (MTS Multi Purpose Test Star™, Eden Prairie, St. Paul, MN, USA) which calcu-

![Fig. 1. Schematic presentation of the knee simulator. The hydraulic actuator is displacement controlled and connected with a force transducer (1). The default signal is equal to the hip joint (2) movement. The specimen (7) is mounted in aluminium cups (3) using cerro bend alloy. Fluidic muscles (4) simulate the quadriceps and hamstrings. A mounting plate connects the muscle configuration with the force transducer (5) and is connected to the tendons of specimen by two cryoclamps (6). A load transducer (9) is fixed between the mounting plate and ankle joint (8) to detect the vertical ground reacting force.](image-url)
lates the analog output to the proportional valves. These valves apply the required air pressure to counteract the main load frame rod simulating the desired ground reaction forces at a correlated knee angle. The knee angle is adjusted by the displacement controlled main rod. The sensor data from the ground transducer and the two tendon force transducers, as well as the load frame rod position e.g. the flexion of the hip joint and the bending angle of the knee are further inputs of the Test star II™ (MTS Eden Prairie, St. Paul, USA) controller.

The knees were transected 25 cm proximal and distal to the knee joint line. Skin and subcutaneous tissue and the articular capsule, ligaments, and tendons were preserved. The bony ends were freed from all soft tissue for fixation in the aluminium cups. The free end of the bone was potted into a 5 cm diameter and 7 cm tall steel cup using a low melting alloy (Cerro Bend Alloy 158F, Cerro Alloys Co., Pittsburgh, PA, USA). A 7–8 cm longitudinal incision was made over the medial compartment with a curve approximately 1 cm distal to the tibial plateau to gain access for sensor placement. A 0.1 mm thin electronic pressure measuring sensor (K-scan 4000, Tekscan, Boston, MA, USA) was placed in the medial compartment above the menisci and fixed to the periosteum and soft tissue of the tibia anteriorly and posteriorly. For the posterior fixation, a 3–4 cm horizontal incision was made to grasp and divert the posterior sensor Trimmtable Tab with a clamp. The Trimmtable Tabs for attachment were enhanced with a piece of Teflon foil and small holes for the sutures were prepared using a soldering iron. Suturing of the sensor was performed under 850 N static loading at 50° flexion. This method of fixation allowed some sensor movement during flexion during the test cycles to follow the natural movement of the meniscus and prevented tearing and displacement of the sensor from its fixation. To allow comparison of the different testing conditions, only one particular sensor was used for every respective specimen. Before applying the sensor in a new testing condition, the sensor was cleaned after each use by wiping it down with a damp rag or cloth with some alcohol. The sensor consists of load-sensing regions oriented in a grid with 1.27 mm spacing between rows and columns. Each region is referred to as a sensel with piezo-resistive pigments to determine the total compressive load within that region. Size of the sensor is 28 × 33 mm with 62 sensels/cm². The K-scan sensor was successfully used in comparable applications by several authors (Lee et al., 2006; Matsuda et al., 1998; von Lewinski et al., 2006; Wirz et al., 2002).

The HemiCAP™ implant (Arthrosurface Inc., Franklin, MA, USA) is a contoured articular prosthetic (CAP) for focal femoral resurfacing and consists of two components, a fixation component and an articular component, that mate together via a taper interlock to provide stable and immobile fixation of the implant and stress bearing contact at the bone/prosthetic interface (Fig. 2). The fixation component is a modified titanium cancellous screw with a tapered lock. After testing with flush implantation (Fig. 3), the sensor was released from its posterior fixation and a piece of Teflon foil inserted between the sensor and the posterior horn of the medial meniscus to protect the sensor (Fig. 3C). All specimens were preconditioned individually performing five knee bending cycles from 0° to 45° flexion and calibrated in the joint with a two-point calibration method at 700 N and 1500 N according to manufacturer guidelines. Definitions of the correct angles of the actual knee position were adjusted with a custom-made goniometer and verified by the displacement controlled main rod. During the test cycles the specimens were sprayed with saline solution to prevent dehydration.

All specimens were tested in three different conditions (Fig. 3): untreated knee (A); flush implantation of a 20 mm resurfacing prosthesis (HemiCAP™) in the weight bearing area of the medial femoral condyle (B); complete radial tear at the posterior horn of the medial meniscus with the femoral resurfacing device in place (C). Each knee was tested in static knee stance positions (5°, 15°, 30°, 45°) and through 10 dynamic knee flexion cycles (5–45°) with body weight ground reaction force. The targeted level of ground reaction force during the trial was set according to the living body weight of the cadaver donor (e.g. 70 kg to 700 N). The loading rate for exploring the dynamic contact pressures was 0.1 Hz/s.

All procedures were performed by the same investigator (CB). All specimens were released at their femoral fixation during preparation of a new testing condition. The exact position was marked and recorded to maintain specimen position in the knee simulator across all testing conditions. The position of the sensor did not change during the preparation. The knee was flexed to 90° to expose the central weight bearing portion. A drill guide was used to place a pin perpendicular to the joint surface representing the center of the defect. The center of the defect was determined by measuring the condylar width and bisecting the distance. The cannulated instrumentation set ensured that the vertical axis was maintained throughout the procedure. After drilling a pilot hole, the fixation component was inserted. A contact probe determined the radius of curvature in two planes. With a matching reamer, the site for the implantation was prepared and a sizing trial with corresponding offsets inserted to confirm the fit to the surrounding articular surface. The selected articular component was oriented in the correct planes and connected to the anchoring screw with a tapered lock. After testing with flush implantation (Fig. 3B), the sensor was released from its posterior fixation and a piece of Teflon foil inserted between the sensor and the posterior horn of the medial meniscus to protect the sensor (Fig. 3C).
tached to the posterior capsule and the knee cycle tests were repeated.

Data was obtained using I-Scan software 4.23 (Tekscan, Boston, MA, USA) with a frequency of 10 values of peak contact pressure per second. Maximum peak contact pressure was assessed and recorded as the highest value at each stance position and the highest value of all measured peak contact pressures during the dynamic knee bending cycle. Furthermore, the average peak contact pressure of all measured values of the dynamic knee bending cycle was determined. Mean and standard deviation values were evaluated using SPSS 11.0 (SPSS Inc., Chicago, IL, USA). For statistical analysis a paired sampled t-test was used to compare means (significance, \( P \leq 0.05 \)).

3. Results

Flush device implantation resulted in no statistically significant differences when compared to the untreated normal knee. An average increase in maximum peak contact pressures ranging from 4\% to 8\% was observed during static testing (5\textdegree, 15\textdegree, 30\textdegree, 45\textdegree) and dynamic knee bending cycle (Table 1, Fig. 4). The average peak contact pressure through the dynamic knee bending cycle was 4.71 (SD 0.55) MPa for the untreated knee compared to 4.66 (SD 0.66 MPa after flush implantation. Starting with the third dynamic knee bending cycle, all consecutive peak contact pressure values appeared to match both testing conditions precisely (Fig. 5).

After placement of a complete radial tear in the posterior horn of the medial meniscus, no significant different maximum peak contact pressures were observed at 5\textdegree stance position. However, a significant increase of mean maximum peak contact pressures was found at 15\textdegree, 30\textdegree and 45\textdegree static stance positions when compared to flush device implantation with a normal meniscus (63\%, 57\%, and 57\%, all \( P < 0.05 \)) (Table 1, Figs. 4 and 6). An average increase of 78\% (\( P < 0.05 \)) was observed throughout the dynamic knee flexion cycle (Table 1, Fig. 4). Accordingly, on average, significant differences were found in any testing condition except the 5\textdegree stance position comparing the normal knee with the knees with a non-functional meniscus after creating the radial tear (Table 1, Fig. 4). Average peak contact pressure through the dynamic knee bending cycle with femoral resurfacing and complete radial tear at the posterior horn of the medial meniscus (6.35 MPa, SD 1.55) was significantly (\( P < 0.05 \)) increased compared to the untreated knee and flush implantation with intact meniscus respectively.

4. Discussion

Although articular cartilage defects can be asymptomatic and the precise likelihood of a defect becoming symptomatic is unknown (Messner and Maletius, 1996), defects have shown to further progress and cause substantial pain and functional impairment (Linden, 1977; Messner and Maletius, 1996; Shelbourne et al., 2003). The treatment of full thickness chondral and osteochondral defects offers a great variety of options depending on the severity of the damage. However, existing treatment options for middle-aged patients (40–50 years old) have several

<table>
<thead>
<tr>
<th>Testing position</th>
<th>Testing condition</th>
<th>Peak contact pressure and standard deviation (MPa)</th>
<th>Mean difference to untreated knee (A) testing condition (%)</th>
<th>( P ) value</th>
<th>Mean difference to flush device (B) implantation (%)</th>
<th>( P ) value</th>
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</thead>
<tbody>
<tr>
<td>5\textdegree stance</td>
<td>(A) Untreated knee</td>
<td>4.22 (SD 1.09)</td>
<td>N/A</td>
<td>N/A</td>
<td>–7</td>
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<td></td>
<td>(B) Flush implantation</td>
<td>4.54 (SD 1.34)</td>
<td>8</td>
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<td></td>
<td>(C) Meniscus tear</td>
<td>4.30 (SD 2.02)</td>
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<td>15\textdegree stance</td>
<td>(A) Untreated knee</td>
<td>3.87 (SD 1.16)</td>
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<td>N/A</td>
<td>–5</td>
<td>ns</td>
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<tr>
<td></td>
<td>(B) Flush implantation</td>
<td>4.09 (SD 1.58)</td>
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<td>N/A</td>
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<tr>
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<td>(C) Meniscus tear</td>
<td>6.66 (SD 3.50)</td>
<td>72</td>
<td>(&lt;0.05)</td>
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<td>(&lt;0.05)</td>
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<td>30\textdegree stance</td>
<td>(A) Untreated knee</td>
<td>4.61 (SD 1.92)</td>
<td>N/A</td>
<td>N/A</td>
<td>–6</td>
<td>ns</td>
</tr>
<tr>
<td></td>
<td>(B) Flush implantation</td>
<td>4.93 (SD 2.58)</td>
<td>7</td>
<td>N/A</td>
<td>N/A</td>
<td>(&lt;0.05)</td>
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<tr>
<td></td>
<td>(C) Meniscus tear</td>
<td>7.73 (SD 3.66)</td>
<td>68</td>
<td>(&lt;0.05)</td>
<td>57</td>
<td>(&lt;0.05)</td>
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<tr>
<td>45\textdegree stance</td>
<td>(A) Untreated knee</td>
<td>5.51 (SD 2.30)</td>
<td>N/A</td>
<td>N/A</td>
<td>–4</td>
<td>ns</td>
</tr>
<tr>
<td></td>
<td>(B) Flush implantation</td>
<td>5.73 (SD 2.44)</td>
<td>4</td>
<td>N/A</td>
<td>N/A</td>
<td>(&lt;0.05)</td>
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<td>(C) Meniscus tear</td>
<td>8.99 (SD 3.51)</td>
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<td>(&lt;0.05)</td>
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<td>Dynamic flexion</td>
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<td>–5</td>
<td>ns</td>
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<tr>
<td></td>
<td>(B) Flush implantation</td>
<td>6.51 (SD 2.67)</td>
<td>6</td>
<td>N/A</td>
<td>N/A</td>
<td>(&lt;0.05)</td>
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<td>(C) Meniscus tear</td>
<td>11.61 (SD 6.39)</td>
<td>89</td>
<td>(&lt;0.05)</td>
<td>78</td>
<td>(&lt;0.05)</td>
</tr>
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</table>
drawbacks (Kirker-Head et al., 2006; Kreuz et al., 2006; Mithoefer et al., 2005; O’Driscoll, 1998). Considered as being too old for biological repair of the defect or repeat procedures, these patients are mostly managed with conservative, non-surgical treatment or unicompartimental or total knee arthroplasty, which represent procedures of final resort for some of the affected patients. Many middle-aged patients are not only affected by articular cartilage defects; meniscal tears often coexist with articular damage involving deep ulcerations (Noble and Hamblen, 1975). This was confirmed in large arthroscopic series finding meniscal tears and concomitant cartilage defects (Curl et al., 1997; Hjelle et al., 2002). The loss of load bearing capability of injured menisci with damage of the integrity of the circumferential fibres and situations after total or partial meniscectomy was shown to result in increased peak contact pressures in the femorotibial joint (Jones et al., 1996; Lee et al., 2006; von Lewinski et al., 2006).

**Fig. 4.** Mean maximum peak contact pressures according to testing position and testing condition with standard deviation. Significant increase of mean maximum peak contact pressures (*) denotes statistical significance at $P < 0.05$ was found with femoral resurfacing and a complete radial meniscus tear (C) compared to the untreated knee (A) and femoral resurfacing without a meniscus tear (B).

**Fig. 5.** Static loading from $5^\circ$ to $45^\circ$ with single body weight ground reaction force, followed by 10 cycles of dynamic loading. The mean values of peak contact pressures are displayed as curves. Marked increase of pressure is found with a resurfaced knee and a non-functional meniscus (C) compared to the untreated knee (A) and resurfaced knee with intact meniscus (B).
This in vitro study aimed to determine the effect of a complete radial tear on peak contact pressure in the tibiofemoral joint comparing the untreated normal knee with a surface matching inlay resurfacing device implanted in the femoral condyle. The HemiCAP® resurfacing prosthesis may offer an alternative treatment strategy for the middle aged patient with a focal full thickness articular cartilage defect. The device is approved for international marketing via CE Mark and other regulatory bodies. The clinical use is currently under investigation. However, clinical data for the device is not yet available. Furthermore, it is unknown if the femoral resurfacing device leads to increased peak contact pressure with deleterious effects to the opposing biological structures such as meniscus and articular cartilage of the tibial plateau.

Although we found an average increase of 4–8% of maximum peak contact pressure after flush implantation of the resurfacing device, our results demonstrated no statistically significant difference compared to the untreated knee. This is consistent with findings of a previous biomechanical study (Becher et al., 2008) and suggests biomechanical safety of the device with correct implantation. Results in a biomechanical model using osteochondral plugs for the treatment of osteochondral defects demonstrated no significant differences in mean peak pressures when comparing intact cartilage with flush, recessed, and angled grafts using a single osteochondral graft with 4.5 mm diameter. However, pressures were higher compared to intact cartilage for flush, 14.5%, and angled grafts (6.7%) (Koh et al., 2006). Kock et al. (2008) reported a significant decrease of contact pressures at the boarder of the defects after using mosaicplasty in 16 mm defects with 3 plugs. Compared to the untreated knee, however, pressures were still 35% increased. In a previous study, increased peak contact pressure was demonstrated with the HemiCAP® resurfacing prosthesis being 1 mm elevated to the adjacent cartilage at an average maximum increase of 217% compared to the untreated condition in human cadaver knees with intact chondral surfaces and menisci (Becher et al., 2008). Thus, proud device implantation to the adjacent cartilage should be avoided.

Radial tears are frequently found among medial meniscus tears. In an arthroscopic study, the authors found that radial tears of the posterior horn accounted for 27.8% of all medial meniscus tears reviewed (Bin et al., 2004). We created a situation with a non-functional posterior horn of the medial meniscus by implementing a complete radial tear close to the medial attachment of the posterior horn according to the typical clinical illustration (Habata et al., 2004). Radial tears extending to the periphery with disruption of circumferential result in the loss of hoop strain and have been shown to be equivalent to total meniscectomy in load bearing terms (Seedhom et al., 1974). Jones et al. (1996) found 50% reduced strains in radial tears anteriorly whilst a complete radial tear completely defunctioned the meniscus. Lee et al. performed serial posterior medial meniscectomies in a human cadaver model resulting in increased peak contact pressures proportionally to the amount of meniscus removed (Lee et al., 2006). Whereas 50% radial width meniscectomy resulted in an average increase of 43% in peak contact pressures, pressures increased on average 136% after total medial meniscectomy (Lee et al., 2006). Increased peak contact pressure of meniscectomized knees at an average of even 260% compared to the control knees was found in an ovine model (von Lewinski et al., 2006). We could demonstrate an average increase of maximum peak contact pressure of 57–89% depending on the testing condition with the non-functional medial meniscus and the resurfacing prosthesis in place compared to the situation with an intact meniscus. These values fall well within the range of values reported and other previous studies (Baratz et al., 1986; Cottrell et al., 2008; Krause et al., 1976; Radin et al., 1984).

As a clinical implication, expectations regarding the clinical success with the prosthetic device should be lowered in comparison to cases with intact meniscus. However, a limitation from our testing setup is that it remains unclear if the elevation in pressure that occurred with the HemiCAP® and the meniscal tear together could have been due to the tear alone, and may have had not been related to the implantation of the resurfacing device. Regarding reported values of the literature with lack of meniscus function (Baratz et al., 1986; Lee et al., 2006) and considerably higher pressures with an elevated implant to the adjacent cartilage (Becher et al., 2008) compared to our results with a complete radial meniscus tear, it could be hypothesised that implantation of the device would cause no significant change in pressure when a tear is present.

Further limitations of this human cadaver study and biomechanical model have to be considered: Comparisons between our and other studies are complicated by differing experimental conditions and methods for measuring knee contact mechanics. Only approximation of the living system can be achieved. However, pre-selection of the applied load by the ground reaction force gave the opportunity to load the specimens with the known body weight of the donor, and thus better approximate the forces that occur in the living system (von Skrbensky and Huber, 2006). Furthermore, the knee bending dynamic measurement may have provided a more accurate reproduction of physiologic weight bearing activity. Peak contact pressures were determined by an electronic pressure sensitive sensor with known limitations including the thickness (0.1 mm), potential for creep, its sensitivity to temperature changes, its disposition for crinkling and the establishment of the position (Beck et al., 2005). Although the position of the
sensor was accurately secured, a small amount of displacement and crinkling could not be excluded. A certain amount of sensitivity loss has to be considered resulting from various testing conditions. A small amount of data point dropouts were observed at some specimens. However, peak contact values did not seem to be affected. Testing was performed in a range from 5° to 45° flexion and the implantation of the device was performed in the central weight bearing area. Results may be different for the implantation in other areas of the medial or lateral femoral condyle and at different ranges of motion angles.

In conclusion, the data suggests that resurfacing with the prosthetic device alone with an intact meniscus does not lead to significant increase in tibiofemoral peak contact pressure. However, results confirm that a non-functional meniscus leads to a biomechanical disadvantage. Possible effects of reduced meniscal tissue and biomechanical integrity of the meniscus must be considered in an in vivo application.

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