Thigh–calf contact force measurements in deep knee flexion

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Abstract

Background. Knee models often do not contain thigh–calf contact which occurs in deep knee flexion. Thigh–calf contact is expected to reduce muscle forces and thereby affects internal stresses in the knee joint. The purpose of this study was to measure thigh–calf contact forces. Two deep knee flexion activities were selected: squatting and kneeling.

Methods. Ten healthy subjects participated in the experiment. Contact pressures between the thigh and calf were measured using the Tekscan Conformat pressure mapping sensor. Knee flexion angles were measured unilaterally using an infrared motion capture system. Contact forces were averaged in terms of means and standard deviations. The magnitude and location of the resultant contact force were calculated. Correlations between anthropometric subject parameters and experimental outcome were studied.

Findings. In general, thigh–calf contact did not take place below 130° knee flexion. The average maximal contact forces for each leg were 34.2% bodyweight during squatting and 30.9% bodyweight during kneeling. Corresponding average maximal knee flexion angles were 151.8° during squatting and 156.4° during kneeling. Thigh and calf circumferences were correlated with the contact force measurements.

Interpretation. The current study shows that thigh–calf contact is substantial (>30% bodyweight on one leg) and likely reduces the forces inside the knee during deep knee flexion. Subsequently, total knee replacements may be subjected to lower loads than assumed before, which reduces the risk of implant failure at large flexion angles. Results presented in this study can be utilized in knee models that focus on deep knee flexion.

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1. Introduction

Total knee replacement (TKR) is a widely used and successful surgical procedure. In 2003 a total of 418000 TKR and 33000 revision procedures were performed in the USA (National Hospital Discharge Survey, 2003). New developments are continuously made to improve implant performance. One of the latest developments is the so-called high-flexion knee prosthesis, which allows flexion angles larger than 120°. The development of this type of prosthesis is mainly a result of two concurrent trends. Firstly, due to its success, TKR is applied to younger and more active patients. The more active way of living demands a prosthesis with a larger range of motion. Secondly, the number of patients undergoing TKR surgery in non-Western countries is growing steadily. Studies on implants used in non-Western cultures report the necessity for high-flexion knee implants due to local daily living activities like kneeling and squatting (Mulholland and Wyss, 2001).

The development of high-flexion knee implants puts higher demands on implants, both in a kinetic and a kinematic perspective. Kinematic studies have been performed to investigate high-flexion knee behavior. As an example, several studies reported an asymmetric femoral roll-back mechanism during squatting and kneeling, which was caused by an internal tibial rotation (Hefzy et al., 1998; Conditt et al., 2006). Other studies reported the separation
of the medial condyle and medial tibial plateau occurring at high knee flexion angles (Nakagawa et al., 2000; Conditt et al., 2006).

Over the last decades the finite element (FE) method has proven its value for TKR research and has recently been utilized in high-flexion knee research. Morra and Greenwald (2005) used FE models to calculate Tibio-femoral contact stresses in different high-flexion knee implants. For some types of implants, they reported stresses above the yield point of the polyethylene insert at high knee flexion angles. Barink et al. (Submitted for publication) used FE models to compare a conventional knee prosthesis with a high-flexion knee prosthesis of the same manufacturer. In general, they found higher implant stresses with increasing flexion angles, which was primarily caused by higher quadriceps forces occurring at these higher flexion angles. They also showed that the high-flexion knee prosthesis did outperform the conventional prosthesis in the high-flexion range.

The outcome reliability of finite element analyses depends on input parameters such as joint forces applied to the finite element models. Joint forces are often estimated by simplified musculo-skeletal models using inverse dynamics. Most of these models do not include thigh–calf contact, which occurs in high knee flexion. Nagura et al. (2002) measured high knee flexion kinematics and calculated corresponding net knee joint forces and moments using an inverse dynamics model. This model contained estimations of segment dimensions and mass distributions. However, the model did not contain thigh–calf contact, which they reported as a shortcoming of their model. Caruntu et al. (2003) calculated knee joint forces during deep knee flexion using a mathematical model, which contained thigh–calf contact. With this model they showed that thigh–calf contact could lead to a considerable reduction of quadriceps and hamstring forces. Nonetheless, the model used in their study was a simplified and two-dimensional representation of the human knee and was not validated.

Because thigh–calf contact is often neglected, musculo-skeletal models would typically predict higher knee joint forces with increasing flexion angles, even in the high-flexion range. This seems to contradict the fact that people can squat for long periods of time in a relaxed manner, which is possibly caused by contact between the thigh and calf. Our hypothesis is that thigh–calf contact is substantial, reduces muscle forces in the knee during high knee flexion and should in that case not be neglected in models that focus on deep knee flexion.

Thus far, no prior studies were found which actually quantified the thigh–calf contact characteristics. The purpose of the current study was to gather information on thigh–calf contact by measuring its pressure distribution in relation to the knee flexion angle. Two high-flexion activities were included in this study: squatting and kneeling. The pressure distributions were used to calculate the magnitude and location of the resultant contact force on the calf, which can be used in further research. Furthermore, we hypothesized that anthropometric properties affect thigh–calf contact characteristics. Hence, we investigated whether we could detect any trends between body mass index related subject properties and the thigh–calf contact characteristics.

2. Methods

2.1. Subjects

Ten healthy subjects (8 male and 2 female) were included in this study as, at this point of time, TKR patients do not yet receive high-flexion TKR components in The Netherlands. A group size of 10 subjects was selected as this group size was deemed to be appropriate to create the data that typically describes the thigh–calf contact conditions. The mean age of the subjects was 28.4 yr (SD 6.0), the mean body mass (BM) was 71.5 kg (SD 15.7), the mean length was 181 cm (SD 9.2) and the mean body mass index (BMI) was 21.6 kg/m² (SD 3.4). Subjects were fully informed and agreed to participate in the experiment.

2.2. Materials

Contact pressure between thigh and calf was measured using the Tekscan Conformat pressure mapping sensor (model #5330, Tekscan, South Boston, MA, USA). This sensor has been developed for research on decubitus and seating optimizations, which produce similar pressure magnitudes as presumably encountered in this study. In other studies, Tekscan sensors have been used to measure, for instance, articular contact areas and stresses (Harris et al., 1999; Wilson et al., 2006). The Conformat sensor has a sensing area of 47.1 by 47.1 cm with 1024 sensing elements (sensels) distributed over 32 rows and 32 columns, a thickness of 1.78 mm, a spatial resolution of 0.5 sensel per square cm and a sensitivity range of 0–33.3 kPa for each sensel. Before application, the Conformat was conditioned by loading and unloading it several times and by equilibrating and calibrating the sensor using the instrumentation provided by the manufacturer. After the sensor was conditioned, it was inserted between the thigh and calf of both legs of the subjects. Tekscan contact pressure distributions were recorded with a frequency of 8 Hz using I-Scan version 5.72 (Tekscan, South Boston, MA, USA).

Knee flexion angles were measured unilaterally using an infrared five-camera motion capture system (Qualisys AB, Gothenburg, Sweden). Reflective markers were attached to the trochanter major, lateral epicondyle and lateral malleolus and represented the hip, knee and ankle joint positions respectively (see Fig. 1). Motion data were collected with a frequency of 60 Hz using Qtrac Capture version 2.77 (Qualisys AB, Gothenburg, Sweden). To link the pressure and kinematic measurements, both systems were synchronized using an infrared led which was switched on
manually when the Tekscan recordings were started and which served as a data marker in the Qualisys recordings.

Leg segment lengths and circumferences of the subjects were measured as well as the distance between the posterior knee and the epicondylar axis. This epicondylar distance (see Fig. 1) was used to transfer from a sensor specific coordinate system to a subject specific coordinate system in order to relate the contact measurements to the calf.

2.3. Squatting and kneeling protocol

The subjects performed two activities: squatting and kneeling. The squatting activity was defined by descending from an erect posture to a squatting position with no heel-ground contact (Fig. 2a) and the kneeling activity by descending from an erect kneeling position to a deep kneeling position with ankles dorsal flexed and knee-ground

2.4. Data analysis

Contact pressure distributions were studied and plotted using I-Scan. Data analysis and processing was done in Matlab 7.1 (The Mathworks, Natick, MA, USA). Tekscan and Qualisys recordings were synchronized and resampled using linear data interpolation. For each subject, the pressure distributions of both legs were averaged to obtain a typical pressure distribution for one leg. Pressure distributions were translated to force distributions by multiplying contact pressures by contact areas. The resultant contact forces and their locations on the calf were calculated using the contact force data and the resultant contact forces were normalized for body weight (BW). Knee flexion angles were determined using the marker positions of the ankle, knee and hip and the cosine rule. Resultant contact forces and their locations on the calf vs. flexion angle curves were obtained and translated to one mean curve per activity for each subject. The starting flexion angle of thigh–calf contact was defined as the flexion angle at which the resultant contact force on one leg became larger than 5% BW.

2.5. Statistical analysis

The maximal contact forces, the maximal contact areas, the maximal flexion angles and the starting flexion angles of thigh–calf contact were averaged for all subjects in terms of means and standard deviations. The Student’s t-test was used to assess differences between the maximal contact force occurring during squatting and kneeling. Moreover, differences between the maximal thigh–calf contact force occurring at the dominant and non-dominant leg were assessed (dominant leg comparison). The Pearson’s linear correlation coefficient (for a 95% confidence interval) was used to quantify trends between BMI related subject properties and the thigh–calf contact characteristics. Hence, relations were determined between anthropometric properties of the subjects (circumferences of thigh and calf, BMI and BM) and experimental outcome (maximal contact force, maximal flexion angle and starting flexion angle of thigh–calf contact).

3. Results

We found that thigh–calf contact pressures exponentially increased with increasing knee flexion angles. Parameters such as the contact area and the resultant contact force were all maximal at the maximal knee flexion angles reached by the subjects (Table 1).
3.1. Contact area

Fig. 3 shows two typical contact pressure distributions at maximal knee flexion, obtained with I-Scan. The average maximal contact area was 215.2 cm² during the squatting activity and 240.8 cm² during the kneeling activity. In most recordings, the peak pressures were located close to the posterior knee, but in some recordings the peak pressures were found more distal on the calf. Most subjects exhibited a slightly asymmetric contact distribution between both legs.

3.2. Contact force

In general, thigh–calf contact did not take place below 130° knee flexion (during both activities, Table 1). Evident thigh–calf contact (>5% BW for one leg) initiated at a lower average flexion angle during squatting (134.8°) compared to kneeling (144.8°). Contact forces were maximal when the knees were maximally flexed. The average maximal flexion angles were lower during squatting (151.8°) compared to kneeling (156.4°). However, the corresponding average maximal contact forces on one leg were higher during squatting (34.2% BW) compared to kneeling (30.9% BW). The average location of the resultant contact force was closer to the epicondylar axis during squatting (15.1 cm) compared to kneeling (16.6 cm). For the squatting activity, a typical relationship between the knee flexion angle and the average resultant contact force is given in Fig. 4, together with a typical relationship between the knee flexion angle and the location of the resultant force on the calf.

3.3. Correlations

Most of the significant correlations and trends ($P < 0.05$) we found were more evident in the kneeling results, in comparison with the squatting results. This was caused by lower standard deviations in the kneeling results. In general, standard deviations were relatively high in both activities, indicating high variability amongst subjects.

The maximal contact force during squatting was not statistically different from the maximal contact force during kneeling ($P = 0.28$) within our results. Thus, thigh–calf contact during both activities was comparable in trend and magnitude. The maximal thigh–calf contact force at the dominant leg (mostly the right leg) was statistically different from the maximal contact force at the non-dominant leg during squatting ($P = 0.02$) and kneeling ($P = 0.04$). Hence, roughly 60% of the maximal thigh–calf contact transferred through contact on the dominant leg (59% during squatting and 55% during kneeling).

As expected, we found that a subject’s BM was significantly related to the maximal thigh–calf contact force (in Newtons) during both squatting ($R = 0.73$) and kneeling ($R = 0.72$). In addition, we found that the thigh and calf

<table>
<thead>
<tr>
<th>Activity</th>
<th>Start of contact</th>
<th>Maximal contact</th>
<th>Contact area (cm²)</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Flexion angle (deg)</td>
<td>Flexion angle (deg)</td>
<td>Res. contact force (% BW)</td>
</tr>
<tr>
<td>Squatting ($n = 10$)</td>
<td>134.8 (5.92)</td>
<td>151.8 (4.39)</td>
<td>34.2 (9.69)</td>
</tr>
<tr>
<td>Kneeling ($n = 9$)</td>
<td>144.8 (3.52)</td>
<td>156.4 (3.43)</td>
<td>30.9 (9.31)</td>
</tr>
</tbody>
</table>

Average values and standard deviations (in brackets) are given.

a Due to erroneous data one subject was not included.

Fig. 3. Two typical thigh–calf contact pressure distributions of one subject (weight = 78 kg) captured at maximal knee flexion angle: (a) squatting: flexion angle $\approx 153°$, contact force = 445.0 N, contact area = 0.0397 m² and (b) kneeling: flexion angle $\approx 155°$, contact force = 424.7 N, contact area = 0.0473 m². The diamond in the figures represents the location of the center of pressure.
circumferences were significantly related to both the starting flexion angle of thigh–calf contact and the maximal contact force (in Newtons, Table 2). Hence, the larger the corpulence of the thigh and calf, the smaller the range of motion of the knee before the start of thigh–calf contact and the higher the maximal contact forces. A similar trend was found for the correlations between a subject’s BMI and the maximal flexion angle during squatting ($R = 0.74$) and the maximal contact force during kneeling ($R = 0.74$).

4. Discussion

In the current study we measured the thigh–calf contact forces occurring during deep knee flexion. Both the thigh–calf contact force and the corresponding knee flexion angle were measured during two high-flexion activities: squatting and kneeling. It was shown that thigh–calf contact is substantial (>30% BW on one leg) and is likely to have a considerable effect on forces inside the knee joint.

Overall, the methodology used in this study functioned well, but there are some limitations. Keeping the pressure mapping sensor in place manually, which possibly led to small variations in the results. Another limitation is the fact that both activities were rather unrestrained and subject dependent, which probably affected the experiment reproducibility. We tried to improve this reproducibility by repeating the activities during one recording as described in the methods.

The instrumentation used in this study also had some limitations. Firstly, the recording frequency of the Tekscan system was 8 Hz, which was low for our measurements. We tried to compensate this frequency by instructing the subjects to perform both activities with a low velocity and to hold the maximal flexion angle for a short period of time. Secondly, Tekscan sensors are designed to measure compressive forces. Shear forces may cause errors in the compressive stress measurements. However, the Tekscan Conformat used in this study differs from other Tekscan sensors (e.g. K-scan) and has been designed for comfort studies of seats and cushions. The Conformat is very flexible and allows almost shear stress free motion of the different sensor elements. Theoretically, the Conformat is therefore less sensitive to shear forces in comparison with other Tekscan sensors. However, we could find no reports in which this was validated. In the current application, the Conformat was subjected to an almost pure compression movement with minimal shear. Hence, cross-talk between shear and compression was most likely negligible in this study.

Finally, it is reported that Tekscan sensor characteristics can slightly drift during measurements (Ferguson-Pell et al., 2000). Although, recent studies report acceptable accuracy compared to current standards such as Fuji film (Bachus et al., 2006; Wilson et al., 2006), we carefully conditioned the sensor before application following the instructions provided by the manufacturer. In addition, we determined the sensor linear calibration curves before and after the measurements to estimate the repeatability of our measurements. Deviations between both calibration curves were within 5% and therefore deemed acceptable for our purpose.

Table 2

<table>
<thead>
<tr>
<th>Subject properties</th>
<th>Start flexion angle</th>
<th>Maximum flexion angle</th>
<th>Maximum contact force in N</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>R(s)</td>
<td>R(k)</td>
<td>R(s)</td>
</tr>
<tr>
<td>Thigh circumf.</td>
<td>–0.54</td>
<td>–0.70*</td>
<td>–0.63</td>
</tr>
<tr>
<td>Calf circumf.</td>
<td>–0.79*</td>
<td>–0.82*</td>
<td>–0.68*</td>
</tr>
<tr>
<td>BMI</td>
<td>–0.42</td>
<td>–0.69*</td>
<td>–0.81*</td>
</tr>
<tr>
<td>Weight</td>
<td></td>
<td></td>
<td>0.73*</td>
</tr>
</tbody>
</table>

*Significant correlations ($P < 0.05$) are marked. $R(s)$ is the Pearson’s linear correlation coefficient for squatting and $R(k)$ for kneeling.

During the squatting activity, the subjects had to keep the sensor in place manually, which possibly led to small variations in the results. Another limitation is the fact that both activities were rather unrestrained and subject dependent, which probably affected the experiment reproducibility. We tried to improve this reproducibility by repeating the activities during one recording as described in the methods.

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Fig. 4. A typical relation (black curve) between the knee flexion angle and the average resultant contact force on one leg (Fig. 4a) and its location on the calf (with respect to the epicondylar axis, Fig. 4b) during the squatting activity. The grey region represents the area in which the curves of all subjects were found.
In this study we considered a relatively small group of young and healthy Western subjects with a relatively low body weight. If the results of this study were to be applied to Western TKR patients, one should realize that these patients often have a considerably higher BMI, which has significant effects on the thigh–calf contact mechanics as demonstrated in this study. In fact, this study indicates that thigh–calf contact load transfer may increase for TKR patients that typically have a higher BMI. Direct application of the results to Asian or Arabic TKR patients is probably more relevant as these patients have BMI's that are comparable to the subjects as measured in this study, although they are used to perform activities at higher flexion angles.

The maximal flexion angles measured in this study (151.8° during squatting and 156.4° during kneeling) are comparable to values we found in the literature. Conditt et al. (2006), Hefzy et al. (1998) and Nagura et al. (2002) reported average maximal knee flexion angles of 155.8°, 157.3° and 155° respectively during kneeling measurements.

The average maximal thigh–calf contact forces were 34.2% BW during squatting and 30.9% BW during kneeling. Thigh–calf contact will substantially affect forces inside the knee, since its moment arm with respect to the epicondylar axis is large (15–17 cm) in comparison with the moment arm of, for instance, the quadriceps muscle (4–5 cm; Ward et al., 2005). The exact effect, however, needs to be assessed in future biomechanical studies.

5. Conclusions

Outcome of this studies supports our hypothesis that thigh–calf contact is substantial (>30% BW on one leg) and should not be neglected in calculations that focus on deep knee flexion. With the data presented in this study more realistic high-flexion knee simulations can be obtained.

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References


