In-Vitro Measurement of Patellofemoral Joint Contact Pressure During Various Kinds of Walking

Shunji HIROKAWA*

The objective of this study is to measure the variation of patellofemoral joint contact pressure under the condition of simulating the variations of the knee flexion angle, and the thigh muscle forces during level walking, climbing and descending stairs. It will compare the results with those of the previous experimental studies in which the thigh muscle forces had been kept constant.

Using the data from the literature and applying a pure mathematical calculation, the values of tensile forces applied to the thigh muscles were determined as a function of knee flexion angle, to simulate various walking conditions. In the experiment, data was obtained by the variation of electrostatic capacity of a specially designed film inserted between the contact surfaces of the patellofemoral joint.

The experimental results demonstrated that when the thigh muscle force was kept constant, the mean contact pressure was almost uniform throughout the whole range of knee flexion, while the contact force showed a dome shape indicating the highest value at about 45° knee flexion. When the values of thigh muscle force and the knee angle simulating various walking conditions were applied, the results demonstrated that the patellofemoral contact pressures were almost constant all of the time of the stance phase of a gait cycle, also the peak values of pressure were almost equal among three kinds of walking. It was also revealed that the articular surface of contact portion where larger contact force occurs produces a smaller curvature.

The patellofemoral articular surfaces are seemingly irregular in shape, however, the complicated shapes may play an important role in moderating steep variations of patellofemoral contact pressure during walking.

Key Words: patellofemoral joint, contact pressure, joint morphology, various walkings

1. Introduction

Although it is widely accepted that mechanical factors are instrumental in the development of Osteoarthrosis (OA), their role in the development of the diseases is not clear and subject to diametrically opposed views.

Recently, the majority of studies on the patellofemoral joint mechanism, not only experimentally but also analytically, have begun to study about the biphasic deformation characteristics of articular cartilage1~4).

Yet, experimental and/or analytical study to inves-


tigate joint contact pressure in association with a knowledge of the morphology and geometry of a joint is essential for the understanding of many biomechanical functions of diarthrodial joints. The pressure distribution including the location and magnitude of peak pressure, is a significant measurement relating to the pathomechanics of osteoarthritis.

There have been many experimental studies performed concerning the patellofemoral contact force and pressure, in which thigh muscle forces have been kept constant. Reilly and Martens5) investigated the quadriceps muscle force and the patellofemoral joint reaction forces for various locomotive activities. Goodfellow et al.6) determined the contact areas in the patellofemoral compartment at various flexion angles of the knee, using a multiple staining technique, which attributed the surface degenerative changes to habitual disuse. Matthews et al.7) also

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calculated the patellofemoral contact stress for various activities by assuming that the patella acts as a frictionless pulley. Roentgenograms from a lateral view were taken and the angle between the line of action of the patellar tendon and that of the quadriceps, (named "patellar mechanism angle"), was used in combination with the quadriceps force to calculate patellofemoral contact force. The data introduced by the above-mentioned pioneers, however, lack credibility for the following reasons; First, the assumption that the magnitude of the quadriceps force is equal to that of the patellar tendon does not hold. Next, for the contact area measurement, serious consideration was lacking about the effects of the magnitude of the applied quadriceps force and the loading duration applied on variation of contact area, thus the contact area data may not be reliable.

As for the experimental results concerning the patellofemoral contact pressure during various activities, there seems to be no other comparable report than Seedhom et al.'s. They have measured the patellofemoral joint forces and pressure using their newly designed measuring tool. They have attempted to relate the sites of the lesions on the femoral surfaces of the patellofemoral compartment to the contact that occurs during the range of flexion of the knee joint and the loads transmitted, hence to the pressure arising during normal activities (i.e. level walking, ramp and stair ascent and descent).

Previously, the author had performed simulation analysis using a three-dimensional mathematical model of the patellofemoral joint and determined variations of patellofemoral pressure for level walking, climbing and descending stairs. In the simulation, the patellofemoral contact pressure was introduced as a function of the thigh muscles force and knee flexion angle. To determine the thigh muscles force and knee angle, a pure mathematical calculation (based on a two-dimensional model) was performed using literature and experimentally determined data. First, a musculoskeletal model of a lower limb was presented, then, the thigh muscle (the quadriceps muscle) force was calculated by substituting angular and momentum data of the lower limb during various kinds of walking for the model equation. The simulation results demonstrated that mean pressures on the patellofemoral joints were rather uniform during the stance phase of gait, and their maximum values did not vary much among the three kinds of walking.

The objective of the present study is to directly measure patellofemoral contact pressure under such conditions that simulate successive variations of the knee flexion angle and the thigh muscle forces for several kinds of walking through in-vitro measurement in order to complement the previously obtained simulation results as well as to compare the results with the above-mentioned Seedhom et al.'s.

2. Material and Method

The experimental setup is shown in Photo. 1. A fresh frozen cadaver knee was mounted vertically. Aluminum grips were fixed to the quadriceps and hamstring tendons with flexible cables, strained so as to approximate the muscle lines of action. The tibia was fixed from full extension to 120° flexion at its distal end. The femur and tibia bones were attached to the fixing devices. Eight centimeters at the proximal end of the femur and the distal end of the tibia were cleaned of all soft tissue for later attachment onto the fixing devices. The bones were firmly secured by several screws, some of which penetrated the bones to prevent any motion. Metal clamps were secured to the quadriceps and the hamstrings, flexible steel cables were extended from the clamps, approximating the lines of action of the muscles. Light aluminum grips were fixed to the quadriceps and hamstrings tendons with flexible cables, extending from the grips over two separate pulleys, approximating the muscles line of action. Weights were attached to the cables. The knee was fixed at one of the desired positions.
angles.

The rod extending from the tibia was allowed motion of five degrees of freedom, necessary to accommodate the knee's shifting center of rotation. The tibia were fixed at 0° flexion (full extension), and at every 15° interval up to 120° flexion at its distal end. Free flexion-extension of the knee was the only degree of freedom which was not allowed in order to create isometric conditions.

An instrument specially designed for measuring contact pressures (SPECTROMASTER Yokohama Inst. Co.) was used in this study. Contact pressure values were measured by the variation of electrostatic capacity of the pressure sensitive elements arranged on the film. Table 1 shows the specification of the instrument and the film, and Fig. 1 shows the size and arrangement of the sensor points of the film. The instrument enjoyed some advantages such as a high S/N ratio, a closely linear relation between input and output data, and a quick response characteristic. The film, 0.4 mm thick, with 24 measuring points, was moderately transformed the surface for good fitting to the articular surface. The sensitive elements could properly detect pressure even though wrinkles ran on the sensitive points of the film.

Longitudinal cuts were made in the medio and lateral fascias and then two films were inserted into the medial and lateral sides respectively, of a gap between the patellar and femur articulating surfaces. The insertions of the films were adjusted so that the areas where the sensors were arranged might fully cover the joint contact areas, which had been predicted through a preliminary trial.

Seven kinds of loads, ranging from zero to 200 Newton, were applied to a bunch of the vastus intermedius and rectus anterior tendons by means of a clamp, for 0 to 120° knee flexion at 15° intervals, while the vastus medialis and lateralis were cut and disregarded. The maximal load of 200 Newton was selected after preliminary trials in which higher loads

<table>
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<th>Table 1 Specification of the instrument and the film</th>
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<td><strong>Instrument</strong></td>
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<th><strong>Film (Sensor)</strong></th>
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Fig. 1 Size and arrangement of sensor points of the film
caused the clamp to slip off the tendon and disrupt the experiment. Load by a half of the value applied to the knee extensor muscle (the quadriceps muscle) was also applied to the hamstrings to restrain an excessive anterior-posterior, or the screw-home movement of the tibia relative to the femur. Choice of angle and load were made at random in order to avoid biased results. Each measurement was repeated three times. Fig. 2 shows typical examples of the contact pressure distribution.

The data collected from the three separate measurements were averaged. Data points in terms of integrated pressure, contact area and mean pressure were plotted against knee flexion angles for successive loads applied to the quadriceps. Integrated pressure means the sum total of pressure values detected at all 48 sensor points: equivalent to contact force. The patellofemoral contact areas, their shift at different loads and knee flexion angles were studied as well.

In the second type of experiment, the values of tensile forces applied to the quadriceps muscles were determined as a function of knee flexion angle, to simulate such conditions as level walking, climbing and descending stairs. Data of variations of the quadriceps force for three kinds of walking were calculated through a musculo-skeletal model of the lower limb, and angular data of the knee joint were obtained from the literature. As stated before, the maximal value applied to the quadriceps was limited to 200 Newton, whereas the corresponding value in-vivo condition is estimated as 5000 Newton during stair climbing. Thus the values of quadriceps force were reduced by 1/25 of the actual values while creating simulated walking conditions in the experiment. Then, joint contact pressure was directly measured and the same kinds of data as in the first type of experiment were obtained. After that, on the assumption that the value of patellofemoral contact force and pressure are in proportion to the value of quadriceps force, the value of 25 times measured value were estimated as joint contact pressure at in-vivo condition.

3. Calculation of Knee-Extensor Muscles Force at Walking

It is necessary to calculate the variation of knee extensor muscle (the quadriceps muscle) force for level walking, climbing and descending stairs as well as the variation of knee flexion angle to study the variation of patellofemoral contact pressure. The musculoskeletal model of a lower limb was presented\(^1\), in which thigh muscle forces were introduced by substituting angular and momentum data for the model equation. Takahama\(^2\) has reported the measurement of angular velocity and moment of the lower limb joints at various activities for the same subject. His data was applied to the present study. The variations of quadriceps force and knee flexion angle for level walking and stair ascent and
decent are shown in Fig. 3. The patterns are consistent with the EMG data, which are available in scientific literature\(^{15,16}\). At level walking and stair descent, the peak contraction of the quadriceps muscle appears at an early phase of knee flexion. Therefore, the patellofemoral contact force does not enlarge. On the other hand, at stair ascent, the peak of the quadriceps force appears in the large knee-flexion phase.

4. Results

Fig. 4 shows a typical example of integrated pressure as a function of knee flexion angle for various load values to the quadriceps, namely Q. Each curve shows a dome shape indicating the highest value at about 45° knee flexion. Fig. 5 shows a typical example of the contact area as a function of knee flexion angle for various loads on the quadriceps. Since the sensors were arranged one by one in each 20 mm\(^2\) square (19.5 mm\(^2\) exactly), the contact area was estimated by the number of weight bearing sensors times 20. Fig. 6 shows a typical example of mean contact pressure calculated by dividing integrated pressure by contact area. One should note that the mean contact pressure is almost uniform throughout the whole range of knee flexion.

Multiple regression analysis was performed by designating the quadriceps force and the knee flexion angle as explaining variables and integrated pressure as the objective variable. Thus, the value of integrated pressure \(P_1\) was expressed by a third order regression equation with respect to the value \(Q\) and knee flexion angle \(\theta\) as,

\[
P_1 = 66.7336 + 5.4398Q - 0.0883Q^2 + 0.0004Q^3 + 3.5070\theta + 0.4246\theta^2 - 0.0094\theta^3
\]

(1)

The value of multiple correlation coefficient was 0.977. Three-dimensional representation of eq. (1) is shown in Fig. 7.

The value of the contact area \(S\), was also expressed...
by the third-order equation of the other two variables, $Q$ and $\theta$, with the value of a multiple correlation coefficient at $0.987$ as,

$$S = 97.4486 + 7.7289Q - 0.0787Q^2 + 0.0001Q^3$$

$$+ 11.3634\theta + 0.0612\theta^2 - 0.0048\theta^3$$

(2)

Three-dimensional representation of eq. (2) is shown in Fig. 8.

The value of the mean contact pressure $P_m$ was also expressed with the value of multiple correlation coefficient at $0.961$ as,

$$P_m = 0.02216 + 0.00068Q - 0.00003Q^2$$

$$+ 0.0021\theta + 0.00024\theta^2$$

(3)

Coefficients of the third order variables were negligible. Three-dimensional representation of eq. (3) is shown in Fig. 9, where that the mean contact pressures change in a moderate manner can be explained.

The values of tensile forces applied to the muscles were determined as a function of knee flexion angle to simulate such conditions as level walking, climbing and descending stairs. Substituting variations of quadriceps force and knee flexion angle for the aforementioned multiple regression equation, the variation of integrated pressure for three kinds of walking can be obtained as shown in Fig. 10. Symbols such as the hollow and solid circles and the triangles indicate actually measured data under conditions that simulate variations of knee flexion and quadriceps forces for several kinds of walking. Individual data points plotted, lie almost on the calculated curves.

Variations of mean contact pressure for three kinds of walking.

![Fig. 7](image1.png) Integrated pressure expressed by a third-order curved surface as functions of quadriceps force and knee flexion angle. A black sphere indicates the actual measurement; Same in Fig. 8 and Fig. 9.

![Fig. 8](image2.png) Contact area expressed by a third-order curved surface as functions of quadriceps force and knee flexion angle.

![Fig. 9](image3.png) Mean contact pressure expressed by a third-order curved surface as functions of quadriceps force and knee flexion angle.

![Fig. 10](image4.png) Variations of integrated pressure for three kinds of walking. The value in the parentheses indicates the estimated value at in-vivo condition: 25 times measured value.

![Fig. 11](image5.png) Variations of mean contact pressure for three kinds of walking.
of walking are shown in Fig. 11. Mean pressures on the patellofemoral joints were rather uniform during the stance phase of gait, and their maximum values did not vary much among the three kinds of walking. This phenomenon may suggest physiological significance for a knee joint as well as the uniformity of mean contact pressure throughout the whole range of knee flexion.

5. Discussion

In the present study, in-vitro measurement of patellofemoral joint contact pressure was performed using a specially designed film. In the first type of experiment, seven kinds of loads were applied to the quadriceps muscle, for 0 to 120° knee flexion at 15° intervals, results of 63 data were obtained, and multiple regression analysis was performed in terms of integrated pressure, contact area and mean contact pressure respectively, by designating quadriceps force and knee flexion angle as explaining it's variables. Hara et al.17), using pressure-sensitive conductive rubber, performed in-vitro measurement of patellofemoral contact pressure. They obtained the result that when 10 kgf (98 Newton) load was applied to the quadriceps, the mean contact pressure is almost uniform throughout the whole range of knee flexion, whereas the integrated pressure and contact area indicated a dome-shape, as a function of the knee angle. Through the present study, their result was verified that to hold for various loads to the quadriceps. The morphological significance of a knee joint was suggested as well.

In the second type of experiment, measurement was performed under the conditions to simulate successive variations of the knee flexion angle and thigh muscle forces for the three kinds of walking. Since the change of the slope of contact pressure may be implicated in the Osteoarthritic lesions in the patella as well as in the magnitude of pressure, special attention was paid to the variation of contact pressure during walking. Fig. 12 shows the previously performed simulation results. In Fig. 12, contact pressure is expressed separately by the medial and lateral ones, σ₁ and σ₂, thus the sum of σ₁ and σ₂ can be comparable with the experimental results shown in Fig. 11. In Fig. 12, variables along the left vertical axis signifies the index value depending upon joint contact force and articulating surface configurations: The principal curvatures and the difference between two principal curvatures. Variable K is determined only by the value of the mechanical property of articular cartilage namely, modulus of elasticity and Poisson's ratio. Numerical value along the right vertical axis indicates the contact pressure calculated, using the following value: modulus of elasticity = 10 MPa, Poisson's ratio = 0.3. Here, the experimental results (Fig. 11) are shown to be in agreement with the simulation results (Fig. 12). Both the experimental and simulation results show that the peak values of contact pressure in each activity are almost equal, and the values of contact pressure are rather uniform during the stance phase of gait. Furthermore, not only the variation patterns but also their peak values of Fig. 11 and 12 are in approximate agreement with Seedhom et al.'s8),9). The results of Fig. 11 and 12 are very reasonable, considering the etiological theory that articular damage is not only caused by extreme pressure, but also by an abrupt change of its value9).

Fig. 13 shows the space curves overlaid on the surface, representing the Ressajus as functions of quadriceps force and knee flexion angle for three
kinds of walking. The figure indicates that the quadriceps force increases along the convexity of the surface, and therefore the value of mean contact pressure does not increase so much. As the patellofemoral articular surfaces are seemingly irregular in shape, these complicated shapes may have a fairly important role in moderating the steep variations of patellofemoral contact pressure during various kinds of walking. On the contrary, this morphological rationality of the normal knee could possibly be inclined to cause articular disease to the knee while pathological walking, during which the lower limb muscles function in fairly different ways from normal walking.

Goodfellow et al.\textsuperscript{7} attributed the surface degenerative changes to habitual disuse. It would be more accurate to say that cartilage in areas of relative disuse commonly develop mild degenerative changes. Thus, as for the pathological walking, the shift of the location of peak pressure and its magnitude should be precisely compared with those of normal walking.

Recently, the biphasic deformation characteristics of articular cartilage have become a matter of great concern\textsuperscript{13-40}. The degeneration of the cartilage is attributed to impaired nutrition from the synovial fluid which is pumped in and out of the pores of the articular surfaces as they are cyclically loaded during the joint activities\textsuperscript{18}. Also, it is a well known fact that the deformation of articular cartilage is time dependent. Unfortunately, the biphasic deformational characteristics of articular cartilage strongly suggest that static measurements or simulation analysis may not be an ideal vehicle for estimating the transient pressure distributions prevailing in-vivo.

Until then the performed measurements on contact area and/or stress in synovial joints had disadvantages such as; The magnitude of the applied load was further below than the physiological loads, and the time of load application was much longer than the duration during which physiological loads act in the joints for usual activities. The measurement in the present study is not the exception on these points. Nevertheless, the experimental results could not be exploded, because the results are in good agreement with the simulation results in which the physiological value of the load was applied. Measurements of dynamic or transient pressure distribution on the joint surfaces is the subject of the future study. Furthermore, a similar experiment to the present study should be performed with respect to such activities as rising from a chair\textsuperscript{19}, and squatting up and down.

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