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Knee Joint Forces When Rising from Kneeling Positions*

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Abstract
Knee joint forces are determined either through mathematical modeling or by in vivo measurement using an instrumented knee prosthesis. In the model studies, significant differences exist among the results and the data for high knee flexion are few. The in vivo measurement data are available for small to moderate flexion but not for high flexion yet. We created a 2D mathematical model of the lower limb incorporating several new features such as a patello-femoral mechanism, a thigh-calf contact at high knee flexion and co-contracting muscles' force ratio, then used it to determine knee joint forces arising from high knee flexions in four kneeling conditions: rising with legs in parallel, with one foot forward, with or without arm use. With arms used, the maximum values of knee joint force decreased to about 60% of those with arms not used. When rising with one foot forward, if arms are not used, the forward leg sustains a force as large as that sustained when rising with legs parallel. By comparing our modeling methodology and results with those in the literature, we determined some of the causes of the differences in the results, thereby providing creditable data especially during high flexion of the knee.

Key words: Knee Joint Force, High Flexion, Kneeling, Mathematical Model, Biomechanics

1. Introduction

Because activities requiring high flexion are necessary to all who wish to enjoy an active lifestyle, regardless of cultural background[1], the relationship between knee forces and physical activity is becoming increasingly important in understanding joint injuries and diseases, evaluating treatment outcomes, planning rehabilitation programs and designing more durable Total Knee Arthroplasty (TKA) to accommodate the needs of younger and more active patients. It is therefore important to learn what knee joint forces are created by all types of flexion from small to high.

At present, knee joint forces are determined either by direct measurement using an instrumented knee prosthesis[(2)-(6)] or through mathematical modeling, i.e. inverse dynamics[(7)-(10)]. The advent of instrumented knee prostheses has made it possible to measure knee joint force *in vivo*; however, the in vivo direct measurement data that is available concerns forces generated by small to moderate flexion. Data from instrumented prostheses about forces generated by high flexion are not yet available[(2)-(6)]. A comparison of data from studies that employed mathematical models reveals significant differences in predicted knee joint forces. The causes of these differences remain unknown; however, once the data are

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published, they serve as the current gold standard[4]. This circumstance makes it crucial to elucidate the cause of the differences, so that the most creditable figures can be determined.

To this end, we conducted an analysis of more sophysticated models. First, we created a 2D musculo-skeletal model. We used Dahlkvist et al.'s model[7] modified with new additional features including simulated muscle co-contraction and antagonistic muscle activations, a patello-femoral mechanism[11]-[13], and the effect of the thigh-calf contact force[14]. The force ratio between the co-contracting muscles was determined from each muscle's physiological cross-sectional area (PCSA)[15]-[18]. Next, to verify the validity of our model, we used it to calculate knee joint forces during the small to middle range of knee flexion and compared our results with the available in vivo data. We then calculated knee joint forces during high knee flexion activities such as rising from squatting and kneeling, and compared the results with predictions reported in the literature. When significant differences were found between our results and those in the literature, we compared the modeling methodologies to determine the possible cause of the differences.

2. Mathematical model and experiment

2.1 Mathematical model

To measure the kinematics and kinetics required to kneel, it would be necessary to look at all the different ways to move from a standing to a kneeling position and back[1]. However, formularizing all the ways is difficult. Instead, we decided to assess forces on the knee joint when rising from a kneeling position in four different ways as shown in Fig.1. To avoid confusion between the expressions "kneeling" and "deep squatting", we will use, "kneeling" to refer to the act of sitting on a floor with one or both knees touching the floor surface as shown in Fig.1.

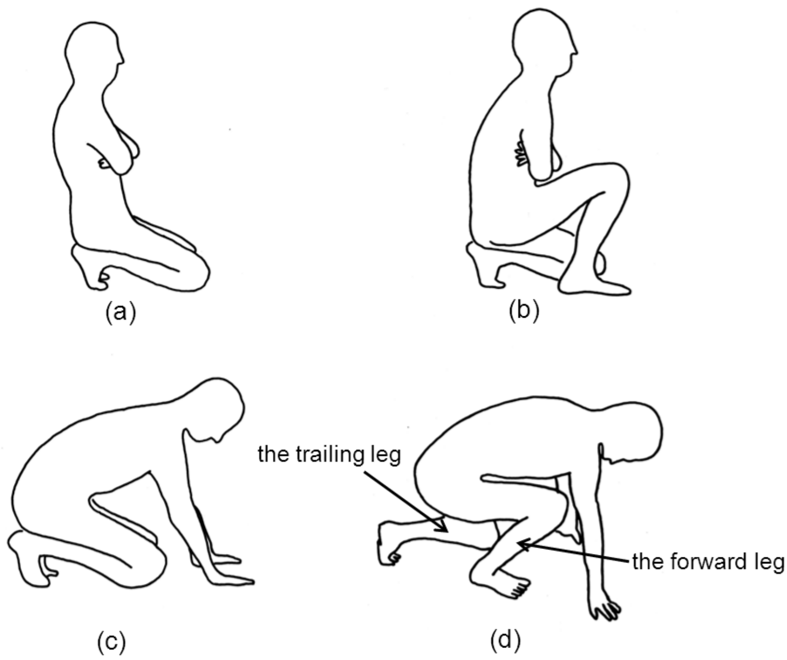

Fig.1 Ways of rising from various kneeling conditions

(a) rising with legs parallel (arms not used), (b) rising with one foot forward (arms not used), (c) rising with legs parallel (arms used), (d) rising with one foot forward (arms used)

Our 2D mathematical model is composed of three segments: upper leg, lower leg and foot. The muscle groups incorporated into our model are shown in Fig.2.

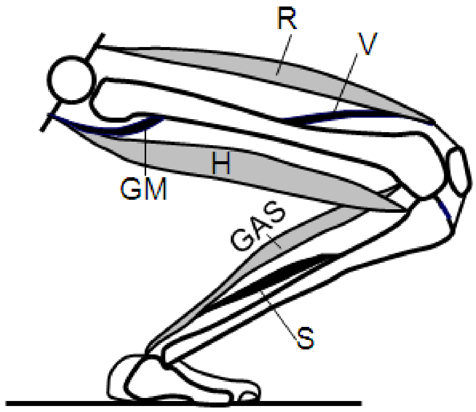

Fig.2 Illustration of the muscles included in the model.

They are the gluteal muscles (GM); quadriceps(Q), including rectus femoris (R) and the vasti (V); hamstrings (H), and calf muscles, including the gastrocnemius (GAS) and soleus (S). Hereafter the same symbols in Italic will be the equation variables representing the forces exerted by the respective muscles, thus *GM* stands for the forces exerted by GM and so on.

The forces acting on the hip, knee and ankle joints are illustrated in Fig.3.

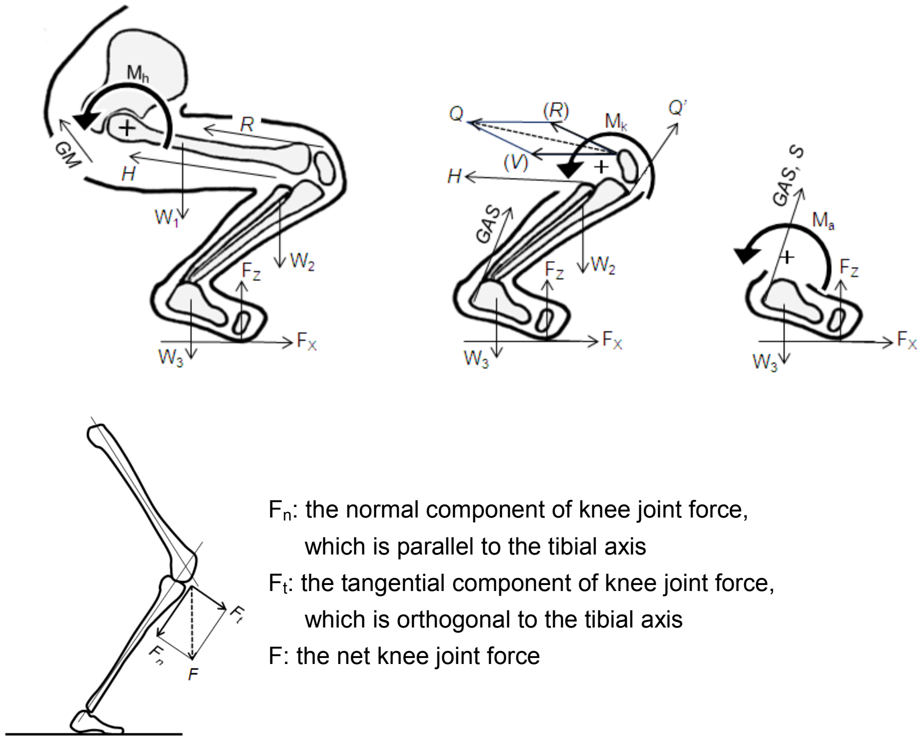

Fig.3 Two dimensional mathematical models for the moments around the hip (upper left), knee (upper center) and ankle (upper right) joints, and the relationship between the net knee joint force and their normal and tangential components (lower).

The variables representing the tensile force generated by the muscles and tendons are

H: hamstrings, *GM*: gluteus, *R*: rectus femoris, *V*: the vasti, *Q*: quadriceps (the vector sum of *R* and *V*), *Q'*: patella tendon, *GAS*: gastrocnemius, *S*: soleus

The variables representing the force acting on the knee joint as illustrated in Fig.3 (defined as the force acting on the tibia from the femur) are

F_n: the normal component of the knee joint force, which is parallel to the tibial axis

F_t: the tangential component of the knee joint force, which is orthogonal to the tibial axis

F: the net knee joint force (the vector sum of F_n and F_t)

The variables representing external forces are

$W_{1,2,3}$: the gravity force acting on the thigh, shank and foot, respectively

F_Z, F_X: the normal and tangential components of the floor reacting force respectively

The variables representing moment are

M_h: about the hip joint

M_k: about the knee joint

M_a: about the ankle joint

We have incorporated three new features into the model. First, we introduce the ratio between the force in the the quadriceps, Q, and the force in the patella tendon, Q' as a function of knee flexion angle, as described in the literature(11)(13). We set the directions of pull on the quadriceps, Q and the patellar tendon, Q' as a function of knee angle based on the patello-femoral mechanism(12)(13). We also introduce the effect of the thigh-calf contact force(14) into the model. When knee angle exceeds 125°, we take the thigh-calf contact force, P and the position (a presumed center of contact pressure, which is determined as the distance from the knee along the tibial axis in 2D model), d of this contact force(14). Per our definition of kneeling above, we incorporate the contact force between the knee and the floor, N, exerted when the knee is touching the floor, where N_Z and N_X represent the normal and tangential components of the floor reacting force on the knee respectively.

In the following equations, the symbols a, b and c stand for the lengths of the moment arm about the hip, the knee and the ankle joints respectively. Thus a_{FZ} means the moment arm of F_z about the hip joint, and b_p means the moment arm of thigh-calf contact force P about the knee joint; the equivalent of variable d above.

Moment M_h created by external forces is expressed as,

$$M_h = F_z a_{Fz} + F_x a_{Fx} - W_1 a_{W_1} - W_2 a_{W_2} - W_3 a_{W_3} \tag{1}$$

Moment M_h created by muscle forces is expressed as,

$$M_h = GMa_{GM} + Ha_H - Ra_R + N_Z a_{N_Z} + N_X a_{N_X} \tag{2}$$

Since the values from equations (1) and (2) must be equivalent to each other, we can eliminate M_h, which gives the following equation,

$$F_Z a_{F_Z} + F_X a_{F_X} - W_1 a_{W_1} - W_2 a_{W_2} - W_3 a_{W_3} - GMa_{GM} - Ha_H + Ra_R - N_Z a_{N_Z} - N_X a_{N_X} = 0 \tag{3}$$

The equation for the knee joint is,

$$-F_Z b_{Fz} + F_X b_{F_X} + W_2 b_{W_2} + W_3 b_{W_3} - GASb_{GAS} + Q' b_Q - Hb_H - N_Z b_{N_Z} - N_X b_{N_X} + Pb_P = 0 \tag{4}$$

and for the ankle joint as,

$$F_Z c_{Fz} + F_X c_{F_X} - W_3 c_{W_3} - (GAS + S) C_{GAS} = 0 \tag{5}$$

The three equations contain six variables, i.e. six muscle muscle forces: GM, H, R, GAS, Q' and S. To solve this statically-indeterminate equation, it is necessary to decrease the number of variables from six to three. To do this we will assume that if the EMG information indicates that more than one muscle group is active at any particular time, the moment of external and gravitational forces about any joint is shared simultaneously by the muscles involved in resisting this moment. Our three specific assumptions are:

1. A moment that tends to extend the hip is shared by the gluteal muscles GM and the hamstrings H, if both are active. Since it is known that the muscle forces are in proportion to their physiological cross-sectional areas (PCSAs)(18), we assume that the force ratio between GM and H would be $GM : H$ = 20.0 : 60.2, according to the

literature[16].

2. A moment that tends to flex the ankle dorsally is shared by the soleus S and the gastrocnemius GAS. Using the assumption about force to area proportion in 1. above, we assume that $S : GAS = 99.1 : 247.6$, according to the literature[15].
3. When the four muscles in the quadriceps are active, since the PCSAs of the individual muscles are almost equal[15][16], the forces exerted by each one of these muscles would be one-quarter of the total quadriceps force. Thus, the rectus femoris R and the vasti V work simultaneously with a strength ratio of $R : V = 3 : 1$. By considering the force triangle composed of R, V and Q, we can calculate the force Q (see Fig.3).

From equations (1) through (5) and the three conditions above, we can introduce the muscle forces acting on the hip, knee and ankle joints respectively. We can then use the values for muscle forces around the knee joint to introduce the forces acting on the knee joint, F_n, F_t and F, as follows,

$$F_n = Q' + GAS + H\cos\theta \tag{6}$$

$$F_t = H\sin\theta \tag{7}$$

$$F = \sqrt{(F_n)^2 + (F_t)^2} \tag{8}$$

where, θ is knee flexion angle.

2.2 Experiment

Ten healthy males (age 26±4years, height 175.1±5.5 cm, and weight 76.6±21.1 kg) and five healthy females (25±3 years, 160.1±7.1 cm, 47.7±6.2 kg) participated in the measurement experiment. Before the experiment, we obtained the approval of the Saga University ethics committee and informed consent from all subjects. To obtain the physical parameters, the length of each subject's upper leg, lower leg and foot was measured directly (The upper leg length was determined by the distance between the lateral illiac crest and the femoral lateral crest, the lower leg length was between the fibulal lateral crest and the malleous of foot, and the foot length was between the malleous and the lateral end of the metatarsus arc. The above mentioned anatomical landmarks were identified by palpation.)

The mass of each segment and the location of its center of gravity were determined by referring to the literature [17], where such regression equations are shown so that we can estimate the above mentioned values by inputting the subject's segment length, total body weight, height and gender into the equations. The equations had been statistically introduced using a large number of measurement data obtained from the cutted pieces of dead bodies.

The lengths of the moment arm about the hip, the knee and the ankle joints, a, b and c were respectively determined as follows. First the distance between the presumed center of rotation of a joint and the insertion site presumed same for the muscles around that joint was determined. Then the component of the above distance normal to the muscles' pull direction at the insertion, again presumed the same for the muscles concern, was determined as the moment arm around that joint. The joint rotation centers and the muscles insertion sites were identified by palpation.

First, three subjects who have similar builds performed activities requiring small/middle knee flexion: standing on one leg, level walking, rising from a chair, ascending and descending stairs, and knee bending. Ground reaction force data and the angles of each joint during the motions were collected by a force plate walkway (Model BP400600, Amti Co., USA) and a video recording system (Vicon Motion Systems, Vicon Co., UK) respectively. Twenty force measuring plates (50cm x 60cm) were installed along the walkway, and further four plates were installed on the midway. Thus, a subject was able to place his/her right and left hands, knees and feet on six individual plates respectively. The subjects repeated each activity three times, and the three sets of data were averaged. The muscle and joint forces were then calculated through our 2D mathematical model, and compared with in vivo data from the literature.

Next, the same subjects performed an activity requiring high knee flexion: rising from a squatting position with legs in parallel. As before, each subject performed the activity three times; the muscle and joint forces were calculated, and compared with mathematical predictions reported in the literature(17).

Finally, all fifteen subjects performed the four different rising motions depicted in Fig.1. Each subject repeated each motion three times. The muscle and joint forces for each kneeling condition were calculated as well.

3. Results

We used the data we obtained to make the following comparisons. The normal and tangential components of knee joint forces for small/middle knee flexion motions made by one subject were compared with the in vivo data(5) for the same motions (Fig.4). In Fig.4, the unit [BW] means the values of forces divided by the subject’s body weight. The right graphs are modifications of those by Kutzner et al.(5) used with permission. The tangential component F_x (anterio-posterior) and F_y (medio-lateral) are always small or negligible in Kutzner et al.'s. Thus, their net resultant force F can be considered to correspond to the normal component F_z (superior-inferior, not shown) as well as to F_n of our study.

Next, the maximum values of the net resultant force acting on the knee joint of one leg during small/middle knee flexion activities were compared with the in vivo data(3)-(6) (Table 1). The maximum calculated values of the net resultant force acting on the knee joints when rising from a squatting position were compared with predictions reported in the literature(7)(9)(10) (Table 1). The data of this study in Table 1 are from the three subjects mentioned in the experiment section.

Thirdly, knee joint forces from all fifteen subjects when rising from a kneeling position were graphed as follows: with legs in parallel (Fig.5) and with one foot forward (Fig.6). In Figs 5 and 6, because the time when rising began and the duration from start to finish varied from subject to subject, the mean and standard deviation curves for the variations in knee joint forces are drawn as a function of knee angle. In Fig.6, the graphs of variation in forces on the forward leg and that on the trailing leg are shown separately in order avoid the impression that both legs share the forces over the same knee angle. In Fig.6, for the sake of clarity, the standard deviation curves are drawn in terms of knee angles or joint forces, depending on whether an inclination in the graph is steep or not.

Finally, the numerical values of the maximum knee joint forces acting on a single knee were tabulated, as were the knee angles at which those forces are exerted when rising from various kneeling conditions (Table 2). Note that the maximum values in Table 2 differ from the maximum values of the mean curves in Figs 5 and 6 because the mean curves were created from individual curves, per Acker et al.(19). In Figs 5 and 6, an extreme value on the mean curve is the mean of all the subject curves at the given angles of the knee, while the values in Table 2 are the means of the maximum values of each individual subject curve, which did not necessarily correspond to the same angle on each curve.

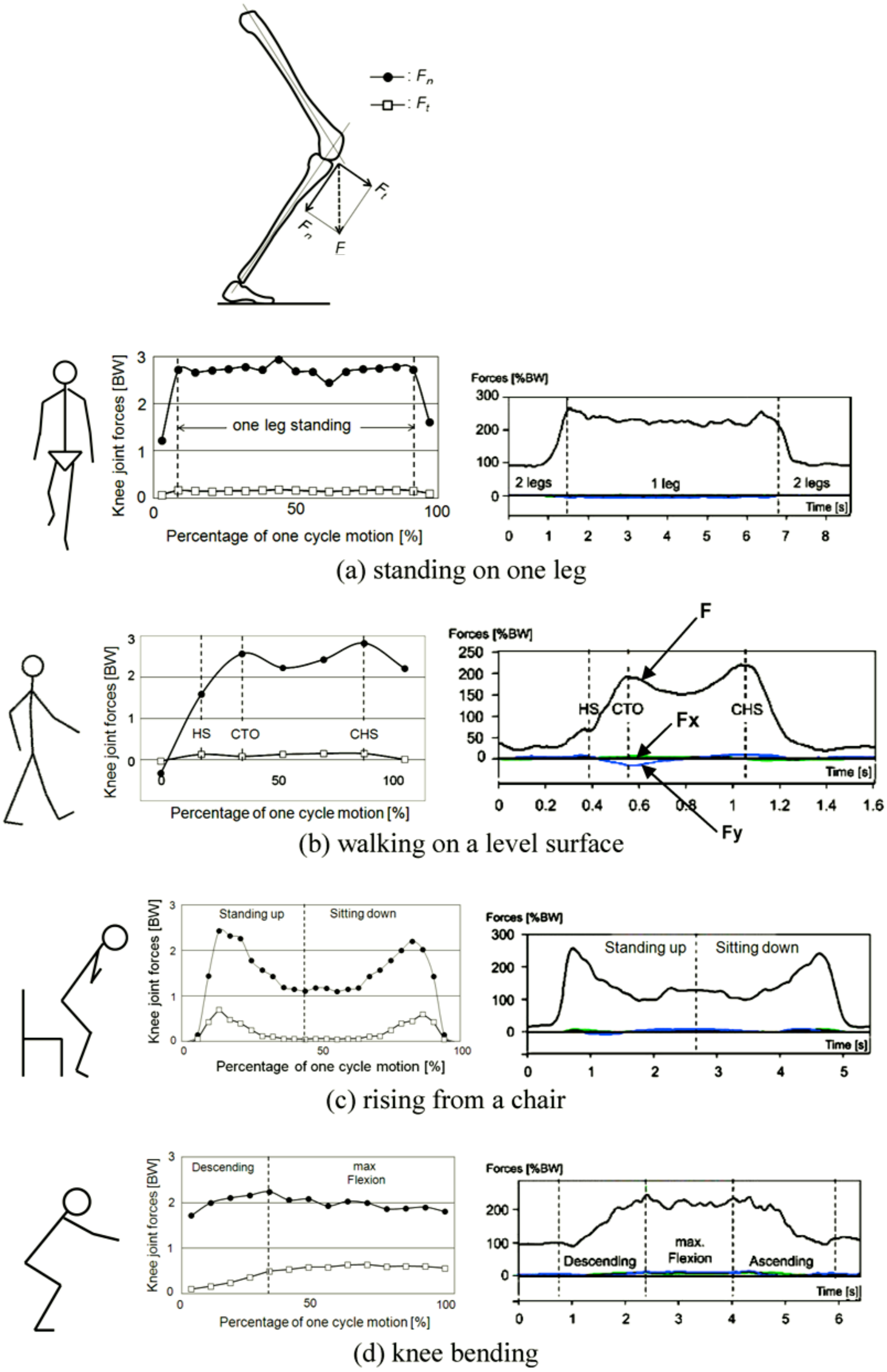

Fig.4 Comparison between knee joint forces from this study (left) and those from the in-vivo measurement (right: modefied those by Kutzner et al.[(5)]). (HS: heel strike, CTO: contralateral toe off, CHS: contralateral heel strike)

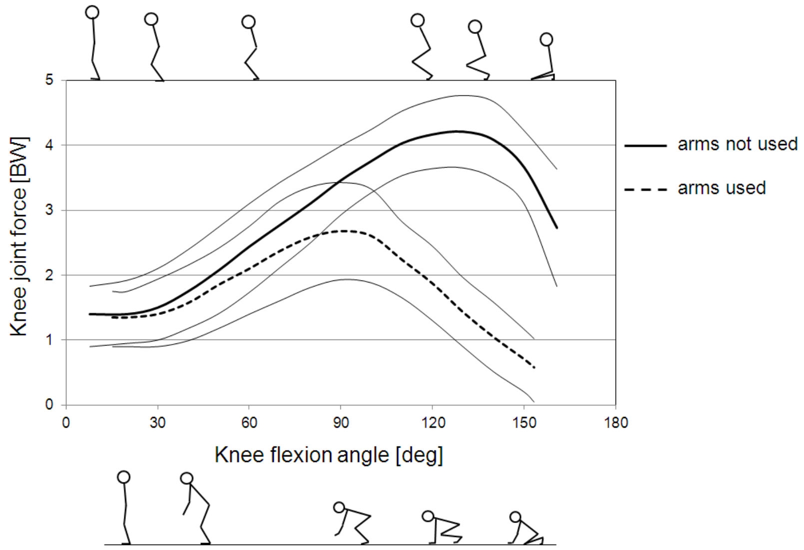

Fig.5 Curves of the mean and standard deviations in knee joint forces as a function of knee angle when rising with legs in parallel.

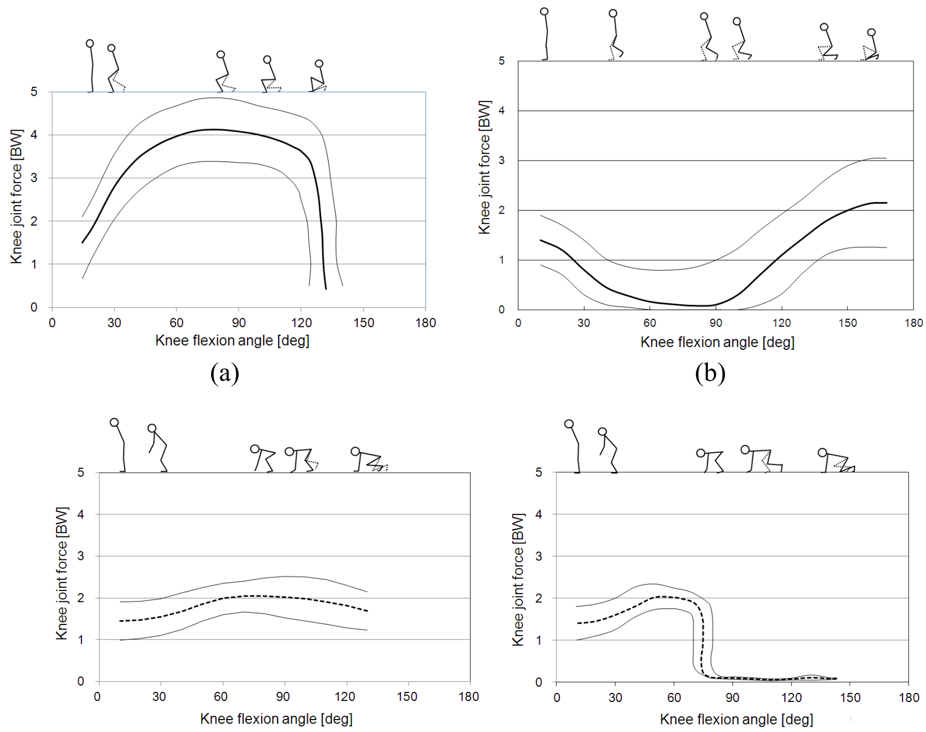

Fig.6 Curves of the mean and standard deviations in knee joint forces as a function of knee angle when rising with one foot forward.
(a) the forward leg with arms not used, (b) the trailing leg with arms not used,
(c) the forward leg with arms used, (d) the trailing leg with arms used.

Table 1 Comparison of knee joint forces derived from various approaches.

Approach	Authors		level walking	stair descent	stair ascent	knee bend	rising from squat	rising from kneeling
in vivo measurement	Taylor	(2001)	2.8	3.1	2.8	—	—	—
〃	D'Lima	(2007)	2.3	—	3	2.1†	—	—
〃	Heinlein	(2009)	2.76	3.52	3.06	—	—	—
〃	Kutzner	(2010)	2.61	3.46	3.16	2.53	—	—
modeling	Dahlkvist	(1982)	—	—	—	—	4.6～5.2	—
〃	Zheng	(1998)	—	—	—	—	4.3	—
〃	Smith	(2008)	—	—	—	—	3.73(0.56)††	—
〃	Nagura	(2006)	—	—	—	—	—	7.3(1.9)††
〃	This study		2.5～2.8	3.0～4.1	3.1～3.4	2.2～2.3	3.5～4.1	4.1～4.5

†They described "kne bend" as "squatting" ††mean (SD)

Table 2 Knee angle when knee joint forces become maximum and the maximum knee joint forces

	Knee angle when knee joint force is max [°] mean (SD)	Maximum knee joint force [BW] mean (SD)
Rising with legs parallel		
Arms not used	125.5(10.7)	4.3(0.5)
Arms used	90.1(11.3)	2.8(0.8)
Rising with one foot forward		
Arms not used		
the forward leg	69.1(10.8)	4.1(0.9)
the trailing leg	150.4(10.2)	2.2(0.5)
Arms used		
the forward leg	88.1(10.8)	2.1(0.5)
the trailing leg	57.8(10.4)	2.3(0.6)

4. Discussion

Although studies on knee joint kinetics and kinematics are extensive, there is still little data on high knee flexion. To obtain data, we created a mathematical model of the lower limb and used it to analyze knee joint forces when rising from a kneeling position. We incorporated innovative features into the model: muscle co-contraction and antagonistic muscle activation were used, and a patello-femoral mechanism was taken into account; that is, the ratio between Q and Q', and the directions of pull on Q and Q' were determined as a function of knee angle respectively(12)(13).

Despite the current notion that large variations exist among the reported knee joint forces, our results for small/middle knee flexion did not differ significantly from the in vivo data (Fig.4, Table 1). The exceptions were in the tangential components found in rising from a chair (Fig.4 (c)) and knee bending (Fig.4 (d)). In this study, when the knee angle was close to 90°, the tangential component values, i.e. the anterio-posterior (AP) forces were about 0.7 times body weight (0.7 [BW]). However, in the in vivo data, the same forces are always small or negligible(5). In our opinion, the AP forces cannot be negligible; if they were, the prosthetic post would remain undamaged or show no wear, which does not happen. One possible explanation for the in vivo data is that interaction with soft tissues and/or friction between the articulating surfaces reduced the AP forces. Another is that when data were taken from the instrumented prostheses, knee flexion angles were not large enough to create strong AP forces. Yet it should be noted that in this study, neither the net resultant forces nor the normal components differed significantly from those in the in vivo data.

Our results for rising from a squatting position did not differ significantly from other data in the literature[9][10] with the exception of Dahlkvist et al.'s[7] (Table 1). Zheng et al.[10] produced their results from their detailed model which incorporated micro and macro structures of the knee. Smith et al.'s 2D model was rather simple, but they applied a unique scaling method to their analysis[9]. We added three new features to Dahlkvist et al.'s mathematical model. By removing these three features one at a time and recalculating the forces each time, we found that the main cause of the difference was the method used to set the muscles' force ratio. Dahlkvist et al. set the ratio on the assumption that the moment at any joint is shared equally by the muscles involved. We set the ratio according to each muscle's PCSA. When we set it using Dahlkvist et al.'s method, then the force values became equivalent to theirs. As our method more closely simulates actual physiological conditions in the knee than theirs does, we may conclude that their predictions were too large.

From Figs 5 and 6 and Table 2, we know how the knee joint forces differ, depending on the alignment of the leg and/or on whether the arms are used or not. Overall, with arms used, the maximum values of knee joint force decreased to about 60% of those with arms not used (Fig.5, Fig.6, Table 2). When rising with one foot forward, if arms are not used, the forward leg sustains a force as large as that sustained when rising with legs parallel.

In conclusion, we may assess the maximum knee joint force when rising from a kneeling position as 4.5 [BW]. On the other hand, Nagura et al.[8] using a simple 2D model of the knee joint, obtained significantly higher value than ours. They reported a value as high as 7.3 ± 1.9 [BW] at 146.3° of knee flexion when rising from a full squat with arms not used. Here a "full squat" in their study corresponds to a "rising from kneeling with legs parallel" in this study (Fig.1 (b)) because knees were obviously touching on a floor at the initiation of rising in their study. There is a question about the angle at which they claim maximum force was exerted on the knee. Irrespective of a subject's corpulence, the maximum knee angle during active flexion does not exceed 140°, even if a *follow-through* is included[20]. At an angle as large as 146°, the tibio-femoral surfaces do not maintain a complete articulation (subluxiate)[21] and therefore the knee joint does not sustain such a force as large at the one they reported. Another question is about reduction of each of the extensor and flexor muscle groups to only one string respectively. They determined the force ratio between the extensor group and the flexor group on the basis of each group's EMG data. EMG data can be used to predict an individual muscle's force but it is doubtful whether the same method could be used to predict the force of a group of many muscles with different lengths and insertions. In view of these weaknesses in their study, we conclude their predictions of the maximum knee joint force are not realistic and are impractically large.

We have found various possible reasons for the large variations among the reported predictions obtained from different mathematical models of the knee. Yet by factoring in each mathematical reason and recalculating the forces each time, we found many of the reasons did not have decisive influence on the results in this study.

Although the limitations of 2D models[4] may be cited as another of the reasons for the large variations, analysis by 2D model can yield useful information about knee joint forces. In sitting-to-standing actions, one major motion of the knee joint is flexion/extension. Musculoskeletal anatomy reveals that the insertions of lower limb muscles involved in this motion and their directions of pull are virtually on the same sagittal plane. Moreover, changes in the muscle lengths involved in non-sagittal motions are relatively small.

By comparing the knee joint force exerted with thigh-calf contact with the knee joint force exerted without it, P in equation (4), we found thigh-calf contact had little effect on the decrease of knee joint force at high flexion. Knee joint force did decrease as knee angle increased to high flexion because of a link mechanism between the thigh and shank, not because of thigh-calf contact. The value of the thigh-calf contact force itself is reported to be less than 0.5 [BW][14], and its influence on the knee joint force is even smaller than that value.

In predicting knee joint force, we found one of the most influential factors was the method for determining the co-contracting muscles' force ratio. Many optimization techniques have been reported to address this indeterminate problem[2]. Yet, the choice of optimization criteria depends on researchers and its validation is indirect. Brand et al.[15] have already mentioned that the optimization criteria have only a small influence on the calculated joint contact forces. Besides optimization techniques, a substantial criterion is needed for determining the muscles' force ratio. Kumamoto et al.[22] claimed the bi-articular muscle function could fill this role because this muscle's sole function is to control the force direction at the end point of a double link system, for example a hand for an arm, a foot for a leg. If their claim is correct, future models should incorporate Kumamoto et al.'s idea in order to introduce further accurate knee joint forces.

Although various problems still remain in model analyses, by refining our model and by excluding some extreme values from those reported in the literature, we feel the values we predicted for knee joint force are creditable. Furthermore the results concern the influence of the legs' alignment and the arms' assistance on the joint force and therefore should be of use in rehabilitation and the design of TKA.

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