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EFFECT OF RELATIVE KNEE POSITION ON INTERNAL MECHANICAL LOADING DURING SQUATTING

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ABSTRACT

Background: There is a widespread notion among physical education teachers, physical therapists and orthopedists that, during any type of squatting, the knee should not be brought forward too much in relation to the tip of the foot, so as to reduce the mechanical loading on the knee. However, there is little quantitative evidence to corroborate this notion. Objective: To estimate the forces and torque on the knee joint in healthy individuals during free squatting exercises using weights performed in two different ways: a) knee not going beyond a vertical line going through the toes; b) knee going beyond this vertical line. Method: Three-dimensional analysis using video cameras and a force platform was performed on squatting movements performed by ten healthy young adults. Fifteen repetitions of each of the two squatting conditions were performed by each subject on the force plate. The forces and joint torque at the ankle, knee and hip were calculated using an inverse dynamic procedure. Results: The results obtained showed that the mean peak torque on the knee was around 38 ± 31% greater, and the mean patellofemoral force was around 28 ± 27% greater, when the knee went beyond the tip of the foot, than when it did not. Conclusions: These results demonstrate that, when the knee does not go beyond the line of the foot, the patellofemoral compression force is less, which leads to lower mechanical demand on this joint.

Key words: knee, injury, inverse dynamics, three-dimensional kinematics.

INTRODUÇÃO

Squatting, performed in different ways, is an important exercise, commonly utilized in training and rehabilitation, and has been the goal of several studies[1-8]. However, as with any other exercise, if performed incorrectly or excessively, squatting may result in injuries on the musculoskeletal system, being the main factor the magnitude of the patellofemoral force, that contributes to the degeneration of the patellar cartilage and femoral surface, which may result in pathologies such as, for example, patellar chondromalacy and osteoarthritis[2]. Understanding how the compressive patellofemoral force behaves during the complete squatting movement is of key importance to delineate the conduct of physical therapists and physical education teachers while prescribing this kind of exercise.

It is known that the patellofemoral force magnitude is affected by how the exercise is performed[5] and that an increase on the knee flexion will increase patellofemoral force[9,3-5,7]. Aligned with these findings there is a widespread concept among health professionals that, during squatting, knee position should not go beyond the position of tip of the foot at the antero-posterior direction.

Only two studies on the literature investigated specifically this question. With a computer simulation, Abelbeck[1] calculated the hypothetical torque of a man (weight 110 kg) squatting on a machine for linear movement squatting ('Smith machine') with a 100 kg load. His analyses were bi-dimensional (only on the sagittal plane) and static (did not include the acceleration terms, nor the ground reaction force variation). Abelbeck[1] found that the peak torque on the knee was 50% greater when the relative knee position varied (going beyond foot line). Fry, Smith and Schilling[6] developed a study in which individuals performed free squatting with the load of body weight. This analysis was also bi-dimensional and static. They found that the peak torque on the knee was around 30% greater when the knee went beyond the tip of the foot. Although both studies have generated relevant information, a more accurate determination (such as a tri-dimensional dynamic analysis) of the knee articulation during squatting can be done. Additionally, it is still unknown how patellofemoral force is affected by this question. Determination of these two mechanical variables will allow a greater understanding of the effect of knee positioning in relation to the foot on the mechanical load on the knee, which will contribute to elucidate how the mechanical component is associated to the possible injury on the knee in squatting practitioners.

In this sense, the objective of this work was to investigate the effect of the placement of the knee in relation to the foot on the knee torque and patellofemoral force during squatting, using a tri-dimensional dynamic analyses of the squatting.

MATERIAL AND METHODS

Ten individuals participated on the present study (seven men, and three women) with a minimum experience of three years in free squatting. None of the participants did squatting with competition finality, but as part of his resisted exercises routine. The average time in which these individuals practiced the analyzed movement was five years (minimum three, and maximum ten years). Average height (± standard deviation) of the individuals was of 171 ± 10 cm, average weight of 68 ± 12 kg and average age was 25 ± 5 years. None of the participants reported any kind of injury on the lower limbs and they all performed the experiment only after signing a consent term according to the local Ethics Committee of the Physical and Sports Education School of the São Paulo University (protocol # 42).

The subjects performed loaded squatting on two different conditions: a) knee not trespassing the vertical line that passes through the toes of the feet (NT); b) knee trespassing this vertical line (T). The load for each individual was adjusted to equal to 40% of his body weight. The execution order was random, in such a way that half the individuals began with the NT (not trespassing) squatting and the other half with the T (trespassing) squatting. Thus, in each condition, the subject ought to perform the squatting movement 15 times. Feet positioning was not imposed, therefore the subjects adopted the most comfortable position for them. The execution rhythm was controlled with a metronome with a 40 beats per minute frequency; each beat delimited the movement's extremes (maximum knee flexion and maximum knee extension when the subject stood up). All participants reported that tiredness or fatigue was negligible.

For the squatting tri-dimensional kinematic analysis were used five digital cameras (four JVC 9800 and one JVC DRV800U, JVC Inc.), all with acquisition frequency of 60 Hz. Reflexive markers were placed in anatomic prominences at the following body localizations[10]: left and right anterior-superior and posterior-superior iliac spines, major throcanter, lateral femoral epicondilus and medial femoral epicondilus, apex of the head of the fibula, tibial tuberosity, distal apices of the lateral and medial malleolus, calcaneus, head of the fifth metatarsus, head of the second metatarsus, head of the first metatarsus of the lower right limb, and right lateral portion of the trunk, at the xiphoid process height, during task performance. To minimize measuring errors of the kinematic data, it was used the anatomical system calibration technique (CAST) proposed by Cappozzo et al.[10]. For such, two clusters were used, composed by one rigid plate with four marks on it, fixed on the segment, one on the leg and one on the thigh.

Participants performed squatting over a force platform (AMTI DAS-6, AMTI) that measured the force and torque components made on the ground by the subject during movement. Escamilla et al.[4] showed that for adults without lesion and movement-experienced, the analysis is similar between the limbs, therefore, only the right limb of the subject had been in contact with the platform during the whole period of data collecting. For acquisition of the force platforms data was utilized a computer with a 16 bits analogical/digital data acquisition plate (PCI 6033, National Instruments) and acquisition frequency of 60 Hz.

Marks digitalization was made on the APAS software (Ariel Inc.) and the tri-dimensional reconstruction was made using the Direct Linear Transformation algorithm implemented in a computational routine on the Matlab environment (Mathworks Inc.) For smoothing of the kinematic data, quintic splines were adjusted to the data by using the '*spaps*' function of Matlab's toolbox Spline. Force platform data were smoothed by a fourth order Butterworth filter, low passing with a cut-off frequency of 20 Hz.

It was determined axis and articulation planes position (anatomically based) as described by Cappozzo et al.[10] so that the torques and forces were represented on the axis of the joint itself. The coordinates of the hip's joint center were determined by numerical optimization and expressed through the pelvic coordinate system, analogously to Piazza[11], being used the method proposed by Bell[12] as initial optimization estimative. Knee joint center was considered as the midpoint between femur epicondilus, and the ankle joint center as being the midpoint between the lateral and medial tibial malleolus.

Internal torques and forces were determined by means of the inverse dynamics method that considered gravitational force on the bar and on the segments, ground reaction force, and the segments acceleration. Positive torques indicate that the torque is extensor, and negative torques that it is flexor. Inertial properties of the segment were calculated accordingly to the adjustments proposed by Leva[13] on the anthropometric model of Zatsiorsky et al.[14]. Hip joint torque was expressed on the anatomical basis of the knee, and not on the anatomical basis of the hip, which was not reconstructed during dynamical trial.

Lever arm of the quadriceps muscle (LM, in meters) in function of the knee angle (α, in degrees) was given by the equation below, which was obtained by a work data adjustment of the experimental work by Van Eijden et al.[15]. The determination coefficient of this adjustment was $R^2 = 0.98$ $(p< 0.0001)$.

$$LM(\alpha) = 7.69E-8\alpha^3 - 1.25E-5\alpha^2 + 2.70E-4\alpha + 4.58E-2$$

Thus, quadriceps force (QF) is determined by the following way:

$$QF = T_{EXT/FLEX}/LM$$

Whereas = $T_{EXT/FLEX}$ is the extensor/flexor component of the torque on the knee.

During knee joint angle changes, the patella perform a movement much more complex than that of a simple pulley, making the distance from the patella to the knee joint center not constant, varying the relation between QF (given by the equation above) and the patellofemoral force (PFF). This relation between QF and PFF may be determined from a data adjustment of the experimental work of Van Eijden et al.[15]. Equation 3 represents the equation obtained from the adjustment of those data. Determination coefficient of this adjustment was foi $R^2 = 0.98 (p < 0.0001)$.

$$K(a)=$$
$$(1.33E-8\alpha^4 - 2.96E-6\alpha^3 + 1.37E-4\alpha^2 + 8.07E-3\alpha + 1.55E-4)$$

Thus, PFF may be obtained by:

$$\mathrm{PFF}(\alpha) = \mathrm{QF}(\alpha) * \mathrm{K}(\alpha)$$

The studied forces (quadriceps and patellofemoral forces) were normalized by the subject's body mass (BM), while the torques (of the ankle, knee, and hip) were normalized by body mass times subject's height (BM*Height).

Pared t-test was used with a significance level of 0.05 to identify differences between the T and NT conditions, and the studied variables were the following: ankle and knee angle; relative knee position (RKP), defined as the projection on the horizontal axis of the vector sagittal plane, determined by the position of the marker localized on the second metatarsus minus the position of the knee joint center; ankle torque, knee and hip, QF and PFF. All the cited variables, with exception of RKP, were studied on the instant that the compressive force was maximal.

RESULTS

All the participants could perform the proposed task since the relative knee position (RKP – the position on the sagittal plane of the mark placed on the second metatarsus minus the position on the sagittal plane of the knee joint center) was different between the conditions ($p < 0.001$). The participants advanced the knee in average 11 ± 5 cm more at the T condition when compared to the NT condition.

Figure 1 represents the behavior of the ankle, knee and trunk angles of a participant in both conditions, considering that the value zero corresponds to erect position (neutral) of the individual. Average angle between the participants, on the instant of maximal knee flexion, was 92 ± 15º at the NT condition and 78 ± 18º at the T condition ($p < 0.001$) for the knee angle, and 54 ± 11º at NT and 70 ± 12º at T ($p = 0.0011$) for the trunk angle. The average ankle angle between the participants was 87 ± 6º at the NT condition and 81 ± 10º at the T condition, which revealed a significant difference ($p = 0.39$) between both conditions.

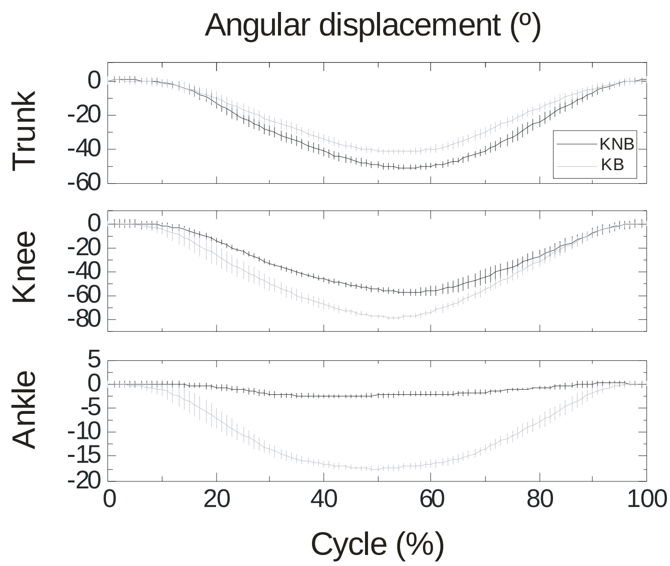

Figure 1. Angular displacement of ankle, knee and trunk of an individual at both conditions of trespassing (T) and not trespassing (NT) the knee.

The peak torques of the ankle, knee and hip joints were different between conditions (respectively, $p = 0.0016$; $p = 0.0011$ and $p = 0.046$), presenting the following normalized values for ankle, knee, and hip respectively: 0.007 ± 0.027; 0.16 ± 0.02; 0.06 ± 0.03 at NT and 0.04 ± 0.04; 0.22 ± 0.05; 0.12 ± 0.06 at T condition. In Figure 2 it is possible to observe the behavior of the average joint torques of an individual. Negative values mean flexor external torques and positive values mean extensor external torques. It is noted that ankle torque during NT condition was kept close to zero, and that, at T, this torque was more plantar flexor. The knee torque, on both conditions, was extensor all the time, as well as the hip torque.

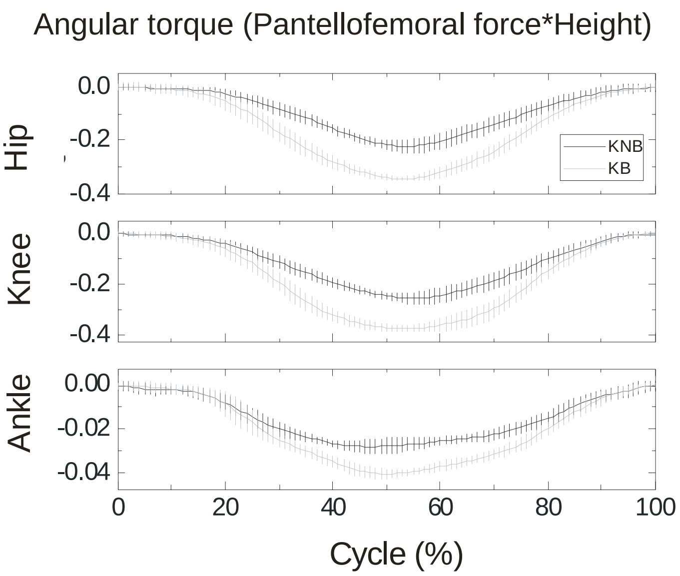

Figure 2. Joint torques for a subject on conditions of not trespassing (NT) and trespassing (T) the knee.

Figure 3 represents the behavior of the PFF of an individual in both conditions. The PFF shows a pattern similar to the one found for QF and at T condition. The PFF was greatest at NT condition.

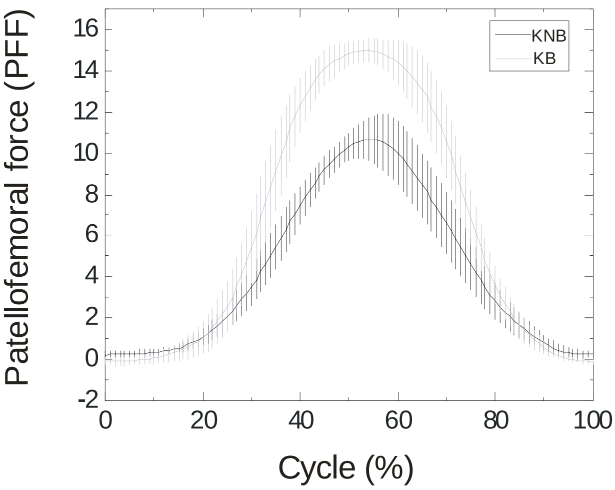

Figure 3. Average patellofemoral force (PFF) and standard deviation of the individual at both conditions, not trespassing (NT) and trespassing (T) the knee.

The compressive force was also normalized by the subjects weight. In this case, the compressive force was in average 28 ± 27% greater at T condition (p= 0.01), being that 12 ± 3 CM at NT and 15 ± 3 CM at T.

DISCUSSION

Our results showed that both squatting forms regarding the relative knee position presented distinct kinetic and kinematic patterns at the three lower limb joints and not only at the knee joint. In order for the participants to perform the squatting with the knee trespassing the horizontal line of the foot, the execution technique had to be altered, as demonstrated by the variation of the joint angles of knee and trunk. At T condition (where the individual trespass the line of the foot with the knee), knee angle was smaller, while the participant inclined less the trunk forward to keep the projection of the center of mass inside the base of support. At NT condition, in which the knee angle was greatest, the individual's hip had to be kept more backwards so that the knee did not trespass the horizontal line of the tip of the foot. To avoid falling back, the participant inclined his trunk forward, keeping, thus, the vertical projection of the center of mass inside the feet base of support, which promoted participant's balance during movement.

Excessive patello-femoral forces (PFF) may contribute to knee injuries such as chondromalacy and osteoarthritis. PFF was greater (around 49 ± 34%) at T condition than at NT condition, indicating a greater mechanical load on the patello-femoral joint when the knee trespasses he vertical line of the foot.

It is not possible a direct comparison with the studies by Fry[6] and by Abelbeck[1] because of the methodological differences and because they have not determined PFF. However, it is possible to compare the relative increases of the peak torque on the knee between these studies. Abelbeck[1] reported a 66% increase and Fry[6] a 30% increase, on the knee peak torque, while, at the present study, it was found an average increase of 38 ± 31%.

Not trespassing the knee over the foot line diminishes the patellofemoral compression force, causing less mechanical demand of this joint. The torque on the hip also increased during the condition in which the knee trespasses the ankle, which, analogously, may lead to a greater lumbar overloading at this condition.

Thus, since the patellofemoral force is greater on the condition in which the knee trespasses the ankle and that, in this condition, it seems to occur a greater torque on the hip joint, to perform free squatting, with the bar trespassing the knee over the line of the foot, does not seem to be the safest way for the execution of this movement. On the other hand, to trespass the knee, while performing squatting, could be justified by the increase of the quadriceps muscle demand during the performance of this movement. However, to this end, it is recommended to increase the exercise load instead of trespassing the knee over the line of the foot.

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