0195-9131/89/2105-0613$2.00/0
MEDICINE AND SCIENCE IN SPORTS AND EXERCISE

Vol. 21, No. 5
Printed in U.S.A.

# Effect of load, cadence, and fatigue on tibio-femoral joint force during a half squat

HEATHER C. HATTIN, MICHAEL R. PIERRYNOWSKI, and KEVIN A. BALL

*School of Physical and Health Education,*
*University of Toronto,*
*Toronto, Ontario, CANADA M5S 1A1*

ABSTRACT

HATTIN, H. C., M. R. PIERRYNOWSKI, and K. A. BALL. Effect of load, cadence, and fatigue on tibio-femoral joint force during a half squat. *Med. Sci. Sports Exerc.*, Vol. 21, No. 5, pp. 613–618, 1989. Ten male university student volunteers were selected to investigate the 3D articular force at the tibio-femoral joint during a half squat exercise, as affected by cadence, different barbell loads, and fatigue. Each subject was required to perform a half squat exercise with a barbell weight centered across the shoulders at two different cadences (1 and 2 s intervals) and three different loads (15, 22 and 30% of the one repetition maximum). Fifty repetitions at each experimental condition were recorded with an active optoelectronic kinematic data capture system (WATSMART) and a force plate (Kistler). Processing the data involved a photogrammetric technique to obtain subject tailored anthropometric data. The findings of this study were: 1) the maximal antero-posterior shear and compressive force consistently occurred at the lowest position of the weight, and the forces were very symmetrically disposed on either side of this halfway point; 2) the medio-lateral shear forces were small over the squat cycle with few peaks and troughs; 3) cadence increased the antero-posterior shear (50%) and the compressive forces (28%); 4) as a subject fatigues, load had a significant effect on the antero-posterior shear force; 5) fatigue increased all articular force components but it did not manifest itself until about halfway through the 50 repetitions of the exercise; 6) the antero-posterior shear force was most affected by fatigue; 7) cadence had a significant effect on fatigue for the medio-lateral shear and compressive forces.

KINETICS, TIBIO-FEMORAL FORCES, THREE-DIMENSIONAL, SQUATS

The human knee joint is surrounded by a large number of muscles which act in different directions and in different phases during any activity. Quantifying the forces across the knee, which includes both the tibio-femoral and patello-femoral joints, is complex due to this muscle activity. Among the earliest researchers of knee joint load are Bresler and Frankel (7). Unfortunately, these investigators did not measure muscular forces, thus the total force transmitted by the joint was not calculated. Kettelkamp and Chao (17) attempted to analyze forces utilizing only radiographic data, but this method was denounced as inaccurate by Harrington (10) and Johnson, Leitl, and Waugh (13).

Submitted for publication June, 1988.
Accepted for publication April, 1989.

An extensively used method for quantifying knee joint loads was that reported by Morrison (21–24) and his followers (6,9,10,12–14,18,25,26,29). Before Morrison, the total force acting across the knee joint had not been investigated in detail. Because this work involved a dynamic rather than a static analysis, two additional factors were taken into account. These included the accelerations and the mass moments of inertia of the lower limb segments. To perform these analyses, ground-foot reactions were directly measured using a force transducer while displacements of segment end-points or joint centers were recorded using optical, electronic, or opto-electronic systems.

Most studies have concentrated on ambulatory activities such as level walking, walking up and down a ramp, and climbing and descending stairs. However, studying the range of loadings carried by the joint during athletic activities is also of importance. The squat exercise has been frequently utilized in the development of the lower body, particularly the quadriceps muscles. The increasingly widespread use of the half squat exercise in weight lifting and training suggests that a better understanding of this movement would be invaluable to both physical practitioners and athletes.

Research concerning the kinetic factors involved in a half squat exercise has been limited. Reilly and Martens (27), using a force plate, along with stroboscopic photography, free body diagrams, equilibrium equations, and experimentally determined parameters, analyzed squatting statically. This likely underestimated the true values for the forces present. Ariel (4) investigated the half squat with a barbell weight using x-rays, cinematography, and a computer program for the analyses. Dahlkvist, Mayo, and Seedhom (8) used a technique similar to Morrison's in analyzing forces during a deep squat but added some improvements by employing a tailor made model for each subject formed through the application of x-rays. Recently, Bauman, Gross, Quade, Galbierz, and Schwirtz (5) have reported the net muscular moments at the knee during the snatch lift.

To date, there is no known 3D dynamic analysis of the forces acting on the knee joint during a repeated half squat. The present study looks at the magnitude of the tibio-femoral joint force in individuals squatting at two cadences carrying three external loads over 50 repetitions. Future work will examine the effect of this cyclical activity on the patello-femoral force during squatting exercises.

## METHODS AND PROCEDURES

**Subjects.** Ten male university student volunteers with mean age, height, and mass of 22.6 ± 2.24 yr, 177.5 ± 5.31 cm and 80.5 ± 11.08 kg, were selected for the experimental trials. All of these subjects had previous experience performing half squat exercises with barbells and could maximally lift 115.1 ± 21.6 kg. Informed consent was obtained from the subjects. None of the subjects had any known genu varum or valgum deformities, previous knee injury surgery, or neurological disorders that could affect their squatting ability.

**Skill.** The guidelines employed in this research for performing a half squat exercise were such that the lifter bent the knees and lowered the body, while keeping the trunk erect, with a weight in the form of a barbell centered across the shoulders, until the posterior surface of the thighs were parallel to the floor; then the subject returned to the standing position. Thus, two stages were defined within each cycle of the exercise: a yielding stage, during which the height of the barbell decreased, and a lifting stage, during which the height of the barbell increased. To determine the effect of fatigue, 50 continuous repetitions at each experimental condition were performed. These 50 repetitions were subdivided into an initial, middle, and final phase for discussion purposes.

Three load conditions were investigated; each was individualized based on the subject's one repetition maximum (RM) for a half squat exercise. Nominal values of 15, 22, and 30 percent of the 1 RM were chosen for the loads based on unpublished observations of Anderson and Haring, as cited by Sale and MacDougall (28).

The half squats were performed at two different cadences. Both the yielding and the subsequent lifting stages were timed to occur smoothly over equal intervals of 1 for the fast and 2 for the slow cadences. An audio metronome and occasional voice commands were used to help each subject maintain the proper tempo.

**Data collection.** To determine the magnitude of muscular and articular forces across the tibio-femoral joint, a four-segment (pelvis, right thigh, lower leg and foot) rigid link-segment model was defined. To custom tailor this model to a subject a photogrammetric anthropometric technique was used (31), similar to that developed by Jensen (11). Each subject stood upon a box, located between two large mirrors such that anterior and sagittal plane views were available to a 35 mm SLR camera (Olympus OM2, 135 mm zoom lens) positioned 13.0 m forward from the subject and 1.2 m above the floor. The resulting slides were traced, then digitized using a HP9874A digitizer controlled by an IBM PC-XT microcomputer. From the segmental endpoint and outline data obtained from two views, the volume and mass, and the 3D lengths, center of mass, and moment of inertia of each segment were determined. In addition, the 3D location of the surface markers were obtained, relative to external (laboratory fixed) and internal (segment fixed) reference systems.

A 3D optoelectronic kinematic data capture system (WATSMART, Northern Digital, Waterloo, Canada) was utilized to record the movements of the half squat motions. Twenty-four small infra-red light emitting diode markers (IREDs) were affixed to the four modeled segments (Fig. 1) and sensed by three infra-red sensitive electronic cameras at 50 Hz. For three dimensional reconstruction using the direct linear transformation (1), the cameras were placed in positions such that two cameras were thought to be able to view each marker concurrently. Camera calibration was done daily using a tubular steel frame with 24 IRED markers

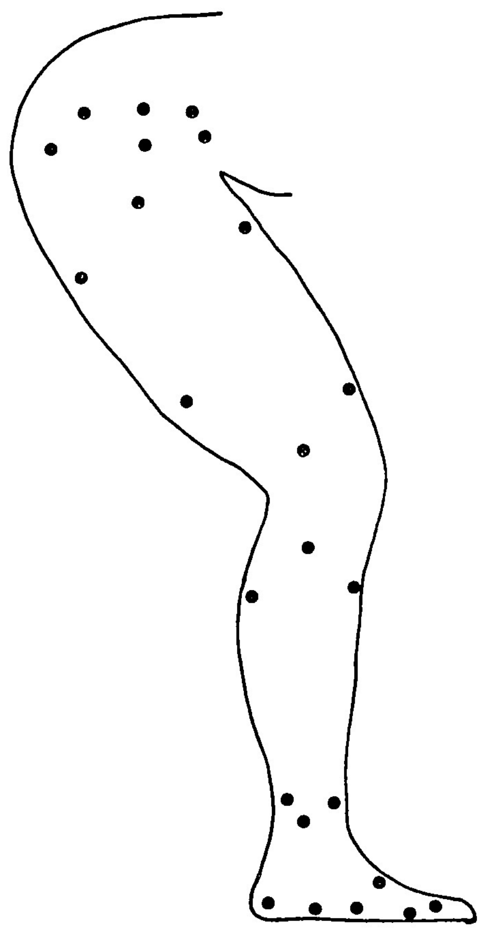

Figure 1—Location of the markers on the subject's right lower extremity.

with known three dimensional values. These markers were simultaneously observed by all three cameras.

The half-squat exercises were performed with the subjects' right leg on a floor mounted piezoelectric force plate (Kistler type Z24852/c and its associated electronic unit). This plate measured the ground reaction forces, center of pressure, and free moment acting on the right foot. An outline of the foot was drawn on the force plate to standardize its placement for every condition. The output from the force plate was connected to a 12 bit AD converter (WATSCOPE, Northern Digital, Waterloo, Canada), which was synchronously triggered with the WATSMART sub-system.

**Data analysis.** The body segment parameters and IRED marker locations were linked to the WATSMART 3D reconstructed data. This defined the location of the segments and IREDs during the movement relative to a laboratory fixed reference system. Each segment's location was defined by six values: the $x$, $y$, and $z$ coordinate of its proximal endpoint and three Cardan angles of rotation from a vertical starting position (32). These data were hand edited to delete inferior data, which were likely the result of active marker reflections obtained by the WATSMART system. A subsequent program searched for missing joint and segment data. Missing data were generated using a cubic spline interpolation and smoothing technique (19). The mean error band was set at 5 mm for translations and 2° for rotations. The translational and rotational accelerations were calculated by differentiating the segmental displacements with respect to time. As the raw displacement data were suitably smoothed, a finite difference approach was used to obtain this information (33). D'Alembert's principles of motion were then used to equate the net joint forces and moments (7) to the inertial, gravitational, and ground reaction forces acting on the body segments during their motion.

To estimate the articular force components, several steps were necessary. First, the angles of pull of the quadriceps ($A_q$) and hamstring ($A_h$) muscles, in degrees, were determined. Because the line of action of the patellar ligament relative to the tibia varies with the angle of flexion of the knee joint ($A_k$), the third order polynomial reported by Morrison (21) was used to specify $A_q$ from $A_k$.

$$A_q = 15 + 0.317 \times A_k - 0.0084 \times (A_k)^2 + 0.000031 \times (A_k)^3.$$

The hamstring muscle angle of pull was defined as the knee joint angle.

Second, the moment arms of the quadriceps ($MA_q$) and hamstring ($MA_h$) muscles were obtained using data obtained from Smidt (30) tailored to the size of the subject (E) from the thigh ($L_t$) and leg ($L_l$) lengths.

$$E = (L_t/0.391 + L_l/0.436) \times 0.05;$$
$$MA_q = E \times [4.406 + 0.0286 \times A_k - 3.915 \times (A_k)^2];$$
$$MA_h = -E \times [2.494 + 0.0696 \times A_k - 7.593 \times (A_k)^2].$$

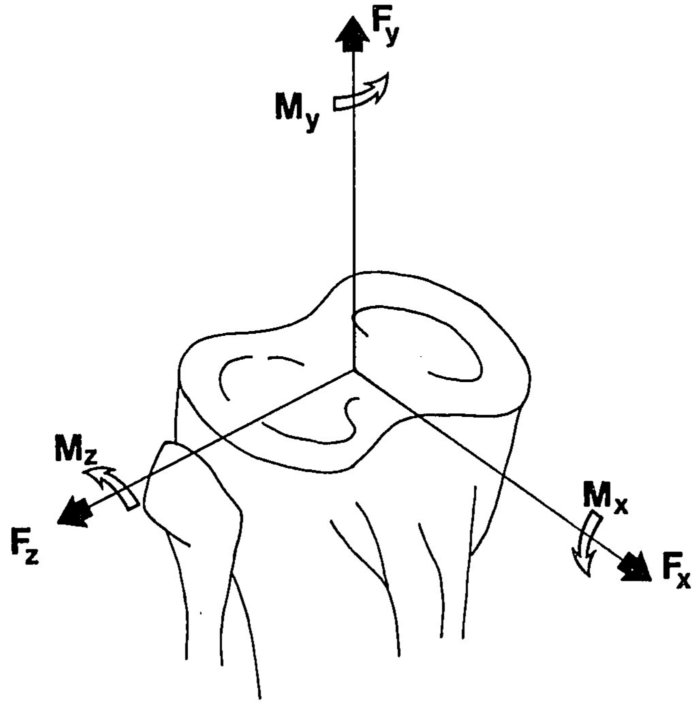


Figure 2—Net joint force ($F^j$) relative to the tibia (x = anterior-posterior, y = up-down, z = right-left).

Third, the net joint force ($F^j$) was determined relative to the tibia and then manipulated to conform to a reference system embedded upon the tibial plateau (Fig. 2) and normalized to body weight.

**Statistical analyses.** Subject and group means and standard deviations were calculated for every condition at each time interval. A multivariate repeated measures analysis of variance was performed on the mean maximal articular force components generated by all subjects for each of the six conditions and three collection intervals. A *post hoc* analysis using Scheffe's test was performed on the resulting data. Unless otherwise stated, the level of significance was set at $P = 0.05$.

## RESULTS AND DISCUSSION

The tibio-femoral joint articular force components ($x$ = anterior-posterior, $y$ = up-down, $z$ = right-left) for one subject performing an initial fast heavy squat is plotted in Figure 3 (mean ± SD). These curves begin and end when the knee was fully extended. As expected, the highest tibio-femoral loads occurred when the subject's knee position was flexed. Typical peak forces were 1, 7, and 0.1 times body weight in the $x$, $y$, and $z$ directions, respectively.

The effects of load, cadence, and fatigue on the antero-posterior shear, compressive, and medio-lateral shear tibio-femoral forces during the half squat exercise were complex. Each variable was proposed to have a direct effect on the forces generated. Interactive effects were also possible, so these were also considered.