Journal of Strength and Conditioning Research, 2002, 16(4), 516–524

Biomechanical Model and Evaluation of a Linear Motion Squat Type Exercise

KEVIN G. ABELBECK

OptiPro, Inc., Los Angeles, California 90064.

ABSTRACT

A biomechanical model of a squat exercise performed on a device using a bar that is restricted to a linear motion was developed. Hip and knee moments were evaluated at varying foot positions. The range of motion of the exercise was limited by the knee joint angle beginning at an 80° angle (flexed) to a 179° joint angle (extended). Variations in foot placement were evaluated for differences in torque applied about the transverse axes of the user's knee and hip joints. Because the user's feet were positioned farther forward (anterior), the moment about the knee decreased whereas the moment about the hip increased. Positive moments were those that resulted in forces to flex the knee and hip joints. Positive knee moments were determined in all conditions when the knee was flexed and became negative when the knee was at or near full extension. The model always produced positive moments about the hip. Thus, foot position is a critical factor in hip and knee moments, and therefore in the muscle groups stressed, in a linear motion squat type exercise.

Key Words: knee moment, hip moment, foot position

Reference Data: Abelbeck, K.G. Biomechanical model and evaluation of a linear motion squat type exercise. *J. Strength Cond. Res. 16(4):516–524. 2002.*

Introduction

The squat exercise is one of the most widely used exercises for increasing physical power and strength. Squats have been used in strength training protocols (6, 10, 16) and rehabilitation (13). Numerous variations of the squat exercise have been used in evaluating functional capacity (12, 11) and back stress (7, 5). Because the basic nature of the lifting movements are the same fundamentally, they are applicable to physical training programs as well as research and rehabilitation protocols.

Although the squat, in its most basic form, is a movement that has been undoubtedly used by human ancestors when they first began to walk upright, it does require a basic degree of balance and coordination. This is especially true with the squat exercise where a loaded bar is positioned on the trapezius muscle of the user. As an alternative form of the exercise, devices that move on linear tracks have been developed. One such device includes a bar with linear bearings on either end of the bar, and the bearings run on steel shafts providing vertical linear motion of the bar called a Smith machine. The same functional device with the addition of a pad to support the lower back and pelvis is commonly called a hack squat. The hack squat is often tilted from the vertical position such that the user is in a semireclined position while using the device, but the function of the linear motion is maintained.

Devices such as these have also been used to test functional performance (8, 14, 15). This removes the necessity of balance from the user because the machine has only a single degree of freedom (if we do not count the locking mechanism of the device).

The variation in loads from a standard squat vs. a squat on a linear motion machine is of interest because of the value of the exercise it provides to different muscles of the user. The loads placed on the body and specifically the moments that are applied about the knee and hip joints are of particular interest. In a traditional squat, the center of gravity of the system (user and weight) must constantly stay over the feet of the user, or the user will fall. Thus all force must be applied to the center of gravity (cg) vertically. The graphical depiction of a traditional squat exercise is shown in Figure 1a, and the modified free body diagram of the system is shown in Figure 1b. The free body is accurate in the force applied to the body from the weight vector $\mathbf{F_W}$ and the reaction of the ground vector $\mathbf{F_{EY}}$, but it is modified to show the center of gravity of the system ($\mathbf{cg_T}$), which includes the force vector $\mathbf{F_W}$. This was done to illustrate graphically how the vector $\mathbf{cg_T}$ must be positioned above the reaction force vector $\mathbf{F_{EY}}$, otherwise a moment would exist and the system would rotate, or in this case, tip over. Machines that run on a linear track, such as a Smith machine, allow variation in anterior-posterior foot placement, thus resulting in horizontal forces that are applied to the machine and floor. These forces have the potential to significantly alter the loads placed on the body.

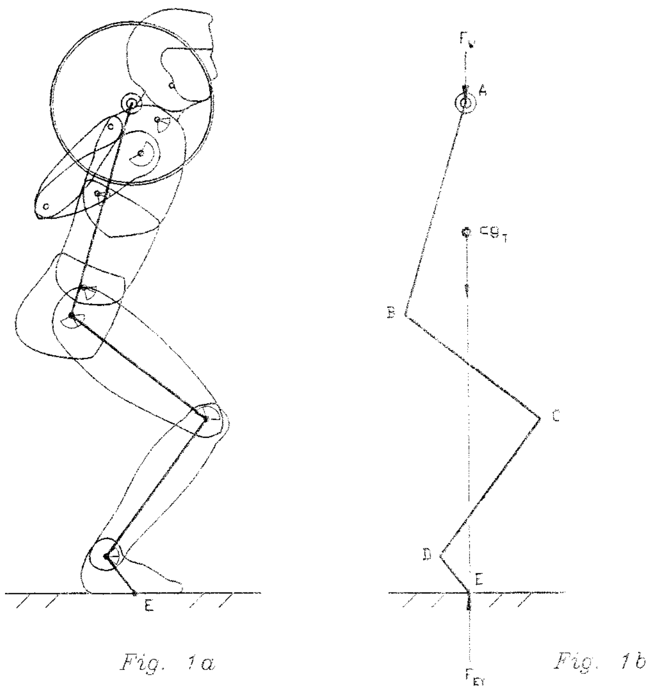

Figure 1. Schematic and free body diagram of traditional squat exercise as shown in the sagittal plane.

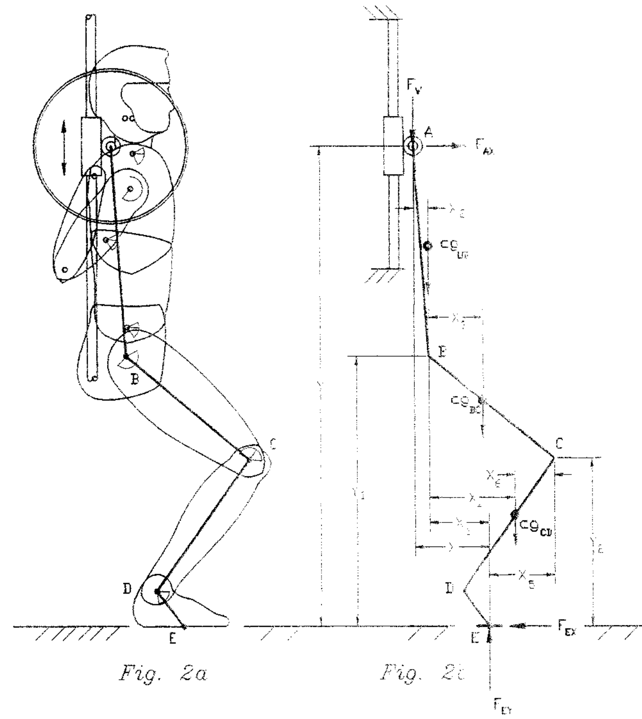

Figure 2. Schematic and free body diagram of linear motion squat as shown in the sagittal plane with X = 0.117 × h.

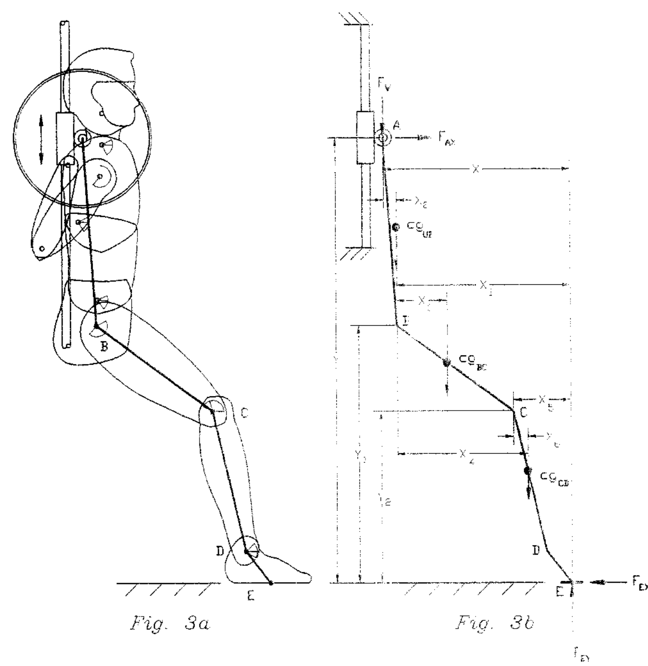

Figure 3. Schematic and free body diagram of linear motion squat as shown in the sagittal plane with X = 0.3107 × h.

Escamilla et al. (3) evaluated muscle recruitment of the knee using a linear motion squat. It was not evident whether the anterior-posterior foot placement was controlled in any way. The authors noted both the width between the feet and foot abduction angle. In a study by Blackburn et al. (1) a foot control was placed in front of the user's feet. This positioned the user's feet directly under the shoulder pads of a Universal Gym so that the users could perform modified squats.

The previously cited study (8) also provides reference to the foot position, but it was only documented for future reference in retesting. No intersubject evaluation was reported. In another study the angle of rotation of the leg in the transverse plane regarding muscle recruitment was evaluated (9). This was done using a traditional squat and not a linear motion device; therefore anterior-posterior foot placement was not an issue.

Variations in the placement of the user's feet can provide different positions for the body to apply and react to forces inherent to the system. Using such a device that restricts movement to 1 degree of freedom, many times in the vertical plane, allows the user to apply forces with a horizontal component. Variations in the anterior-posterior foot stance will alter the moments about the hip and knee joints of an individual while performing a squat type exercise on a machine restricted to motion in 1 degree of freedom.

Methods

Experimental Approach to the Problem

A two-dimensional mechanical model was created using a free body diagram of a linear motion squat; in this case a Smith machine was used. Various foot positions were used. In the most posterior position the ankle was placed directly under the hip of the user. In the most anterior position the feet were placed in front of the body such that when the user's knee joint angle was 90°, the upper leg was parallel to the floor. The foot position is designated by the dimension X; in Figures 2 and 3. This dimension is given as the perpendicular distance from the center of the bar (point A) to the vertical component of the reaction force vector offered to the feet by the floor ($\mathbf{F}_{EY}$). The location of this force vector is in the center of the foot and remains stationary through a single repetition of the exercise. The bar is positioned at the base of the neck, and the hip flexed forward at 2°. This slight hip flexion is reasonable partly because of the point of force application of the bar being positioned on the back of the neck, thus being slightly behind the vertical centerline of the body. The slight forward flexion minimizes the moment caused by the weight vector ($\mathbf{F}_W$) to open or extend the hip.

The values of X were chosen as 5 even increments between the posterior position where the ankle is under the hip (position #1) to the anterior position where the knee is at 90° when the upper leg is parallel to the floor (position #6), including these end positions, thereby generating the 6 positions. The knee angle is a traditional limit and guide for evaluating a squat exercise (1, 3, 8, 9, 14, 15, 16). The model uses relative relationships of each of these 6 values of X multiplied by the total height of the user (h) to give the following set of values:

Position #1: $X = 0.0607 \times h$

Position #2: $X = 0.1107 \times h$

Position #3: $X = 0.1607 \times h$

Position #4: $X = 0.2107 \times h$

Position #5: $X = 0.2607 \times h$

Position #6: $X = 0.3107 \times h$

This range in foot positions is considered typical and is commonly seen in the field. The foot position used in Figure 2 ($0.1107 \times h$) is commonly used. Position #2, as shown in Figure 2, depicts how the vertical reaction force offered by the ground to the user's feet ($\mathbf{F}_{EY}$) creates a force of flexion about the transverse axis of the knee (point C). The same force shown in Figure 3 results in a force that extends the knee (point C).

A mathematically scaled model of a man of average height (1.753 m) was created. The lengths of the limb segments of the legs were determined from anthropometric limb lengths from cadavers as reported by Duggar (2) from the original work of Braune and Fischer. In this study the upper and lower limb lengths were both determined to be 25.0% of the entire body length with a standard deviation of 1.13 and 0.78, respectively.

The model that was developed includes the components as illustrated in the free body diagram shown in Figure 2b. This includes the force applied by the

Table 1. Reported and mean anthropometric data.

	% Body weight	Center of gravity (% segment length)
Braune and Fischer (1889)* $n = 3$		
Upper leg	10.75	43.9
Lower leg	4.8	41.95
Foot	1.68	
Braune and Fischer (1892)* $n = 2$		
Upper leg	11.23	43.4
Lower leg	4.53	42.4
Foot	1.88	
Dempster (1955)† $n = 8$		
Upper leg	9.9	43.3
Lower leg	4.6	43.3
Foot	1.4	
Weighted means		
Upper leg	10.30 (20.60)‡	43.45
Lower leg	4.64 (9.28)‡	42.85
Foot	1.54 (3.08)‡	

* Reported in Duggar (2).
† Reported in Duggar (2) and Grieve and Pheasant (4).
‡ Weights are doubled due to two limbs.

weight ($\mathbf{F}_W$), the weight of the user broken down into composite centers of gravity for the components of the body, the reaction force offered by the ground through the feet of the body ($\mathbf{F}_{EY}$) and the horizontal components of force applied by the bar ($\mathbf{F}_{AX}$), and the force due to friction from the floor ($\mathbf{F}_{EX}$). The figure illustrates a designated 4-link body as specified by points A, B, C, D, and E, where B denotes the transverse axis of the hip and C is a generalized transverse axis of the knee. The body is evaluated in a static or constant velocity state, and therefore the body can be evaluated as a rigid member.

The centers of gravity contribute to moments about specific points, namely the points of interest, the hip (point B) and the knee (point C). Approximate locations of the centers of gravity of the component parts have been determined and a series of vectors designating these forces have been specified, namely the upper body ($\mathbf{cg}_{UB}$), the upper leg or segment BC ($\mathbf{cg}_{BC}$), and the lower leg or segment CD ($\mathbf{cg}_{CD}$). This composite method was done to assess the contribution of the mass of the body components in the presence of gravity, especially considering the variation in foot positions. The relative centers of gravity for each segment were compiled from data presented by Duggar (2) and by Grieve and Pheasant (4) as determined by Dempster. From these data, a mean of the weight of each segment, relative to the total body weight of the individual was determined, and the mean location of the center of gravity of each segment was calculated as the distance from the proximal joint as a percentage of the total segment length. The given data is relatively consistent, but to extrapolate the most accurate results the reported mean was weighted to the number of subjects used in each study. The results of the data are summarized in Table 1.

The weight of the user's feet were also evaluated from this data and used as a contribution to total body mass but not as a component that moves during the exercise. Therefore the mass of the feet does not affect the work done. The point of contact of the feet with the ground is also generalized and set at a stationary point in the center of the user's foot, designated by point E.

The range of motion was from a starting position of a knee angle of 80° (1.396 radian) to a final position of just before full knee extension, specifically a knee angle of 179° (3.125 radian). The summation of forces and moments, maintaining equilibrium in the system, was used to determine resultant forces in this free body including:

$$\mathbf{F_{EY}} = -(\mathbf{F_W} + \mathbf{cg_{UB}} + \mathbf{cg_{BC}} + \mathbf{cg_{CD}} + (\text{weight of the feet}(.0308 \times \text{BW}))) \quad (1)$$

Summing the moments about point A ($\mathbf{M_A}$) to find $\mathbf{F_{EX}}$ gives:

$$\mathbf{F_{EX}} = (\mathbf{F_{EY}} \times \text{X})/(\text{Y}) \quad (2)$$

Again summing forces yields:

$$\mathbf{F_{AX}} = -\mathbf{F_{EX}} \quad (3)$$

where $\mathbf{F_{AX}}$ is the horizontal component of the force applied by the bar to the back of the user and $\mathbf{F_{EX}}$ is the frictional force applied to the user's feet to maintain the horizontal position. The negative sign has been added to Equations (1) and (3) to designate the direction of the vectors. This direction has also been corrected in the figures and the above equations.

As with any model some generalizations must be made. In this case, these include the upper body being 1 rigid member maintaining its position through the vertical movement of the exercise, with the center of gravity of the upper body ($\mathbf{cg_{UB}}$) positioned directly above the hip (point B). Subtracting the sum of the masses of the lower limbs (including the feet) from the total yields the upper body weight, which is 67.04% of the total body weight (BW). The consistent orientation of the upper body is well maintained as observed by users in the field, and in some cases it is mandated by the machine. This is the case in the previously cited studies (3, 8) and in many types of training equipment that simulate the squat exercise, such as a hack squat. In these devices a pad supports the back and pelvis of the user, thus maintaining their relative positions. Therefore this assumption is considered reasonable for the purposes of this evaluation and is used consistently in this model.

To calculate moments about the hip (point B) and the knee (point C), all external forces and forces resulting from the mass of body were evaluated. The sign convention for this model was a positive moment that applies a load to cause joint flexion. This mandates positive work being done from the hip and knee joints when they extend. Thus, given the orientation, for the moment about point B ($\mathbf{M_B}$) a positive moment is one such that the resultant force applied tends to decrease angle ABC. For the moment about point C ($\mathbf{M_C}$), a positive moment is one such that the resultant force applied tends to decrease angle BCD. The sum of the moments about point B is:

$$\mathbf{M_B} = [-(\mathbf{F_W} \times \text{X}_2) - (\mathbf{cg_{BC}} \times \text{X}_3) - (\mathbf{cg_{CD}} \times \text{X}_4) + (\mathbf{F_{EY}} \times \text{X}_1) + (\mathbf{F_{AX}} \times (\text{Y} - \text{Y}_1)) - (\mathbf{F_{EX}} \times \text{Y}_1)] \quad (4)$$

and similarly for the moment about point C:

$$\mathbf{M_C} = [(\mathbf{F_W}(\text{X} + \text{X}_5)) + (\mathbf{cg_{UB}}(\text{X} + \text{X}_5 - \text{X}_2)) + (\mathbf{cg_{BC}}(\text{X}_5 + \text{X}_1 - \text{X}_3)) - (\mathbf{cg_{CD}} \times \text{X}_6) + (\mathbf{F_{EY}} \times \text{X}_5) - (\mathbf{F_{AX}} \times (\text{Y} - \text{Y}_2)) + (\mathbf{F_{EX}} \times \text{Y}_2)] \quad (5)$$

where the contribution of $\mathbf{F_{EY}}$ is positive when point E is posterior to point C and negative when point E is anterior to point C.

If Equations (4) and (5) are expanded, users of any height with similar proportions can be represented by the model. The distances can be proportionally scaled to any user's height by normalizing them to the 1.753-m height by multiplying all X and Y values by h/1.753, where h designates the height of the user in meters. This model uses 1.753 m consistently, but because this body would be scaled proportionally, the moments generated would increase by the same proportionate amount.

Body weight can be normalized assuming the values for the model as previously determined. As shown in Table 1, the mass of the upper body is 67.04% of the total body weight (BW), the upper legs comprise 20.60% of BW, the lower legs comprise 9.28% of BW and feet comprise 3.08% of BW. Thus:

$$\mathbf{cg_{UB}} = 0.6704 \times \text{BW}; \quad \mathbf{cg_{BC}} = 0.2060 \times \text{BW}; \quad \text{and} \quad \mathbf{cg_{CD}} = 0.0928 \times \text{BW}$$

where with the addition of 0.0308 × BW for the feet, the sum is 1.00 × BW. The dimensions are given as a ratio of the total segment length to universally normalize the lengths according to any height subject, and the model also used weight as a relative ratio to

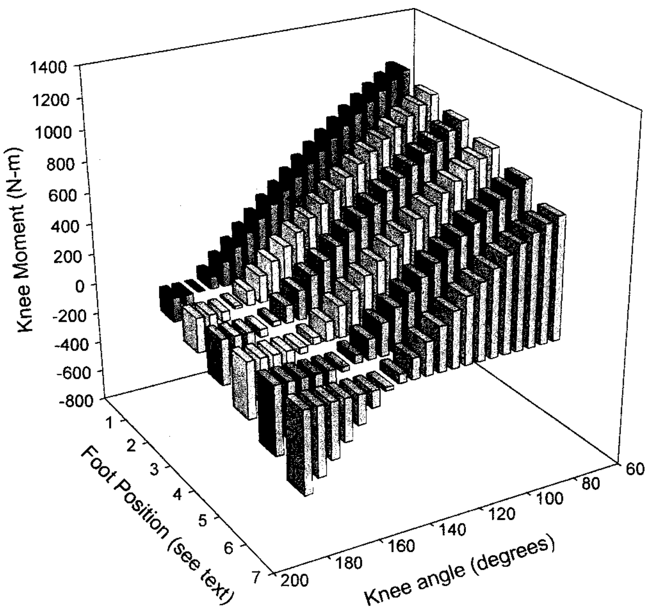

Figure 4. Knee moments at relative knee joint angles with foot positions of varying anterior placement while performing a linear squat.

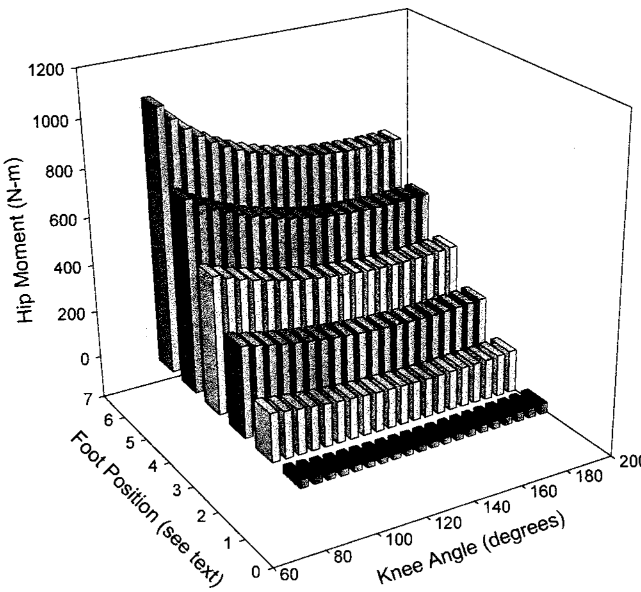

Figure 5. Hip moments at relative knee joint angles with foot positions of varying anterior placement while performing a linear squat.

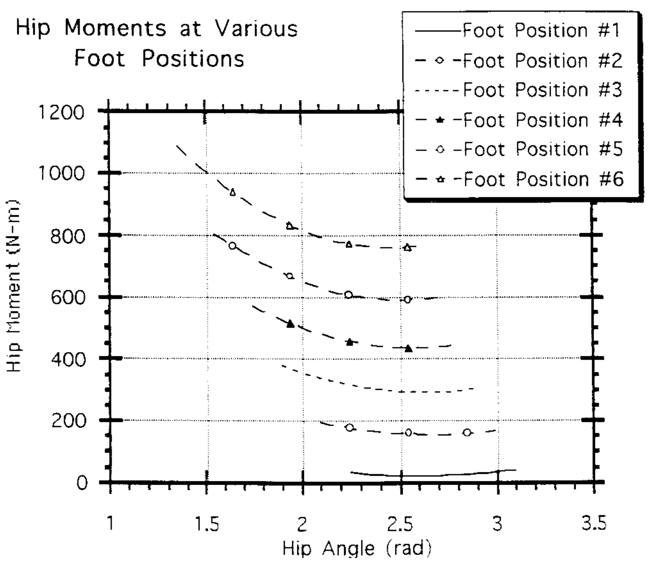

Figure 6. Hip moments at relative hip joint angles with foot positions of varying anterior placement while performing a linear squat.

total body weight. This was done to enable variations in body weight to be evaluated by the model.

Anthropometric data about the distance from the bottom of the foot to the transverse axis of the ankle when in a standing position were not available. The data were not readily available because they had to be gathered from cadavers. Because the variation in shoe sole thickness would also vary from one user to another, a determined value of $0.0523 \times h$ was used throughout this (1.753 m high) model for the vertical position, and a distance of $0.0387 \times h$ for the horizontal distance between points D (ankle) and E ($\mathbf{F}_{EY}$) was also used.

Results

A person with a body weight of 110 kg capable of lifting a 110-kg load ($\mathbf{F}_W$) was chosen for the initial evaluation. The model generated moments about the knee that were evaluated at foot positions 1–6 and are displayed in Figure 4. With the foot at position #1, where the foot is under the user's hip, the torque about the knee is significantly greatest with the knee flexed and decreased with extension and also at regular intervals at the foot positions that were more anterior to the body. The foot position #1 graph depicts a negative value at knee angles greater than 2.890 radian or 165.6°. This knee angle is the actual angle of the joint, 180° being fully extended. Subsequent foot positions start at a lower positive value with the same knee angle of 1.396 radian (80.0°), and the torque values decrease at a similar rate and shifts to a negative value at a smaller joint angle. This angle at position #6 is approximately 2.552 radian (146.2°). This negative moment means the resultant torque about the knee is now being applied such that it is assisting the knee to extend.

A similar depiction, relative to the hip joint is shown in Figure 5. Here the relationship between the hip moments and the knee angle is shown at the various foot positions. All moments are positive in that the resultant moment is such that it is always opposing extension of the hip. The relationship as to the value of the moment increases as the feet are positioned more anteriorly. This is the opposite of the knee moment. This trade off is logical because if the total work

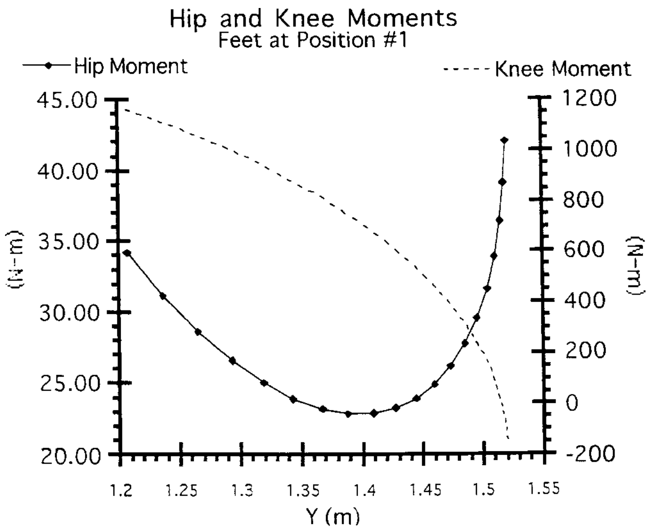

Figure 7. Hip and knee moments at relative vertical bar distance in a linear squat with feet positioned under the body.

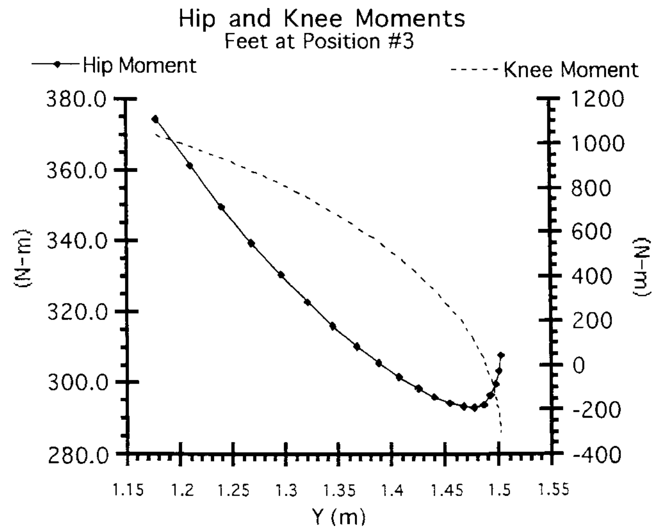

Figure 8. Hip and knee moments at relative vertical bar distance in a linear squat with feet positioned anterior to the body.

done is relatively constant, inherently one would increase at the other decreases.

The hip angle relationship to hip moment is shown in Figure 6. The hip joint angle is taken from the orientation of the spine (which is positioned at 2° of flexion) to the midline of the upper leg (segment BC in Figures 2 and 3). Because the model was developed around the knee angle, the angular displacement of the hip was not consistent at various foot positions. Therefore fourth-order polynomial curve fit equations were developed for the knee moment vs. hip angle data. A correlation coefficient of $r = 0.9999$ or more was maintained for each curve fit, and the range limits were set to the actual hip angle generated by the model using the knee angle that limits the range of motion. The knee moment at position #1 stayed under 45 N m for the entire range, whereas it was more than 1000 N-m in position #6.

The hip and knee moments together are evaluated in Figure 7 with the foot position #1. Here the values of the hip and knee moments are set against the vertical position (Y) of the center of the bar as it starts from its position with the 1.753-m tall individual, with the bar on the user's trapezius, beginning with the knees at 80° (1.396 radian) and extending the hip and knee joints to a 179° (3.125 radian) knee angle. This vertical position reference (Y) presented along the X axis was done to show the relationship of each curve as it relates to a variable common to both the knee and the hip angular displacements, at the vertical position of the bar. The inverse relationship of the shape of the curves can be easily evaluated in the starting and ending positions of each curve.

The same relationship is shown in Figure 8 as it relates to the hip and knee moments with the feet at position #3. The knee moment decreases slightly as the feet move farther away from the user's body (position #3) and the hip moment, in a flexed position, increases more than 10 times as that seen with the feet at position #1.

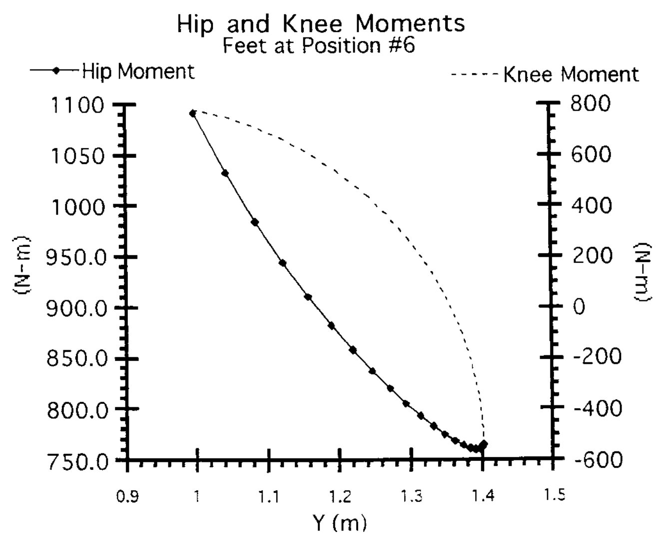

Figure 9. Hip and knee moments at relative vertical bar distance in a linear squat with feet positioned far anterior to the body.

Again, the hip and knee moments were evaluated and depicted in Figure 9, only at foot position #6, out in front, anterior to the user's body. The knee moment decreased again while maintaining the same general shaped curve. The hip moment drastically increased to a starting position more than 30 times that seen in position #1. The concavity is maintained in all hip moment curves, with the ending point being approximately 18 times that of the moment at the same hip angle at position #1.

The net area under the curves, shown in Figures 4

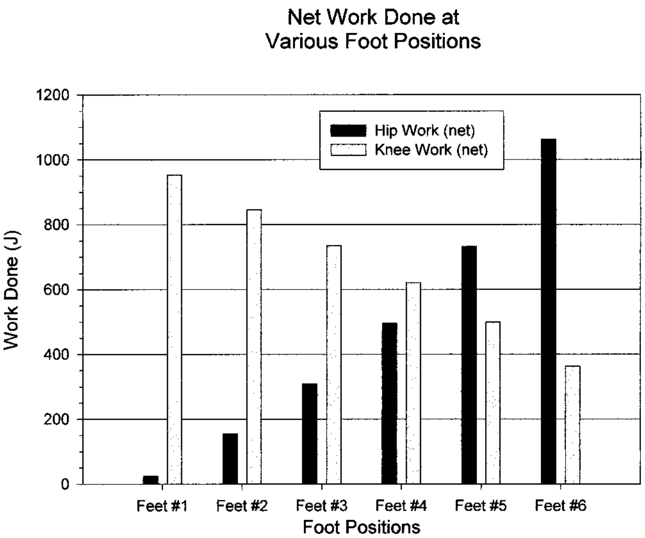

Figure 10. New work done about the hip and knee joints in a linear squat with foot positions of varying anterior placement.

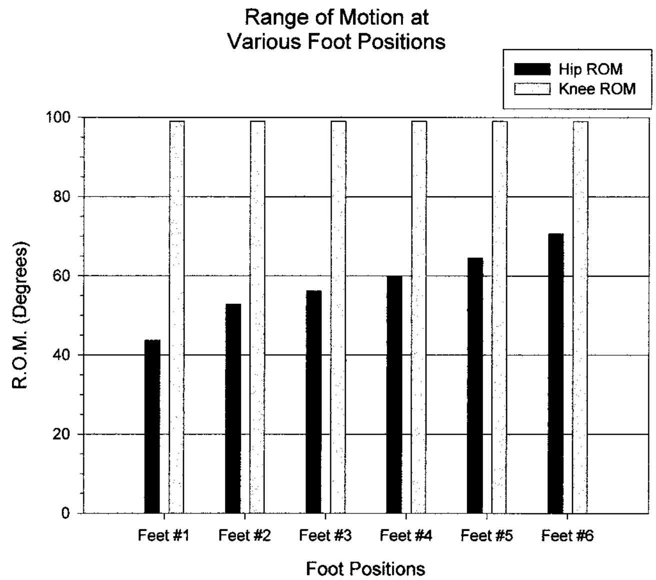

Figure 11. Range of motion of the hip and knee joints in a linear squat with foot positions of varying anterior placement.

and 6, were measured to determine the work done by both the hip and knee being limited by the angular displacement of the knee. The net work was determined by the positive area under the curve. Thus, the positive work is the integral of the function of the hip moment vs. the radian angular displacement ($f(\mathbf{M_B}\theta)$), where θ is the knee joint angle in radians. For the hip, the net work done ($\mathbf{W_{Bnet}}$) was determined by the equation:

$$W_{Bnet} = \int_{1.396}^{3.125} f(M_B\theta)\, d\theta \qquad (6)$$

A similar process was used to determine the net work done about the knee joint. In this case, negative work is done in some foot positions near full extension of the knee. The negative work was the integral of the equation relative to the knee ($f(\mathbf{M_C}\theta)$), with the X value being the limit as the function ($f(\mathbf{M_C}\theta)$) crosses the X-Y plane at each foot position. The other limits are the beginning and end points of the knee angle of the function (1.396 and 3.125 radian, respectively). The negative work was subtracted from the positive work to determine the net work. This was determined by the equation:

$$W_{Cnet} = \int_{1.396}^{x} f(M_C\theta)\, d\theta - \int_{x}^{3.125} f(M_C\theta)\, d\theta \qquad (7)$$

The absolute value of the knee moment ($\mathbf{M_C}$) was always used, and therefore the negative of the second half of the equation was used even though the function would generate a negative moment when crossing the X-Y plane. The absolute values were used to evaluate the positive and negative work values independently.

A graph of the relative work done is shown in Figure 10 at various foot positions. This graph depicts the decrease in positive work done about the knee joint and an increase in work done about the hip joint as the feet move anteriorly, away from the user's midline.

The range of motion of the hip is depicted relative to the knee in Figure 11. Because the limits of the movement were determined by the knee angle limits, this bar graph is at a constant height of 99°. The hip angle increases as does the work done as the feet move further anterior to the user's body.

Discussion

When the user's feet are positioned under the user in a linear motion squat exercise with a 110-kg user and a 110-kg load, the knee moment starts (knees flexed) at a value that is over 30 times greater than the hip moment and ends maintaining a positive hip moment while generating a small negative knee moment (fully extended knee). As the user positions his feet anterior to the body, the knee moment decreases slightly while the hip moment increased by an average of 10.9 times from position #1 to position #3 and an average of over 29 times from position #1 to position #6.

The data show an enormous variance in hip to knee moments. When there is a negative knee moment, such as with the feet positioned anterior to the body, it does not mean that the knee extensors are not forcefully contracting. A free body diagram of that segment would have to be done to evaluate muscle loading. These are internal forces to the free body chosen for this study. The long head of the biceps femoris will be contracting to assist in the large moment placed on the hip, and this muscle applies load to produce flexion of the knee, thus contributing to a positive knee torque due to internal forces.

A slightly curvilinear increase in net work done about the hip joint is seen as the feet move incrementally anteriorly from the user's midline and a similar decrease in work done is seen about the user's knee. This data shows a nearly inverse and linear relationship between these variables.

The general form of the knee torque curve generated by the model (Figure 4) is consistent with the electromyography findings and the percent maximum voluntary isometric contraction (%MVIC) for the vastus medialis and vastus lateralis muscles (3, 9) as well as the rectus femoris muscles (3) of both the traditional squat and a linear motion squat. Also, the peak muscle activity for these muscles (knee extensors) is significantly greater than that of the knee flexors, rated as a percentage of maximum. This correlates well with the results of the model. The data of the model are generally consistent in that a greater positive moment about the knee relates to a greater activation of the knee extensors, but to truly evaluate specific muscles a free body diagram of segments of the leg must be analyzed.

As seen in Figures 5 and 6, a slight increase in the hip moment can be seen as the knee extends (and so does the hip) when the feet are in the position #1, under the body. The moment decreases as the knee is extended when the foot positions are further anterior to the body, and do so at an increasing rate as the feet get farther in front of the user. Each curve is slightly concave. This concavity, though more drastically ascending, is seen in the relationship to %MVIC of the EMG regarding the traditional squat (9). The EMG data of the linear leg press shows a relatively constant relationship through the range of motion. This data is derived from muscular activity from the biceps femoris muscles and the medial hamstrings and is of a value between 20 and 40% MVIC with the knee extensors being much greater (initially near 80%) at knee flexion (3). Although anterior-posterior foot position was not documented in the aforementioned study, if the user's feet were positioned at a midpoint (positions #3-#4) the model (Figure 5) is consistent with these findings.

These muscles would be activated relative to the torque applied to the hip in the model because these muscles contribute to hip extension. The primary hip extensor, the gluteus maximus (glutes), was not evaluated by the studies cited. Even so, it is understandable that the relationship of moment applied to the muscle activation would still be consistent in the linear motion squat but quite different with the traditional squat (3).

A relative increase in the horizontal component of force may be seen in a hack squat machine or other devices where a supportive pad is placed at or below point B on the model. If the moment arm of the perpendicular distance from the floor to point A is reduced by half, the force doubles. As the user lowers, point B gets closer to the ground (platform), and this value continually increases during the downward movement. This reaction force from the pad decreases its perpendicular distance in a negative moment as it nears the knee and the force at the foot ($\mathbf{F}_{EX}$) increases along with its moment because of the relatively small decrease in perpendicular distance of this moment arm. The same movement on a hack squat would appear to increase the knee moment relative to the hip moment.

Practical Applications

In all aspects of athletics, conditioning, and rehabilitation, the user desires to stress specific muscles groups in certain exercises. The range of this stress can be very broad when applied to compound movements and especially explosive power movements such as Olympic lifts. For more specific training such as in the hypertrophy and strength phases in athletic programs, bodybuilding programs and rehabilitation where more specific muscle groups are targeted, it is vitally important to control relative joint loading and therefore stress in specific muscle groups. A higher movement about the knee will result in more stress on the quadriceps group and less stress on the glutes and hamstrings. The reverse is true for greater hip moments. Placement of the user's feet closer under the body results in greater stress on the quadriceps and more work done by these muscles. Placement of the feet farther in front of the body generates more stress on, and work done by the glutes and hamstrings. In addition, regardless of the foot placement, the depth of the squat is also relevant to the joint loading. From the low position to the extended position, the knee moment decreases rapidly. By contrast, the hip moment drops and then plateaus at roughly the midpoint in the upward stroke of the squat. Therefore the depth of the squat should also be considered when targeting specific muscle groups.

References

1. Blackburn, J.R., and M.C. Morrissey. The relationship between open and closed kinetic chain strength of the lower limb and jumping performance. *J. Orthop. Sports Phys. Ther.* 27:430–435. 1998.
2. Duggar, B.C. Anthropometry and human engineering. In: *The Human Body in Equipment Design.* A. Damon, H.W. Stoudt, and R.A. McFarland, eds. Cambridge, MA: Harvard University Press, 1966. pp. 157–186.
3. Escamilla, R.F., G.S. Fleisig, N. Zheng, S.W. Barrentine, K.E. Wilk, and J.R. Andrews. Biomechanics of the knee during closed kinetic chain and open kinetic chain exercises. *Med. Sci. Sports Exerc.* 30:556–569. 1998.
4. Grieve, D., and S. Pheasant. *The Body at Work.* W.T. Singleton, ed. London: Cambridge University Press, 1982. pp. 71–200.

5. HAGEN, K.B., AND K. HARMES-RINGDAHL. Ratings of perceived thigh and back exertion in forest workers during repetitive lifting using squat and stoop techniques. *Spine* 19:2511–2517. 1994.
6. HICKSON, R.C., K. HIDAKA, AND C. FOSTER. Skeletal muscle fiber type, resistance training, and strength-related performance. *Med. Sci. Sports Exerc.* 26:593–598. 1994.
7. KUMAR, S. Lumbosacral compression in maximal lifting efforts in sagittal plane with varying mechanical disadvantage in isometric and isokenetic modes. *Ergonomics* 37:1975–1983. 1994.
8. MUNICH, H., D. CIPRIANI, C. HALL, D. NELSON, AND J. FALKEL. The test-retest reliability of an inclined squat strength test protocol. *J. Orthop. Sports Phys. Ther.* 26:209–213. 1997.
9. NINOS, J.C., J.J. IRRANG, R. BURDETT, AND J.R. WEISS. Electromyographic analysis of the squat performed in self-selected lower extremity neutral rotation and 30 degree of lower extremity turn-out from the self-selected neutral position. *J. Orthop. Sports Phys. Ther.* 25:307–315. 1997.
10. PANARIELLO, R.A., S.I. SHERRY, AND J.W. PARKER. The effect of the squat exercise on anterior-posterior knee translation in professional football players. *Am. J. Sports Med.* 22:768–773. 1994.
11. ROBERTSON, J.M., P. WITT, AND M.T. GROSS. A comparison of trunk extensor strength and squat lifting ability. *J. Orthop. Sports Phys. Ther.* 25:137–144. 1997.
12. SUNI, J.H., S.I. MIILUNPALO, T.M. ASIKAINEN, R.T. LAUKKANEN, P. OJA, M.E. PASANEN, K. BOS, AND I.M. VUORI. Safety and feasibility of a health-related fitness test battery for adults. *Phys. Ther.* 78:134–147. 1998.
13. THEIN, J.M., AND L.T. BRODY. Aquatic-based rehabilitation and training for the elite athlete. *J. Orthop. Sports Phys. Ther.* 27:32–38. 1998.
14. VOLEK, J.S., W.L. KRAEMER, J.A. BUSH, M. BOETES, T. INCLEDON, K.L. CLARK, AND J.M. LYNCH. Creatine supplementation enhances muscular performance during high-intensity resistance exercise. *J. Am. Diet Assoc.* 97:765–770. 1997.
15. WILSON, G.J., A.D. WALSHE, AND M.R. FISHER. The development of an isokinetic squat device: Reliability and relationship to functional performance. *Eur. J. Appl. Physiol.* 75:455–461. 1997.
16. WISLOFF, U., J. HELGERUD, AND J. HOFF. Strength and endurance of elite soccer players. *Med. Sci. Sports Exerc.* 30:462–467. 1998.

Note: This is a technical paper and not a human experiment, therefore no human subjects were used in this study.

Address correspondence to Kevin G. Abelbeck, kabelbeck@yahoo.com.