Joint Torques and Joint Reaction Forces During Squatting With a Forward or Backward Inclined Smith Machine

Andrea Biscarini, Fabio M. Botti, and Vito E. Pettorossi
University of Perugia

We developed a biomechanical model to determine the joint torques and loadings during squatting with a backward/forward-inclined Smith machine. The Smith squat allows a large variety of body positioning (trunk tilt, foot placement, combinations of joint angles) and easy control of weight distribution between forefoot and heel. These distinctive aspects of the exercise can be managed concurrently with the equipment inclination selected to unload specific joint structures while activating specific muscle groups. A backward (forward) equipment inclination decreases (increases) knee torque, and compressive tibiofemoral and patellofemoral forces, while enhances (depresses) hip and lumbosacral torques. For small knee flexion angles, the strain-force on the posterior cruciate ligament increases (decreases) with a backward (forward) equipment inclination, whereas for large knee flexion angles, this behavior is reversed. In the 0 to 60 degree range of knee flexion angles, loads on both cruciate ligaments may be simultaneously suppressed by a 30 degree backward equipment inclination and selecting, for each value of the knee angle, specific pairs of ankle and hip angles. The anterior cruciate ligament is safely maintained unloaded by squatting with backward equipment inclination and uniform/forward foot weight distribution. The conditions for the development of anterior cruciate ligament strain forces are clearly explained.

Keywords: Smith squat, joint load, cruciate ligaments, line of gravity

Squatting, a fundamental strengthening exercise, is an integral part of many training programs in sports and fitness, and is commonly prescribed in rehabilitative interventions. Squat biomechanics have been the subject of extensive research studies with particular focus on muscle activity and safety for knee structures.1–12 A variety of different squat techniques and equipment has been developed over the years to comply with the postural, joint, and muscular needs and demands of individual people and special populations. These squat variants are primarily characterized by different modalities of administration of the resistance load, which is represented by body weight, and by the gravitational overload directly provided by weighted barbells, dumbbells, and belts, or transmitted to the body by lever and/or cable/pulley systems. The resistance load system, together with the instantaneous values of the kinematical parameters, determines muscle forces, and the joint reaction forces resulting from bony contact forces and ligament tensions. Knowledge and control of muscular and joint forces occurring during exercises is of fundamental importance in the design of suitable rehabilitation and conditioning programs, as well as for injury prevention.13,14

Different equipment imposes specific degrees of mechanical constraint to the squat exercise. For example, in the free barbell squat, the center of mass (C) of the system consisting of the user’s body and the weighted barbell should fall between forefoot and heel, and consequently hip, knee, and ankle joint angles take “in-phase” values which, within small inter-subject variability ranges, are tightly related to each other. Thus, each phase of the exercise is characterized by a well-defined joint torque distribution, i.e., by specific ratios of the torques at different joints.

The Smith machine (Technogym, Cesena, Italy) is a popular piece of equipment used in weight training and particularly in squat exercises (Smith squat). It consists of a barbell constrained to move up and down sliding along rectilinear steel tracks (Figure 1). In the Smith squat, the tracks’ reaction forces ($R_s$ in Figure 1) acting on the barbell compensate forward or backward imbalances of C determined by backward or forward foot displacements or trunk tilts. Therefore, the value of the joint angles may be changed independently of the other, enabling a wider range of exercise positions and, concurrently, a wider
range of possibilities for modulating the distributions of muscle activity and joint loads.

In contrast to the free barbell squat, little attention has been devoted in the literature to understand the specific biomechanical properties of the Smith squat, despite its great popularity and its extensive use for different purposes. For long time the safety and effectiveness of the Smith squat has only been the subject of a continuous and heated debate in the fields of athletic training, fitness, and rehabilitation. Only in very recent times, Biscarini et al developed a specific biomechanical model for this exercise and derived the knee, hip, and lumbosacral joint torques, the shear and compressive components of the tibiofemoral joint loads, and the patellofemoral force, as a function of external load, foot positioning, trunk tilt, and body configuration, for all the combinations of hip, knee, and ankle joint angles allowed by the Smith machine. It was clearly highlighted that, compared to the free barbell squat, the Smith squat may be largely patterned to modulate the distribution of muscle activities and minimize the mechanical load on specific joint structures.

Muscle activity and joint load distribution may be further modulated as the equipment’s tracks are inclined by an angle $\alpha$ with respect to the vertical, by an equal $\alpha$ rigid rotation of the Smith machine and its ground platform (Figure 1). This is equivalent to a reverse $\alpha$ rotation of the line of gravity with respect to the machine. The different body positioning (backward/forward trunk tilt, foot placement, and ankle, knee, and kip angles) and equipment inclinations can be managed concurrently to get specific biomechanical outcomes.

Joint torques and forces in Smith squat have previously been assessed only for a vertical orientation ($\alpha = 0$) of equipment tracks. The aim of this paper is to gain full understanding of the equipment potential by extending the calculations to squat exercises performed in a backward or forward inclined Smith machine (with different degrees of inclination $\alpha$). The final goal is to define the most favorable combinations of equipment inclination and body positioning to unload specific joint structures while focusing the activation on selected muscle groups. The possible applications in training and rehabilitation are discussed with particular focus on the control of the strain-forces on cruciate ligaments and lumbosacral joint. Finally, we will address the postural implications associated with the forward/backward equipment inclination.

**Methods**

The human body is modeled by 14 linked rigid segments (head, trunk, upper arms, forearms, hands, thighs, shanks and feet), whose relevant biomechanical parameters (weight, dimensions, center of mass, and moments of inertia) have been deduced from extensive anthropometric studies. The weighted barbell, placed on the user’s shoulder, may only slide along the rectilinear tracks of the Smith machine (Figure 1). The direction of the tracks (y-axis) may be inclined by an angle $\alpha$ with respect to the vertical, by an equal $\alpha$ rigid rotation of the Smith machine and its ground platform (x-axis), that is assumed to be integral with the machine (Figure 2). The joint angles $\theta_{\text{ankle}}, \theta_{\text{knee}},$ and $\theta_{\text{hip}}$ take a value of $0^\circ$ when, due to an ideal full flexion, the corresponding bony segments overlap each other. The angle $\theta_{\text{trunk}}$ defines the trunk tilt with respect to the tracks.

The external forces acting on the system (WB) constituted by the user’s body (B) and the weighted barbell (W) are equivalent to the following vectors (Figure 1):

- The weight $Mg$ of the system, applied in the center of mass $C = (x_C, y_C)$ of the system ($M$ is the sum of the mass $M_B$ of the weighted barbell and the mass $M_B$ of the user’s body).
- The two equal and frictionless (x-axis oriented) reaction forces $\vec{R}$, exerted by equipment’s tracks on the barbell, and applied at the two contact points $(0, y_w)$ between the barbell and the tracks.
- The shear (parallel to the ground platform) and normal (to the ground platform) ground reaction force $\vec{R}_s$ acting on the barbell, constrained to two rectilinear tracks, may only move along a line (y-axis). The external forces acting on the system constituted by the user and the weighted barbell are reported in the figure.

**Figure 1** – Sketch of the fundamental elements of the Smith squat exercise: the barbell, constrained to two rectilinear tracks, may only move along a line (y-axis). The external forces acting on the system constituted by the user and the weighted barbell are reported in the figure.
forces $\tilde{A}_{GR}$ and $\tilde{N}_{GR}$, acting symmetrically on each foot, and applied at a point $(x_{GR}, 0)$ within the contact surface between each foot and the ground.

The location of the application point of the ground reaction forces depends upon the user-controlled weight distribution between forefoot and heel, and can be instantly determined during the exercise with the use of a force platform. In a quasi-static Smith squat exercise, the external forces are subjected to the equilibrium condition

$$2\tilde{N}_{GR} + 2\tilde{A}_{GR} + 2\tilde{R}_s + Mg = 0$$

(1)

Projection of this equation onto the $y$ and $x$ axes gives

$$\begin{align*}
2N_{GR} - Mg \cos \alpha &= 0 \\
2(R_s)_x + 2(A_{GR})_x - Mg \sin \alpha &= 0
\end{align*}$$

(2)

(3)

where $N_{GR}$ and $Mg$ are the magnitudes of $\tilde{N}_{GR}$ and $M\tilde{g}$, while $(A_{GR})_x$ and $(R_s)_x$ are the scalar $x$-components of $\tilde{A}_{GR}$ and $\tilde{R}_s$, respectively. The equilibrium condition for the axial moments of the external forces takes a simple form when the moments are calculated in respect to the axis crossing the $(x_{GR}, 0)$ point and normal to the $xy$ plane:

$$2y_w (R_s)_s - y_c Mg \sin \alpha - (x_{GR} - x_c) Mg \cos \alpha = 0$$

(4)

Equations (2–4) determine the reaction forces of the ground and the equipment tracks:

$$N_{GR} = \frac{Mg \cos \alpha}{2}$$

(5)

$$(R_s)_s = \frac{y_c Mg \sin \alpha + (x_{GR} - x_c) Mg \cos \alpha}{2y_w}$$

(6)

$$(A_{GR})_x = \frac{Mg \sin \alpha}{2} - \frac{y_c Mg \sin \alpha + (x_{GR} - x_c) Mg \cos \alpha}{2y_w}$$

(7)

With the knowledge of such reaction forces, one can calculate a specific joint torque (the torque provided by the muscles crossing the joint and by the passive joint reaction forces due to ligament tensions and bone-to-bone contacts) from the moment equilibrium equation applied to an appropriately selected mechanical subsystem delimited by that
joint. For example, the hip torque $\tau_{\text{hip}}$ is derived from the equilibrium moment equation for the system (WUB) constituted by the upper body UB (trunk, head, and upper limbs) and the weighted barbell W:

$$2\tau_{\text{hip}} = (y_W - y_{\text{hip}})2(R_s) + (x_{\text{c-w}} - x_{\text{hip}})(M_{\text{GR}} + M_W)g \cos \alpha - (y_{\text{c-w}} - y_{\text{hip}})(M_{\text{GR}} + M_W)g \sin \alpha$$  \hspace{1cm} (8)

Here, $M_{\text{UB}}$ is the UB mass, and $C_{\text{WUB}}$ is the WUB center of mass. The torque $\tau_{\text{LS}}$ on the lumbosacral joint is directly obtained by the right-hand-side of Equation (8), by simply replacing the whole trunk with the portion of the trunk above the lumbosacral joint, and the hip coordinates with the lumbosacral joint coordinates. As these joints are very close to each other, it is approximately $\tau_{\text{LS}} = 2\tau_{\text{hip}}$. The knee torque $\tau_{\text{knee}}$ is deduced from the same equilibrium equation for the lower leg and foot system (LF):

$$\tau_{\text{knee}} = -y_{\text{knee}}(A_{\text{GR}}) + (x_{\text{c-w}} - x_{\text{GR}})N_{\text{GR}} - (x_{\text{c-w}} - x_{\text{cLF}})m_{\text{LF}} g \cos \alpha + (y_{\text{c-w}} - y_{\text{knee}})m_{\text{LF}} g \sin \alpha$$  \hspace{1cm} (9)

where $m_{\text{LF}}$ is the LF mass, and $C_{\text{LF}}$ is the LF center of mass.

The determination of a joint reaction force (the force due to ligament tensions and bone-to-bone contacts) requires the selective knowledge of the force for each muscle crossing that joint. Unfortunately, in general, this information cannot be deduced from the knowledge of the joint torque, without the use of arbitrary optimization procedures.\textsuperscript{21} However, if the torque of the knee flexor muscles is negligible compared to the knee extensor torque provided by the quadriceps, then the patellar tendon force $F_{\text{PT}}$ can be given as the ratio of $\tau_{\text{knee}}$ to the patellar tendon moment arm $a_{\text{PT}}$:

$$F_{\text{PT}} \equiv \frac{\tau_{\text{knee}}}{a_{\text{PT}}}$$  \hspace{1cm} (10)

With this approximation, the patellofemoral joint reaction force $\phi$ is obtained analytically from the equilibrium condition for the external forces acting on the LF system

$$\phi + F_{\text{PT}} + A_{\text{GR}} + N_{\text{GR}} + m_{\text{LF}} g = 0$$  \hspace{1cm} (11)

Projection of this equation on the longitudinal shank axis, and on a normal to this axis, gives the compressive ($\phi_c$) and shear ($\phi_s$) components of $\phi$, respectively:\textsuperscript{22,23}

$$\phi_c = -F_{\text{PT}} \tan \gamma_{\text{PT}} + N_{\text{GR}} \cos(\theta_{\text{ankle}}) - (A_{\text{GR}}) \sin(\theta_{\text{ankle}}) - m_{\text{LF}} g \cos(\alpha + \theta_{\text{ankle}})$$

$$\phi_s = -F_{\text{PT}} \tan \gamma_{\text{PT}} - N_{\text{GR}} \sin(\theta_{\text{ankle}}) - (A_{\text{GR}}) \cos(\theta_{\text{ankle}}) + m_{\text{LF}} g \sin(\alpha + \theta_{\text{ankle}})$$  \hspace{1cm} (12, 13)

In Equations (12) and (13), $\gamma_{\text{PT}}$ is the patellar-tendon traction angle, that is the $\theta_{\text{ankle}}$-dependent angle of the vector $F_{\text{PT}}$, and the longitudinal shank axis. In this study, the well-known Herzog and Read\textsuperscript{24} polynomial functions are adopted for $a_{\text{PT}}(\theta_{\text{ankle}})$ and $\gamma_{\text{PT}}(\theta_{\text{ankle}})$.

It is commonly known that the patello-femoral compressive force $\phi_c$ increases with patellar-tendon force and degree of knee flexion.\textsuperscript{23} Thus, within the approximation of Equation (10), for any given value of $\theta_{\text{ankle}}$ (that sets the value of $a_{\text{PT}}$), $\phi_c$ is only determined by the knee torque $\tau_{\text{knee}}$, and comes to be a monotonic increasing function of $\tau_{\text{knee}}$.

In the following, we will direct our attention to the dependence of $\tau_{\text{hip}}$, $\tau_{\text{LS}}$, $\tau_{\text{knee}}$, $\phi_c$, $\phi_s$, and $\gamma_{\text{PT}}$ on the inclination angle $\phi$ of the Smith machine, for all the different body configurations corresponding to the following ranges of joint angles: $60^\circ \leq \theta_{\text{ankle}} \leq 100^\circ$, $90^\circ \leq \theta_{\text{ankle}} \leq 180^\circ$, $-10^\circ \leq \theta_{\text{ankle}} \leq 30^\circ$, and $70^\circ \leq \theta_{\text{hip}} \leq 180^\circ$ ($\theta_{\text{hip}} = 90^\circ + \theta_{\text{knee}} - \theta_{\text{ankle}} - \theta_{\text{radial}}$); and for different locations of the application point of the ground reaction force: just below the ankle joint ($X_{\text{GR}} = X_{\text{ankle}}$), and midway between the heel and the toe tip ($X_{\text{GR}} = X_{\text{ankle}} + 0.07m$). For comparison, the same joint torques and forces are also reported for a typical free barbell squat exercise ($\alpha = 0$, $A_{\text{GR}} = R_s = 0$ and $X_C = X_{\text{GR}}$). Maple software package (Maplesoft, Waterloo, Canada) was used for the calculation.

**Results**

The joint torques $\tau_{\text{knee}}$ and $\tau_{\text{hip}}$ (and $\tau_{\text{LS}}$) and the joint reaction forces $\phi_c$ and $\phi_s$ can be accurately controlled by selecting an appropriate overload resistance $M_W$, and by changing both the equipment inclination $\alpha$, and the parameters $\theta_{\text{ankle}}$, $\theta_{\text{knee}}$, $\theta_{\text{radial}}$, and $X_{\text{GR}}$ that define the body configuration within the Smith machine (Figures 3 to 8). The ratios of the quantities $\tau_{\text{knee}}$, $\tau_{\text{hip}}$, $F_{\text{PT}}$, $\phi_c$, and $\phi_s$, to the total resistance $M_g$, turn out to be nearly insensitive to the overload resistance $M_g$. For this reason, only the condition $M_W = M_B$ has been considered (Figures 3 to 8), and the forces $\phi_c$ and $\phi_s$ have been normalized to $M_B$ (Figures 4 to 8). In fact, a change in the ratio $M_W/M_B$ slightly influences the location of the centers of mass C and $C_{\text{WUB}}$, and, consequently, slightly changes the lever arms, about the joint axes, of the weight forces acting on WB and WUB systems.

For a given value of knee angle $\theta_{\text{ankle}}$, $|\phi_s|$ decreases and torque shifts from the knee to the hip joint, by increasing the backward equipment inclination or by reducing its forward inclination (increasing $\alpha$), bending the trunk...
Backward/Forward Inclined Smith Squat

forward (increasing $\theta_{\text{ankle}}$), moving the foot forward with respect to the knee (increasing $\theta_{\text{ankle}}$), and displacing the weight distribution towards the forefoot (increasing $x_{\text{GR}}$), and vice versa (Figures 3, 4, 7, and 8). The torques $\tau_{\text{knee}}$ and $\tau_{\text{hip}}$, and the intensity $|\phi_n|/M_g$ of the compressive tibiofemoral force, increase almost linearly when the knee flexion angle is increased ($\theta_{\text{knee}}$ is decreased) keeping constant the other parameters (Figures 3 and 4). Negative knee (hip) torques occurring for high (low) values $\theta_{\text{ankle}}$ indicate a knee (hip) flexor torque to be provided by the knee (hip) flexors. Notably, with changing the equipment inclination $\alpha$ from $-30^\circ$ to $30^\circ$, the peak values of $\tau_{\text{knee}}$ and $|\phi_n|/M_g$ (obtained for $\theta_{\text{knee}} = 90^\circ$, $\theta_{\text{ankle}} = 60^\circ$, $\theta_{\text{trunk}} = -10^\circ$, $x_{\text{GR}} = x_{\text{ankle}}$) reduce from 200 N·m to 150 N·m (Figure 3) and from 3.3 to 2.65 (Figure 4) respectively, whereas the peak value of $\tau_{\text{hip}}$ (obtained for $\theta_{\text{knee}} = 120^\circ$, $\theta_{\text{ankle}} = 100^\circ$, $\theta_{\text{trunk}} = 30^\circ$, $x_{\text{GR}} = x_{\text{ankle}} + 0.07m$) rises from 175 N·m to 225 N·m (Figure 7).

Figure 3 – Dependence of the knee and hip torques ($\tau_{\text{knee}}$ and $\tau_{\text{hip}}$) on the trunk inclination ($-10^\circ \leq \theta_{\text{trunk}} \leq 30^\circ$), for different values of knee angle ($\theta_{\text{knee}} = 150, 120, 90^\circ$), ankle angle ($\theta_{\text{ankle}} = 100, 90, 80, 70, 60^\circ$), and equipment inclination ($\alpha = -30^\circ$ and $30^\circ$) under the conditions $M_W = M_B$ and $x_{\text{GR}} = x_{\text{ankle}}$ (application point of the ground reaction located under the medial malleolus). Data are displayed only in the range $70^\circ \leq \theta_{\text{hip}} \leq 180^\circ$ ($\theta_{\text{hip}} = 90^\circ + \theta_{\text{knee}} - \theta_{\text{ankle}} - \theta_{\text{trunk}}$), and the circle points represent typical knee and hip torques for the free barbell squat.
The shear component $\phi_t$ of the tibiofemoral joint reaction force to anterior (posterior) drawer in non-weight-bearing cadaveric knees. We conventionally assume that a positive (negative) shear force $\phi_t$ opposes the posterior (anterior) translation of the tibial plateau with respect to the femur, reflecting a load on the PCL (ACL). Both ACL and PCL can be stressed appreciably in the range $180^\circ \geq \theta_{knee} \geq 130^\circ$, whereas only PCL stress occurs for $\theta_{knee} < 130^\circ$. Nevertheless, the ACL stress is completely suppressed.

The shear component $\phi_t$ of the tibiofemoral joint reaction force displays a non-monotonic dependence on $\theta_{knee}$ (Figures 5 and 6) and can be largely patterned by changing the equipment inclination $\alpha$. $\phi_t$ determines to a great extent the strain-forces on cruciate ligaments. For example, Butler et al.\(^{26}\) highlighted that the anterior cruciate ligament, ACL, (posterior cruciate ligament, PCL) provides 86\% (95\%) of the total restraining force

\[
\phi_t = \frac{M_W}{M_B} \gamma_G \theta_{ankle} = 60, 70, 80, 90, 100^\circ
\]

\[
\alpha = -30^\circ, \theta_{knee} = 120^\circ
\]

\[
\alpha = -30^\circ, \theta_{knee} = 90^\circ
\]

\[
\alpha = 30^\circ, \theta_{knee} = 120^\circ
\]

\[
\alpha = 30^\circ, \theta_{knee} = 90^\circ
\]
Figure 5 – Dependence of the shear component $\phi_t$ of tibiofemoral joint reaction force (normalized to the overall resistance $Mg$) on the trunk inclination ($-10^\circ \leq \theta_{\text{trunk}} \leq 30^\circ$), for different values of knee angle ($\theta_{\text{knee}} = 170, 160, \ldots, 90^\circ$), and ankle angle ($\theta_{\text{ankle}} = 100, 90, 80, 70, 60^\circ$), and for a backward equipment inclination ($\alpha = 30^\circ$), under the conditions $M_W = M_B$ and $x_{GR} = x_{\text{ankle}}$ (application point of the ground reaction located under the medial malleolus). Positive (negative) shear forces $\phi_t$ correspond to loads on the posterior (anterior) cruciate ligament, and the circle points represent typical $\phi_t$ values for the free barbell squat.
Figure 6 – Dependence of the shear component $\phi_t$ of tibiofemoral joint reaction force (normalized to the overall resistance $Mg$) on the trunk inclination ($-10^\circ \leq \theta_{\text{trunk}} \leq 30^\circ$), for different values of knee angle ($\theta_{\text{knee}} = 170, 160, \ldots, 90^\circ$) and ankle angle ($\theta_{\text{ankle}} = 100, 90, 80, 70, 60^\circ$), and for a forward equipment inclination ($\alpha = -30^\circ$), under the conditions $M_W = M_B$ and $x_{\text{GR}} = x_{\text{ankle}}$ (application point of the ground reaction located under the medial malleolus). Positive (negative) shear forces $\phi_t$ correspond to loads on the posterior (anterior) cruciate ligament, and the circle points represent typical $\phi_t$ values for the free barbell squat.
on the whole knee ROM, independently of the values of $\theta_{\text{knee}}$ and $\theta_{\text{ankle}}$, by squatting with $30^\circ$ backward equipment inclination, and displacing the center of foot weight distribution in forward direction (Figure 7), at least up to midway between the heel and the toe tip ($x_{\text{GR}} = x_{\text{ankle}} + 0.07m$). This last condition is obtained with a uniform weight distribution between forefoot and heel, as commonly suggested in free barbell squat guidelines. The strain-force on ACL is maximum for a backward equipment inclination ($x_{\text{GR}}$) to midway between the heel and the toe tip ($x_{\text{ankle}}$). For example, with a $30^\circ$ backward equipment inclination, the shear component $\phi$ of the tibiofemoral joint reaction force may be suppressed in the range $180^\circ \geq \theta_{\text{ankle}} \geq 120^\circ$ by selecting, for each value of $\theta_{\text{ankle}}$, one or more specific pairs of ankle and trunk angles (Figure 5). This compares favorably with the standard vertical Smith squat, where the corresponding range is $180^\circ \geq \theta_{\text{ankle}} \geq 130^\circ$.16 and also with open kinetic-chain leg extension exercises, where the condition $\phi = 0$ can be achieved only for $180^\circ \geq \theta_{\text{ankle}} \geq 140^\circ$ by a controlled displacement of the application point of the resistance along the shank during the exercise.29

The equipment inclination can also be effectively managed to minimize the force on specific joint structures. For example, with a $30^\circ$ backward equipment inclination, the shear component $\phi$ of the tibiofemoral joint reaction force may be suppressed in the range $180^\circ \geq \theta_{\text{ankle}} \geq 120^\circ$ by selecting, for each value of $\theta_{\text{ankle}}$, one or more specific pairs of ankle and trunk angles (Figure 5). This compares favorably with the standard vertical Smith squat, where the corresponding range is $180^\circ \geq \theta_{\text{ankle}} \geq 130^\circ$.16 and also with open kinetic-chain leg extension exercises, where the condition $\phi = 0$ can be achieved only for $180^\circ \geq \theta_{\text{ankle}} \geq 140^\circ$ by a controlled displacement of the application point of the resistance along the shank during the exercise.29

The conditions for the development of ACL strain-force in squat exercises is a subject of paramount importance in sport rehabilitation. A great amount of data has been collected over the years, with the use of different techniques, spanning from theoretical modeling to the surgical implants of strain gauges on the ACL. However, on the whole, these research activities give contrasting results, and a clear understanding of the problem is still lacking. In fact, negligible ACL forces were reported in several studies, whereas forces with widely variable intensities were reported in the range $180^\circ \geq \theta_{\text{ankle}} \geq 120^\circ$ in a comparable number of studies.2,3,13,32–36 We recently highlighted that, for the Smith squat exercise, these differences can be accounted for in terms of differences in $\theta_{\text{ankle}}$, $\theta_{\text{trunk}}$, and $x_{\text{GR}}$, and, most importantly, that the ACL loading can be nearly completely eliminated by squatting with increased forward trunk tilt and by displacing the weight distribution toward the forefoot.16 The present results indicate that a backward equipment inclination ($\alpha = 30^\circ$) further protects the ACL, and definitely eliminates the ACL strain-force, when associated with a forward foot weight distribution ($x_{\text{GR}} = x_{\text{ankle}} + 0.07m$), independently of trunk tilt and foot positioning (Figure 7).

It was previously recognized that some extreme body configurations allowed by the Smith machine may be dangerous for lower back and knees.16 The equipment inclination may even worsen these critical situations. For example, one might strengthen hip and back extensors with knee safety by squatting with backward inclined equipment, bending the trunk forward and moving the feet forward in front of the knees. However, this body configuration entails considerably high hip flexion angles, and the backward inclination of the equipment, due to the action of gravity, tends to further enhance the hip flexion. In this condition, the lumbar spine may dangerously compensate by flexing more than usual under loading, especially in the presence of inflexibility of hamstrings (when knees are nearly straight), gluteus maximus and adductor magnus (when knees are bent). Likewise, squatting with a forward inclined equipment, backward trunk tilt, and feet backward behind the knees, preserves the back joints and maximizes both the quadriceps involvement and the peak force on the knee joint structures. However, this body positioning induces a full hip extension, and the forward equipment inclination, due to the

Discussion

In quasi-static free barbell squat, the knee, hip, and ankle angles are closely related one another, and the torque is nearly equally shared between the knee and the hip (symbols ● and ○ in Figure 3). Conversely, in the Smith squat, each joint angle may be changed independently of the other, enabling a free modulation of the torque distribution among the joints: the torque may be intentionally lowered at the knee and enhanced at the hip and back, or vice versa, in comparison with the free barbell squat.16 A forward or backward inclination of the Smith machine greatly extends the range of possibilities for modulating the torque distribution among the joints. For example, with a forward (backward) equipment inclination and a backward (forward) foot weight distribution, the hip (knee) torque may be maintained exquisitely small in the whole range $180^\circ \geq \theta_{\text{ankle}} \geq 90^\circ$, selecting suitable values of the ankle and trunk angles. Furthermore, a change in equipment inclination $\alpha$ from $-30^\circ$ to $30^\circ$ (from $30^\circ$ to $-30^\circ$) gives a $33\%$ increment (25% decrease) in hip peak torque and a $25\%$ decrease (33% increment) in knee peak torque, when $x_{\text{GR}} = x_{\text{ankle}} \pm \theta_{\text{ankle}}$, $\theta_{\text{knee}}$, and $\theta_{\text{trunk}}$ span the selected ranges (Figure 3).
Figure 7 – Dependence of knee and hip torques ($\tau_{\text{knee}}$ and $\tau_{\text{hip}}$), and of axial and shear ($\phi_n$ and $\phi_s$) components of tibiofemoral joint reaction force (normalized to the overall resistance $Mg$), on the trunk inclination ($-10^\circ \leq \theta_{\text{trunk}} \leq 30^\circ$), for different values of knee angle ($\theta_{\text{knee}} = 150, 120, 90^\circ$) and ankle angle ($\theta_{\text{ankle}} = 100, 90, 80, 70, 60^\circ$), and for a backward equipment inclination ($\alpha = 30^\circ$), under the conditions $M_B = M_B$ and $X_{GR} = X_{\text{ankle}} + 0.07\,m$ (application point of the ground reaction located midway between the heel and the toe tip). Data are displayed only in the range $70^\circ \leq \theta_{\text{hip}} \leq 180^\circ$ ($\theta_{\text{hip}} = 90^\circ + \theta_{\text{knee}} - \theta_{\text{ankle}} - \theta_{\text{trunk}}$) and the circle points represent typical values for the free barbell squat.
Figure 8 – Dependence of knee and hip joint reaction forces (τ_knee and τ_hip), and of axial and shear (ϕ_n and ϕ_t) components of tibiofemoral joint reaction force (normalized to the overall resistance \( M_g \)), on the trunk inclination (−10° ≤ \( \theta_{\text{trunk}} \) ≤ 30°), for different values of knee angle (\( \theta_{\text{knee}} = 150, 120, 90° \)) and ankle angle (\( \theta_{\text{ankle}} = 100, 90, 80, 70, 60° \)), and for a forward equipment inclination (\( \alpha = −30° \)), under the conditions \( M_w = M_b \) and \( x_{GR} = x_{\text{ankle}} + 0.07 \) m (application point of the ground reaction located midway between the heel and the toe). Data are displayed only in the range 70° ≤ \( \theta_{\text{hip}} \) ≤ 180° (\( \theta_{\text{hip}} = 90° + \theta_{\text{knee}} - \theta_{\text{ankle}} - \theta_{\text{trunk}} \)), and the circle points represent typical values for the free barbell squat.
action of gravity, further enhances this condition. The lower back may dangerously hyperextend under loading, especially in the presence of hip flexor inflexibility and/or abdominal weakness.

One major limitation affects the present study. The specific contribution of hamstrings and gastrocnemius muscles has been neglected in Equations (12) and (13), that is, in the calculation of the shear and compressive components of tibiofemoral joint load (ϕ, and ϕt). However, in the Smith squat, these contributions are not realistically predictable, because the user may freely change the weight distribution between the foot and heel, due to the support given by the Smith machine, especially if the equipment is inclined. The qualitative effects of such muscle activations on ϕ and ϕt have been addressed in great detail in a previous work:16 the calculated values of the tibiofemoral reaction forces ϕ and ϕt only constitute reference lower limits for the compressive (ϕt) and PCL-loading (ϕt > 0) components, and reference upper limit for the intensity |ϕt| of the ACL-loading component (ϕt < 0). Actually, the current biomechanical study represents the necessary framework for further experimental investigations in which motion analysis, EMG, and force platform data will be simultaneously recorded. This will entail a huge experimental activity due to the number of parameters that can be changed independently of the other (Mt, ω, ϑ, θknee, θtrunk, xGR, and α) in the inclined Smith squat exercise.

Further limitations stem from the simplified biomechanical joint model adopted in this study. For example, a model which includes the soft tissue structures of the knee and the complex geometry of the articular surfaces would be necessary to know how the net tibiofemoral loads ϕt and ϕk are distributed across the different structures of the knee. Such information is needed to achieve a quantitative assessment of the strain-force on the cruciate ligaments.

The present model refers to a quasi-static Smith squat exercise. In dynamic squat, high joint accelerations and inertial effects associated with the overload may appreciably affect the values of joint torques and joint reaction forces. Typically, the joint torques and loadings are increased (decreased) at the beginning (end) of the concentric phase and at the end (beginning) of the eccentric phase of the exercise. However, these effects can be quantified only on a case-by-case basis, and do not affect the general trends set by the equipment inclination (Figures 3–8), which are the subject of this article.

In conclusion, a controlled change of inclination of the Smith squat equipment can be effectively used to unload specific joint structures while focusing the activation on selected muscle groups. The information on the equipment inclination can be integrated with those concerning body positioning and foot weight distribution to get a complete set of indications for a full and careful use of the Smith squat in strengthening and rehabilitation programs. These indications are summarized in the following.

Patellar tendon force (Fpt), knee torque (τknee), patellofemoral (ϕk) and tibiofemoral (ϕt) compression forces increase with decreasing ϑknee, ϑtrunk, ϑankle, xGR, and α (inclining the equipment forward). The hip torque (τhip) and the spine torque occurring at the lumbar sacral joint (τls) increase with decreasing ϑknee and with increasing ϑtrunk, ϑankle, xGR, and α (inclining the equipment backward).

The ACL and PCL strain-force may be simultaneously suppressed (ϕt = 0) in the range 180° ≥ ϑknee ≥ 120° by a backward equipment inclination (α = 30°) and selecting, for each value of ϑknee, one or more specific pairs of ankle and trunk angles. The ACL loading can be definitely eliminated by squatting with a backward equipment inclination (α = 30°) and at the end of the concentric phase and at the beginning (end) of the eccentric phase of the exercise.

For small (large) knee flexion angle, in the range 180° ≥ ϑknee ≥ 150° (ϑknee < 130°), the PCL strain-force decreases (increases) with decreasing ϑknee, ϑtrunk, ϑankle, and α, that is, inclining the equipment forward. In the range 150° > ϑknee ≥ 130°, the PCL strain-force still decreases by decreasing α (inclining the equipment forward). The peak value of both ACL and PCL loading, occurring in the range 180° ≥ ϑknee ≥ 90°, is minimized by a backward equipment inclination.

For any given value of the equipment inclination α, the ratios to the total resistance Mg of the calculated torques (τknee, τhip) and joint reaction forces (ϕk, and ϕt) are nearly insensitive to the value of overload resistance Mg, at least for Mg ≤ 2Mg. Thus, Figures 3 through 8 actually give the trends of τknee, τhip, ϕk, and ϕt for any real value of overload resistance.

**References**


