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Kinematic and kinetic analysis of knee joint during squatting

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Abstract: Knee function in squatting is an important consideration for activity of daily living. However, there are only a limited number of studies reporting biomechanics of deep knee flexion beyond 90°. The joint and muscle forces calculated from the data collected from six healthy subjects performing squatting and rising from a deep squat using a force plate, a marker system and EMG. The estimated tibio-femoral joint forces were between 4.7 and 5.6 times body weight in vertical direction and 2.9-3.5 times body weight in horizontal direction. These forces are much larger than the forces during normal walking. A mathematical model shows the external forces acting on the segments of lower limbs are the gravity forces and inertia forces due to mass segments of limb and the reactive forces transmitted from the ground. The ground reaction forces have the enormous effect on the leg during stance phase of walking whereas the gravity forces and the inertia forces are predominantly responsible for the forces at joints during swing phase of walking. The moment developed by the external forces at the respective joint is counterbalanced by the moment developed by the contraction of muscle forces about the joint to keep the joint in dynamic equilibrium.

Keywords
kinematics, kinetics, squatting, ground reaction forces, lower limbs

1. Introduction

Many countries peoples are in the habit of squatting posture for many activities starting from the use of toilet to farming operation, relaxation and even during their daily prayer to Almighty (Akagi 2005, Mulholland 2001). Therefore, along with normal walking squatting is an important consideration of a person for their daily activity. Studies carried out on squatting posture for its kinematics and kinetics action are very limited. Murrey (1967) carried out some studies on the ground reaction forces and position of body centre of gravity during squatting activity. Natarajan et al. (1974) studied some aspect of muscular activity during squatting. With the electromyographic study they observed there exists a correlation between the activity of tibialis anterior muscle with the shift of centre of gravity of body during squatting. The posterior shift is accompanied by persistent activity of the tibialis anterior and when the shift towards the front, the activity of the muscles stops. Dahlkvist et al. (1982) had studied the forces developed in different muscles of lower limb acting in sagittal plane during squatting and rising from deep squat. They came to the conclusion that patella-femoral forces are the largest and the values varies between 4.7 times the body weight during slow ascent and 7.6 times body weight during fast descent. Brechter JH et al. (2002) demonstrated greater joint stress during stair ascent and descent for different subjects with patellofemoral pain and without patellofemoral pain. Brechter JH, Christine E. Draper (2005) estimated patellofemoral joint contact areas in a group of healthy, pain-free subjects during upright, weight-bearing conditions. Sixteen subjects (8 female, 8 male) were scanned by them in a GE Signa SP open configuration MRI scanner, which allowed subjects to stand or squat while reclining...
25° from vertical with the knee positioned at 0°, 30°, or 60° of flexion. Male subjects displayed mean unloaded patellofemoral joint contact areas of 210, 414, and 520 mm² at 0°, 30° and 60° of knee flexion, respectively. Vucina, A et al. (2005) stated that during upstairs movement of person with AK prosthesis lies in a need to introduce an external source of energy, which would provide the user with energy required for lifting a body when climbing the stairs. Takeo Nagura et al. (2002) calculated the knee biomechanics in deep flexion beyond 90°. In that study, authors found that the large moments and forces will result in high stress at high angles of flexion. Moreover, the peak net moments and the net posterior forces were generated between 90° and 150° of flexion. These loads can influence pathological changes to the joint and are important considerations for reconstructive procedures of the knee.

2. Kinematic squat analysis
It is observed that during squatting, from standing posture, the leg rotates laterally about 30-35 degrees from the sagittal plane. The ankle and hip joints provide three degrees of freedom, thus ankle and hip remaining in the same vertical plane, the knee joint moves laterally during squatting posture (Figure 1).

![Figure 1 A typical stick diagram showing the pattern of thigh and shank motion during squatting](image)

3. Mathematical Modelling of lower limb
It is assumed that the lower limb consists of three rigid points-thighs, shank and foot which are connected at joint by single hinge joint and internal muscle forces are acting about this joint to counterbalance the moment due to external forces. The relationship of the internal forces and moments developed at ankle, knee and hip joint during walking and squatting are determined in terms of the weight of the segments, moment of inertia of the segments about its centroidal axis normal to sagittal plane, accelerations of the proximal joint of the segment and the orientation, angular velocity and angular acceleration of the segments. Since the study has been done on the left leg of subjects walking from right to left the axes have been chosen positive from right to left for motion along the plane of progression. All the absolute rotations and the derivatives for the respective segments have been taken positive for clockwise rotation measured from the positive axis (which is along the plane of progression i.e., from right to left).

The derivation of the relationship for forces and moments at the joints is given in detail in Appendix A. The components of forces along the positive direction of axes and clockwise moment at joints are taken to be positive. The dynamic equations in these final for the segments foot, shank and thigh are written as follows:

3.1 Hip Joint Model:
Component of joint reaction forces $X_H$ and $Y_H$ in figure 2 along plane of progression x and vertical y respectively are given by

$$X_H = X_k + \frac{w_r}{g} \left[ X_H - \ddot{\theta}_H \left( r_r \cos \theta_H \right) - \ddot{\theta}_H \left( r_r \sin \theta_H \right) \right]$$

----------1
\[
Y_H = Y_K + W_T + \frac{W_T}{g} \left[ \dot{y}_H - \dot{\theta}_H \left( r_T \sin \theta_H \right) + \ddot{\theta}_H \left( r_T \cos \theta_H \right) \right] \quad \text{---2}
\]

And joint moment \(M_H\) is given by Figure 2
\[
M_H = M_K - X_K (l_T \sin \theta_H) + Y_K (l_T \cos \theta_H) + W_T (r_T \cos \theta_H) + \left( I_T + \frac{W_T}{g} r_T^2 \right) \ddot{\theta}_H \\
- \frac{W_T}{g} \left[ \ddot{x}_H (r_T \sin \theta_H) - \dddot{y}_H (r_T \cos \theta_H) \right] \quad \text{---3}
\]

where, \(W_T\) is the weight of the shank segment, \(\ddot{x}_H, \dddot{y}_H\) are the component of translation acceleration of knee joint along axes of reference, \(\dot{\theta}_H, \theta_H\) and \(\ddot{\theta}_H\) are the angular rotation, angular velocity and angular acceleration of axis of shank measured clockwise from x direction, \(I_T\) is the length of shank, \(r_T\) is the distance of mass center of shank from its proximal joint and \(I_T\) is the mass moment of inertia of the shank about its centroidal axis.

3.2 Knee Joint Model:

Force components \(X_K\) along plane of progression x and \(Y_K\) in figure 3 along vertical y are given by,
\[
X_K = X_{AN} + \frac{W_s}{g} \left[ \dot{X}_K - \dot{\theta}_S \left( r_s \cos \theta_S \right) - \dot{\theta}_S (r_s \sin \theta_S) \right] \quad \text{---4}
\]
\[
Y_K = Y_{AN} + W_S + \frac{W_s}{g} \left[ \dot{Y}_K - \dot{\theta}_S \left( r_s \sin \theta_S \right) + \ddot{\theta}_S (r_s \cos \theta_S) \right] \quad \text{---5}
\]
and joint moment \(M_K\) is given by
\[
M_K = M_{AN} - X_{AN} (l_S \sin \theta_S) + Y_{AN} (l_S \cos \theta_S) + W_S (r_S \cos \theta_S) \\
+ \left( I_S + \frac{W_s}{g} r_s^2 \right) \ddot{\theta}_S - \frac{W_s}{g} \left[ \ddot{x}_K (r_S \sin \theta_S) - \dddot{y}_K (r_S \cos \theta_S) \right] \quad \text{---6}
\]

Where, \(W_s\) is the weight of the shank segment, \(\ddot{x}_K, \dddot{y}_K\) are the component of translation acceleration of knee joint along axes of reference, \(\dot{\theta}_S, \theta_S\) and \(\ddot{\theta}_S\) are the angular rotation, angular velocity and angular acceleration of axis of shank measured clockwise from x direction, \(I_S\) is the length of shank, \(r_s\) is the distance of mass center of shank from its proximal joint and \(I_S\) is the mass moment of inertia of the shank about its centroidal axis.
3.3 Ankle joint Model:

Force components $X_{AN}$ along plane of progression $x$ and $Y_{AN}$ in figure 4 along vertical direction $y$ are given by,

$$X_{AN} = -X_G + \frac{W_F}{g}[\ddot{X}_{AN} + \dot{\theta}_F^2 (r_F \cos \theta_F) - \ddot{\theta}_F (r_F \sin \theta_F)] \quad --------7$$

$$Y_{AN} = -Y_G + W_F + \frac{W_F}{g} [\ddot{Y}_{AN} - \dot{\theta}_F^2 (r_F \sin \theta_F) + \ddot{\theta}_F (r_F \cos \theta_F)] \quad --------8$$

And moment at the joint $M_{AN}$ is given by

$$M_{AN} = -X_G Y_{AN} - Y_G (X_G - X_{AN}) + W_F (r_F \cos \theta_F) + [I_F + \frac{W_F}{g} r_F^2] \dot{\theta}_F - \frac{W_F}{g} (\ddot{X}_{AN} (r_F \sin \theta_F) - \ddot{Y}_{AN} (r_F \cos \theta_F)) \quad --------9$$

Where, $X_G$, $Y_G$ are longitudinal and vertical components of ground reaction forces, $W_F$ is the weight of the foot, $X_{AN}$, $Y_{AN}$ are coordinated ankle joint, $\ddot{X}_{AN}$, $\ddot{Y}_{AN}$ are the translation accelerations of the ankle joint along axes, $X_G$ is the longitudinal coordinate of ground reaction force, $r_F$ the distance of mass center of foot from proximal joint of foot, $\dot{\theta}_F$, $\theta_F$ and $\ddot{\theta}_F$ are angular rotation, angular velocity and angular acceleration of foot axis with reference to the forward direction, $I_F$ is the mass moment of inertia of foot about its centroidal axis and $g$ is the acceleration due to gravity.
In above equations, the instantaneous values of linear acceleration of joint and angular velocities and angular accelerations of segments, obtained from kinematic analysis. The anatomical data for weight of segments, position of mass center from the proximal joints and their mass moment of inertia about the respective mass centers are determined from the graphs and tables given by Contini (1972) by knowing the height and weight of the subjects. Herman I.P. (2016) also presented human moment of inertia and the stability equations during squatting. The components of GRF (Ground reaction forces) are obtained from the data recorded on the force platform.

4. Results and discussions

Table 1 Lower limb segment parameters of normal subjects

| Subject code | Height (m) | Weight (N) | Weight (N) | Length (m) | Distance of mass centre from proximal joint (m) | Mass moment of inertia about centroidal axis normal to sagittal plane (Kg-m²) |
|--------------|-----------|------------|------------|------------|-----------------------------------------------|---------------------------------------------|
| F            | S         | T          | F          | S          | T                                              | F                                           | S   | T   | F    | S    | T   |
| S1           | 1.74      | 680        | 9.12       | 28.38      | 62.43                                          | .426                                        | .402 | .157 | .177 | .0028 | .0415 | .0812 |
| S2           | 1.75      | 546        | 7.31       | 22.53      | 49.90                                          | .429                                        | .404 | .158 | .178 | .0023 | .0334 | .0655 |
| S3           | 1.60      | 574        | 7.64       | 23.78      | 52.34                                          | .392                                        | .369 | .144 | .163 | .0020 | .0293 | .0661 |
| S4           | 1.71      | 467        | 6.26       | 19.50      | 45.92                                          | .416                                        | .392 | .153 | .173 | .0018 | .0272 | .0526 |
| S5           | 1.68      | 485        | 6.31       | 19.69      | 43.02                                          | .411                                        | .388 | .151 | .171 | .0018 | .0268 | .0521 |

F=FOOT; S=SHANK; T=THIGH   S1; S2; S3; S4; S5=SUBJECTS

The GRF variation is curves are divided into parts one is longitudinal component (X₀) and vertical component (Y₀). For any instant the components of X₀ and Y₀ can estimated from the graph as shown in figure.

5. Squatting Kinetic Analysis

During kinetic study of squatting it is observed that vertical GRF decreases initially due to downward acceleration of body centre of the gravity and then it increases and reaches a peak value at instance of hip strikes the shank to stop the motion. The internal moment developed at the hip, knee and ankle joint due to external ground reaction forces, gravity forces and forces due to inertia of body segments for eight normal subjects. During initial phase of squatting the line of action of GRF moves posterior from its initial standing position and the magnitude of the GRF also decreases due inertia forces acting on the body C.G up to about 20% of the squatting time.
Figure 5 Typical patterns of components of ground reaction forces and variation of co-ordinate of point of application of GRF during normal walking of the subjects

Subsequently, the point of application of GRF moves forward and magnitude of GRF starts increasing and become almost same as that in the standing posture at about 50% of squatting time. Then with the downward deceleration of body C.G, the inertia forces acts downward and magnitude of ground reaction forces increases (figure 5).

Figure 6 Variation of hip, knee and ankle moments during squatting for four normal subjects.
The variation of the joint moments during squatting posture is shown in (Figure 6) with the increase in the lever arm and the magnitude of forces, the joint moments starts increasing after 29% of the squatting time. The maximum mean extension movements developed at the hip joint and knee joint are found to be \(-56\pm 22\) (Nm) and \(-103\pm 26\) (Nm) respectively. The mean maximum ankle plantarflexion moment is \(27\pm 10\) (Nm). Thus, moment developed at the knee joint to be greatest in comparison with moments at other joints. The Joint moments show that the magnitude is mostly dependent on the duration of squatting. The increase in the extensor force during squatting will increase the stress on the patellar tendon and joint contact forces. In conclusion, large net quadriceps moments and net posterior forces at the knee were seen during the deep flexion activities. The results indicate that the loads on the knee during squatting are important considerations both for the design of the knee linkage mechanism.

\[\text{Table 2: Comparison of the squat and walking}\]

|                  | Walking     | Squatting  | P-value |
|------------------|-------------|------------|---------|
| Knee Flexion Moment(%BW*Ht) | 2.3(1.2)    | 6.6(1.2)   | <0.001  |
| Knee adduction Moment(%BW*Ht) | 2.3(0.6)    | 1.4(0.8)   | 0.01    |
| Peak lateral Compartment Knee Force(N) | 772(279)   | 1417(616)  | 0.002   |
| Peak Medial Compartment Knee Force(N) | 1343(247)  | 1468(405)  | 0.27    |
| Peak Net Anterior Shear Force(%BW) | 36.6(5.7)  | 42.8(6.7)  | 0.06    |

It is found that the knee flexion moment during the squat was almost three times as large as that seen during walking. (P<0.001, Table 2). Conversely the adduction moment during squatting was significantly lower (P=0.02, table 2) than during level walking. During squatting anterior shear force also increased than that of normal walking.

**Conclusion**

This work extends the analysis of lower limb biomechanics during normal walking and squatting. Prosthetic knee design should be capable to cover a high-flexion and allow deep knee flexion without affecting the stability and durability of the joint. Knee forces during squatting are extensively studied. It was found that the forces tibio-femoral joint are between 4.7 and 5.6 times body weight which are very high as compared to the normal walking. So the prosthetic components designer has to apply high factor of safety while designing a prosthetic knee joint. A mathematical model developed also gives the indication that the knee flexion moment is three times as compared to walking and lateral compartment knee force is 1417N in squatting which is very high and can damage the prosthetic knee joint.

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**APPENDIX**

**Derivation of the force and moment at a point on a rigid body under motion**

The expressions for forces and moments developed at hip, knee and ankle joints are obtained from the following derivation for motion of rigid body under external forces and moment acting at a point. Considering the rigid body figure of weight acting at its centroid C to be moving in a two dimensional space x-y under force components X and Y and clockwise moment M acting at a point A on the body.

The acceleration of point C can be expressed in terms of acceleration of the point A is

\[
\ddot{\ell} = \ddot{a} + \ddot{\theta} \times \Delta \ell + \dddot{\theta} (\Delta \ell \times \dot{\ell}) \quad \text{(A.1)}
\]

In the scalar form

\[
\ddot{\ell}_x = \ddot{a}_x - \dot{\theta}^2 (r \cos \theta) - \dot{\theta} (r \sin \theta) \quad \text{(A.2)}
\]

\[
\ddot{\ell}_y = \ddot{a}_y - \dot{\theta}^2 (r \sin \theta) + \dot{\theta} (r \cos \theta) \quad \text{(A.3)}
\]

Where \( \ddot{\ell}_x, \ddot{\ell}_y \) are x and y components of accelerations of centroid C; \( \ddot{a}_x, \ddot{a}_y \) are x and y component of acceleration of point A; \( r \) is the distance between A and C; \( \dot{\theta} \) is the rotation of rigid body about A clockwise with positive ‘x’ direction and \( \ddot{\theta} \) are the angular velocity and angular acceleration of the rigid body about A.

![Figure 7 Rigid body in motion under external forces and moments](image)

Considering the dynamic equilibrium of the rigid body (Figure 7) and taking moment of all the forces about the point A, we have moment equation given by:

\[
-\bar{F} + (\ddot{\ell} - \dot{\ell}) x \bar{W} + I_c \ddot{\theta} + (\dddot{\ell} - \ddot{\theta}) x \frac{W}{g} \ddot{\ell} = 0 \quad \text{(A.4)}
\]

In scalar form equation (A.4) becomes

\[
M = Wr \cos \theta + I_c \ddot{\theta} - \frac{W}{g} \ddot{\ell}_y (r \cos \theta) - \frac{W}{g} \ddot{\ell}_x (r \sin \theta) \quad \text{(A.5)}
\]

Substituting the value of \( \ddot{\ell}_x \) and \( \ddot{\ell}_y \) in terms of \( \ddot{a}_x \) and \( \ddot{a}_y \) from equation (A.2) and (A.3) respectively,

\[
M = [I_c + \frac{W}{g} r^2] \ddot{\theta} + Wr \cos \theta - \frac{W}{g} \ddot{a}_x (r \sin \theta) - \frac{W}{g} \ddot{a}_y (r \cos \theta) \quad \text{(A.6)}
\]

Equating the net forces acting on the body with inertia forces acting on the body with inertia forces, the force components X and Y acting at point A can be written as:

\[
X = \frac{W}{g} \ddot{\ell}_x \quad \text{----------------------(A.7)}
\]

\[
Y = \frac{W}{g} \ddot{\ell}_y + W \quad \text{----------------------(A.8)}
\]

Substituting for \( \ddot{\ell}_x \) and \( \ddot{\ell}_y \) from equations (A.2) and (A.3)

\[
X = \frac{W}{g} \left[ \ddot{a}_x - \dot{\theta}^2 (r \cos \theta) - \dot{\theta} (r \sin \theta) \right] \quad \text{(A.9)}
\]

\[
Y = W + \frac{W}{g} \left[ \ddot{a}_y - \dot{\theta}^2 (r \sin \theta) + \dot{\theta} (r \cos \theta) \right] \quad \text{----------------------(A.10)}
\]