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RESEARCH ARTICLE

TORQUE ANALYSIS OF DETERMINING THE KNEE JOINT INJURY

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ABSTRACT

This paper describes mathematical model to determining the knee joint injury using kinematics knee model. The model consists of rotations and translations vectors. The result modeling maximum torque is angle 70º knee joint injury,in the present study, there significant difference between angles 60ºand 90º.

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INTRODUCTION

In recent years, with the advances in medicine, physiology, biology and the development of mechanics, information theory, cybernetics, the study of people's life system has stepped into an in-depth development stage. It has important meaning not only for understanding the laws of human motion and discovering the coordination of limbs between decision-making, but also in the fields of clinical diagnosis, medical rehabilitation (Song, 2009; Senden, 2009; Zhongwu, 2002; Xiangping Li, 2012], ergonomics, sports science, bionic mechanism and humanoid robot to detect and analyze human motion, as well as research the gait model of knee. Modeling dynamical knee joint available in the literature (Moeinzadeh, 1981, 1983]. The knee is more susceptible to twisting or stretching injuries (hyper flexed/hyperextended), taking the joint through a greater range of motion than it was meant to tolerate. If the knee is stressed from a specific direction, then the ligament trying to hold it in place against that force can stretch or tear. Twisting injuries to the knee put stress on the cartilage or meniscus and can pinch them between the tibial surface and the edges of the femoral condyle, potentially causing tears. Injuries of the muscles and tendons surrounding the knee are caused by acute hyperflexion or hyperextension of the knee or by overuse. Strains are graded similarly to sprains, with first-degree strains stretching muscle or tendon fibers but not tearing them, second-degree strains partially tearing the muscle-tendon unit, and third-degree strains completely tearing it. Anatomically, many of the structures that support the knee are interconnected. A knee that is injured may cause damage to one or more structures depending upon the mechanism. Contact between the femur and the tibia, is not considered a real mathematical model that predicts knee response under dynamic loading. Most of the remaining dynamic models can be perceived as different versions of a single dynamic model. Such a model is comprised of two rigid bodies: a fixed femur and a moving tibia connected by ligamentous elements and having contact at a single point. In this paper will study dynamical analysis to determining the knee joint injury.

Kinematic of Knee: Six quantities are used to fully describe the relative motions between moving and fixed rigid bodies: three rotations and three translations. These rotations and translations are the components of the rotation and translation vectors, respectively.

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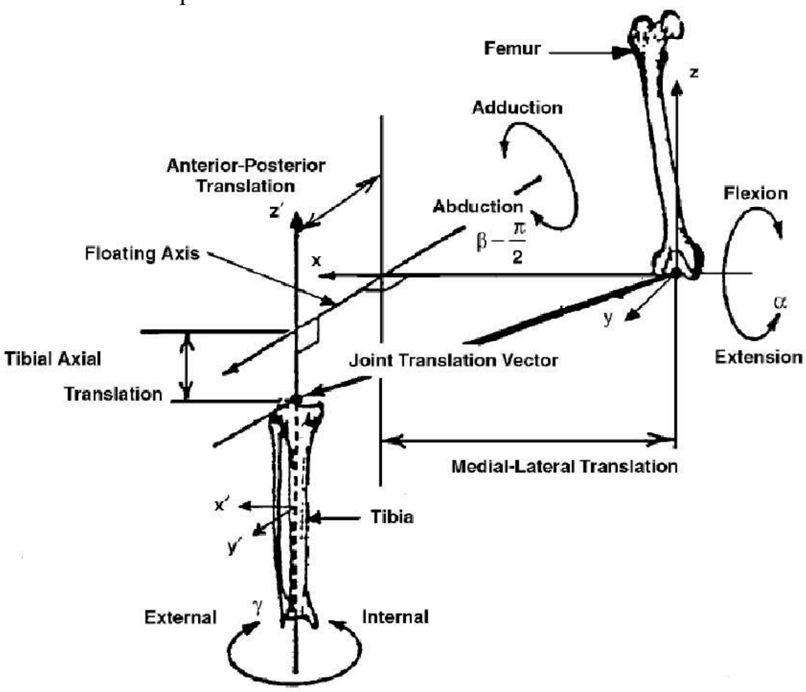
The three rotation components describe the orientation of the moving system of axes (attached to the moving rigid body) with respect to the fixed system of axes (attached to the fixed rigid body). The three translation components describe the location of the origin of the moving system of axes with respect to the fixed one.



The joint coordinate system is shown consist of an axis (x-axis) that is fixed on femur (i is a unit vector parallel to the x-axis), an axis (z-axis) that is fixed on the tibia (k' parallel to the z' axis), and and a floating axis perpendicular to these two fixed axes ($\vec{e}_2$ is a unit vector parallel to the floating axis). The three components of the rotation vector includeflexion-extension, tibial internal-external, and varus-valgus rotations. Flexion-extension rotations, α, occur around the femoral fixed axis; internal-external tibialrotations, γ , occur about the tibial fixed axis; and varus-valgus rotations, β , (ad-abduction)occur about the floating axis. The rotation vector $\vec{\theta}$is written as

$$\vec{\theta} = -\alpha\vec{\imath} - \beta\vec{e}_2 - \gamma k'$$

The six parameters (three rotations and three translations) describing tibio-femoral motions are used to determine the transformation between the two coordinate systems

$$\vec{R} = \vec{R}_0 + [R]\vec{r}$$

where $\vec{r}$describes the position vector of a point with respect to the tibialcoordinate system, and $\vec{R}$ describes the position vector of the same point with respect to the femoral coordinate system. The vector $\vec{R}_0$ is the position vector which locates the origin of the tibial coordinate system with respect to the femoral coordinate system, and (*R*] is a (3×3) rotation matrix

$$[R] = \begin{bmatrix} \sin\beta\cos\gamma & \sin\beta\cos\gamma & \cos\beta \\ -\cos\alpha\sin\gamma - \sin\alpha\cos\beta\cos\gamma & \cos\alpha\cos\gamma - \sin\alpha\cos\beta\sin\gamma & \sin\alpha\sin\beta \\ \sin\alpha\sin\gamma - \cos\alpha\cos\beta\cos\gamma & -\sin\alpha\cos\gamma - \cos\alpha\cos\beta\sin\gamma & \cos\alpha\sin\beta \end{bmatrix}$$

where α is the knee flexion angle, γ is the tibial external rotation angle, and β is(π/2 ± abduction); the positive sign indicates a right knee and negative sign indicates a left knee. A separate mathematical function was determined as an approximate representation for each ofthe medial femoral condyle, the lateral femoral condyle, the medial tibial plateau, and the lateral tibial plateau. The femoral articular surfaces are approximated as parts of spheres, while the tibial plateaus are considered as planar surfaces. The equation of the medial and lateral femoral spheres expressed in the femoral coordinate system of axes is written as

$$f(x, y) = -\sqrt{r^2 - (x-h)^2 - (y-k)^2} + 1$$

where values of parameters (*r, h, k*, and *l*) are obtained as 21, 23.75, 18.0, 12.0 mm and 20.0, 23.0, 16.0, 11.5 mm for the medial and lateral spheres, respectively. The equation of the medial and lateral tibial planes expressed in the tibial coordinate system of axes is written as

$$g(x', y') = my' = c$$

where values of parameters (*m, c*) are obtained as 0.358, 213 mm and –0.341,212.9 mm for the medial and lateral planes, respectively. Translation motion model accommodates two situations: a two-point contact and a single point contact. Initially, a two-point contact situation is assumed with the femur and tibia in contact on both medial and lateral sides. In the calculations, if one contact force becomes negative, then the two bones within its compartment are assumed to be separated, and the single-point contact situation is introduced, thus maintaining contact in the other compartment. The contact condition requires that the position vectors of each contact point in the femoral and the tibial coordinate system, $\vec{T_c}$ and $\vec{t_c}$, respectively, satisfy as follows

$$\vec{T_c} = \vec{T_0} + [T]\vec{r_c}$$

Where

$$\vec{T_c} = x_c\vec{i} + y_c\vec{j} + z_c\vec{k},$$
$$\vec{t_c} = x'_c\vec{i'} + y'_c\vec{j'} + z'_c\vec{k'}$$

where x_c, y_c, z_c and x'_c, y'_c, z'_c are the coordinates of the contact points in the femoral and tibial systems, respectively. Since contact occurs at points identifiable in both the femoral and tibial articulating surfaces, we can write at each contact point

$$z_c = f(x_c, y_c),\ z'_c = g(x'_c, y'_c)$$

there $f(x_c, y_c)$ and $g(x'_c, y'_c)$ are given by eqs.(3.1) and (3.2), respectively. Eq.(3.3) can thus be rewritten as three scalar equations

$$x_c = x_0 + T_{11}x'_c + T_{12}y'_c + T_{13}\,g(x'_c, y'_c)$$
$$y_c = y_0 + T_{11}x'_c + T_{12}y'_c + T_{12}\,g(x'_c, y'_c)$$
$$f(x_c, y_c) = z_0 + T_{31}x'_c + T_{32}y'_c + T_{33}\,g(x'_c, y'_c)$$

where T_{ij} is the ij-the component of the rotational transformation matrix $[T]$

Cross product of these two tangent vectors is then employed to determine the unit vector normal to the femoral surface, $\vec{n_f}$, at the contact point. Using eq.(3.5) we have

$$\vec{n_f} = \frac{\left.\frac{\partial f}{\partial x}\right|_{(x_c, y_c)}\vec{i} + \left.\frac{\partial f}{\partial y}\right|_{(x_c, y_c)}\vec{j} - \vec{k}}{\sqrt{1 + \left(\frac{\partial f}{\partial x}\right)^2 + \left(\frac{\partial f}{\partial y}\right)^2}}, \quad (x, y) = (x_c, y_c)$$

Unit vector normal to the tibial surface, $\vec{n'}_t$

$$\vec{n'}_t = \frac{\left.\frac{\partial g}{\partial x'}\right|_{(x'_c, y'_c)}\vec{i} + \left.\frac{\partial g}{\partial y'}\right|_{(x'_c, y'_c)}\vec{j} - \vec{k}}{\sqrt{1 + \left(\frac{\partial g}{\partial x'}\right)^2 + \left(\frac{\partial g}{\partial y'}\right)^2}}, \quad (x', y') = (x'_c, y'_c)$$

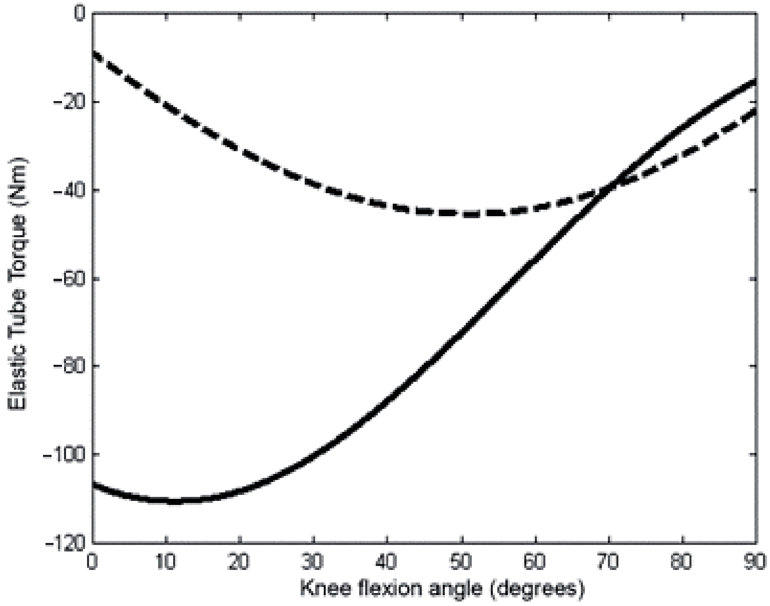
Applying the rotational transformation matrix to eq. (3.8) yields the unit normal vector to the tibial surface, $\vec{n_t}$, expressed in the femoral coordinate system as

$$\vec{n_t} = (T_{11}n'_{tx} + T_{12}n'_{ty} - T_{13}n'_{tz})\vec{i} + (T_{21}n'_{tx} + T_{22}n'_{ty} - T_{23}n'_{tz})\vec{j} + (T_{31}n'_{tx} + T_{32}n'_{ty} - T_{33}n'_{tz})\vec{k}$$

Since the unit vectors normal to the surfaces of the femur and tibia are coli near, they are equal $\vec{n_t} = \vec{n_f}$. The scalar form of this vectorial equation represents the geometric compatibility conditions at each contact point.

RESULTS

It is well established that maximal torque output of the quadriceps muscle group is dependent on knee angle (Becker, 2001; Williams, 1959]. It has previously been shown that maximal of the quadriceps occurs at a knee angle of 70 flexions. Becker and Awiszus demonstrated an increase in maximal torque production between angles 35º to 70º which was replicated in the current investigation with an increase in maximum torque output from angle 15º to 60º. Although angle 70º was not tested in the present study, there was no significant difference between angles 60º and 90º, which were tested.



Graph 1. Torque vs knee angle

The lack of difference at 60º and 90º coincides with (Marginson, 2001] who demonstrated that maximum torque production was equal at angles of 60º, 80º, and 90º. There could be many reasons for a nonmonotonic increase in maximum torque production, reaching a peak at 60ºfollowed by a decline. Many attribute the increase in maximal torque production to the increase in sarcomere length up to the optimal length that allows for maximal cross-bridge. Because there is a decrease in torque after the optimal length, it is maintained that the sarcomere length is stretched beyond optimum length and no longer allows maximal cross-bridge attachment (Desbrosses et al., 2006]. Despite the logic of this argument, there is little evidence that in intact quadriceps muscles sarcomere length has any influence. Other possible explanations include an inability to fully activate the quadriceps at shorter muscle lengths and interactions between monosynaptic excitation and muscle length (Becker, 2001]. Becker and Awiszus (Becker, 2001] suggested that an increase of the discharge rate resulting from stretching of the muscle spindles of the quadriceps in the greater joint angles leads to an increase in excitatory drive an increase in torque output.

REFERENCE

Song, Y. and Hong, Y. Y. 2009. "Design of an exoskeleton knee based on parallel mechanism for lower limb rehabilitation," Chinese High Technology Letters, vol. 08, 2009, pp. 367-372.

Senden R, Grimm B and Heyligers IC. 2009. "Acceleration-based gait test for healthy subjects: reliability and reference data," Gait and Posture, vol. 30, pp. 192-196.

Zhongwu Guo and Guangzhi Wang, 2002. "Research on kinematic parameters of gait in normal young people," Rehabilitation theory and practice of China, vol. 18 , pp. 632-534.

Xiangping Li, Bin Shu And Xiaohong Gu, 2012. "Temporal-spatial parameters of gait: reference data of normal subjects from Chinese adults," Chinese Journal of Rehabilitation Medicine, vol. 27, pp. 227-230.

Moeinzadeh, M.H. 1981. *Two- and three-dimensional dynamic modeling of the human joint structures with special application to the knee joint*, Ph.D. dissertation, Ohio State University,Columbus, OH.

Moeinzadeh, M.H., Engin, A.E., Akkas, N. 1983. *Two-dimensional dynamic modeling of the human knee joint*, J. Biomechanics, 16, p. 253.

Moeinzadeh, M.H., Engin, A.E. 1983. *Response of a two-dimensional dynamic model to externally applied forces and moments*, J. Biomedical Eng., 105, p. 281.

Moeinzadeh, M.H., Engin, A.E. 1988. *Dynamic modeling of the human knee joint,* in: Computational Methods in Bioengineering, Vol. 9, American Society of Mechanical Engineering, Chicago, 145.

Becker, R. andAwiszus, F. 2001. Physiological alterations of maximal voluntary quadriceps activation by changes of knee joint angle. *Muscle Nerve*, *24*, 667–672.

Williams, M. andStutzman, L. 1959. Strength variation through the range of joint motion. *Physical Therapy Review*, *39*, 145–152.

Marginson, V. andEston, R. 2001. The relationship between torque and joint angle during knee extension in boys and men. *Journalof Sports Sciences*, *19*, 875–880.

Desbrosses, K., Babault, N., Scaglioni, G., Meyer, J.P. andPousson, M. 2006.Neural activation after maximal isometric contractions at different muscle lengths. *Medicine & Science in Sports &Exercise*, *38*, 937–944

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