Human Knee Inverse Dynamics Model of Vertical Jump Exercise

Dumitru I. Caruntu
Department of Mechanical Engineering
University of Texas Rio Grande Valley
1201 W University Drive
Edinburg, TX 78539
e-mail: dumitru.caruntu@utrgv.edu
caruntu2d2@asme.org
ASME Member

Ricardo Moreno
Department of Mechanical Engineering
University of Texas Rio Grande Valley
1201 W University Drive
Edinburg, TX 78539
e-mail: rmoreno851@gmail.com

ABSTRACT

This work deals with the dynamics of the human knee during vertical jump exercise. The focus is on the joint forces necessary to produce the jump, and to dissipate energy during landing. A two-dimensional sagittal plane, inverse dynamics knee model is developed. This model uses data from a motion capture system and force plates in order to predict knee joint forces during the vertical jump exercise. The knee model consists of three bony structures femur, tibia, and patella, ligament structures to include both cruciate and collateral ligaments, and knee joint muscles. The inverse dynamics model is solved using optimization in order to predict knee joint forces during this exercise. Matlab software package is used for the optimization computations. Results are compared with data available in the literature. This work provides insight regarding contact forces and ligaments forces, muscle forces, and knee and hip contact forces in the vertical jump exercise.

1 Corresponding author
INTRODUCTION

Biomechanics is an active field of research that gives insight in areas such as sports, ergonomics and bioengineering. Biomechanics research may improve procedures in rehabilitation, product design, and work environments, Nishida et al. [1]. There have been many advances in computer modeling, data acquisition, motion simulations, and image rendering with biomechanical data. Biomechanical data has been used to replicate movements in robots and make their motion more lifelike with the “ability to perform fast movements, other properties, and particularities,” Babic and Lenarcic [2]. In biomechanics, models are built to investigate joint motion, and contact, ligament, and muscle forces during exercises such as walking and running.

Anatomical, biochemical, and physiological characteristics, all contribute to musculoskeletal system, Bobbert et al. [3], which is continuously acting since the human body is an “inherently unstable system,” Winter [4]. Understanding the behavior of the musculoskeletal system gives perspective to daily life tasks. Few investigations have been dedicated to leg dynamics with faster execution speeds to include loads on the hip and knee joints, Cleather et al. [5]. Understanding these motions allows for more insight in the human potential, since sporting activities produce forces 3-4.5 times the person’s bodyweight [5]. These performance-based movements could be simplified to a simple jump and landing motion.

Vertical jump is a ballistic movement, which can be considered “one of the most ‘explosive’ tests due to both, its very short duration, and the high intensity involved,” Samozino et al. [6]. The vertical jump exercise can be broken down into four distinct phases: 1) standing position, 2) jumping, 3) flight time, and 4) landing. Landing, often described as deceleration and stabilization of the body after contact, Spägele et
al. [7], is the essential part for injury prevention, which can vary given the situation. Joint kinematics and kinetics, energy absorption strategies, muscle activation patterns, and landing style are a few factors that influence landing mechanics, Pflum et al. [8]. Landing styles can vary from toe-heel, flatfoot, toe-only, and heel-only, Dufek and Bates [9]. The overall goal in a landing strategy is to dissipate the force effectively produced by the contact, Prilutsky and Zatsiorky [10].

Deterring from injury is the goal in executing a movement. The knee is the most commonly injured joint and with the most severity [9]. The most “catastrophic knee injuries that debilitate athletic careers” are ligament ruptures, Bates et al. [11], of which 70-80% occur in non-contact situations, without any external interference, person-to-person contact, or object-to-person contact, Norcross et al. [12]. From these non-contact situations, the rupture of the anterior cruciate (ACL) ligament could be attributed to landing [8]. Injuries of this nature are usually in one-legged landings and are “considered more dangerous because of the decreased base of support and the increased demand required by absorption of the impact of landing,” Pappas et al. [13]. However, injuries can happen during two-legged landings as well.

Understanding these mechanisms is beneficial for performance-based tasks or rehabilitation. Mathematical models are used to predict the characteristics of human movement. The equations of motion are used as equality constraints in inverse dynamics models. However, the difficulty lies in finding a “physiologically feasible set of controls” for the system, Schellenberg et al. [14]. In doing so, an objective function could be defined on the motor task [7].

Recent research reported in the literature and dedicated to vertical jump exercise includes strength and conditioning [15-17], and physiotherapy [18]. Hirayama
Dumitru I. Caruntu, corresponding author, CND-18-1432


Investigations on vertical jump exercise have been reported by Cleather et al. [5], Spägele et al. [7], and Blajer et al. [19]. Specifically, Cleather et al. [5] used a biomechanical model of the right lower limb to calculate the internal joint forces experienced by the lower limb during vertical jumping with a particular emphasis on the forces experienced by the knee. They used an inverse dynamics approach in which the experimental data used was from twelve athletic males (age 27.1 ± 4.30 years; mass 83.7 ± 9.90 kg) who performed 5 maximal countermovement jumps with their hands on their hips and the highest jump (height 0.38 ± 0.05 m). They reported the tibio-femoral joint and hip joint loadings experienced by a typical subject during vertical jumping and landing to be 6.50 times body weight (BW), and 3.70 BW, respectively. Spagele et al. [7] applied a multi-phase dynamic optimization approach to a real human vertical one-legged jump consisting of an upward propulsion, an airborne and a landing phase. They aimed to understand of how the central nervous system coordinates muscle excitations in order to accelerate and decelerate body segments of the lower limb for a measured one-legged, vertical jump. They reported
the normalized muscle excitations of their nine muscles model of the lower limb. Biceps femoris long head and biceps femoris short head showed very little normalized muscle excitations. Blajer et al. [19] presented a two-dimensional biomechanical model of a human body for determination of the muscle forces and joint reaction forces in the lower extremities during sagittal plane movements such as vertical jump. The experimental results used in the model were from a vertical jump performed by an athlete (basketball player) of mass 114.8 kg and height 194 cm. While the hip, knee and ankle joints were modeled as enforced directly by the muscle forces applied to the foot, shank, thigh and pelvis at the muscle attachment points, the actuation of the other joints was simplified to the torques representing the respective muscle action. They reported a peak of the total force of the vasti muscles (medialis, lateralis, and intermedius) of 3.40 times body weight.

Present work investigates the muscular, ligament, and contact forces, and tibio-femoral contact point trajectory associated with vertical jump exercise, i.e. countermovement jump (CMJ) with the arms kept akimbo. Present vertical jump consists of larger jumping duration and similar landing duration when compared to the explosive vertical jump in Ref. [5]. To investigate internal forces experienced during the vertical jump exercise, a two-dimensional knee model was developed using a Newton-Euler formulation. Using an inverse dynamics approach, the mathematical model predicted the internal forces during the exercise. Although, the vertical squat jump exercise has been previously investigated, it is not completely understood. To the best of our knowledge, the novelty of the present work consists of reporting the significant forces experienced by the lower limb during the vertical jump exercise, namely 1) tibio-femoral, patello-femoral, and hip contact forces, 2) gluteus,
quadriiceps, and gastrocnemius muscle forces, and 3) posterior cruciate ligament (PCL) and medial collateral ligament (MCL). Also this work reports 4) the tibiofemoral contact point trajectory during the moderate vertical squat jump exercise. In this research, the human subject was a recreational athlete, an advent runner, familiar to consistent execution. The subject was 1.75 m tall and had a mass of 84 kg. The largest values reached during jumping and landing were for knee flexion angle 95.4° and 55.4°, vertical ground reaction force of 1.10 times body weight (BW) and 2.70 BW, quadriceps muscle 1.10 BW and 2.50 BW, gluteus muscle 2.10 BW and 1.12 BW, gastrocnemius muscle 1.40 BW and 2.80 BW, tibio-femoral joint contact 2.10 BW and 5.60 BW, patello-femoral contact 1.10 BW and 2.00 BW, and hip contact force 1.60 BW and 2.40 BW, posterior cruciate ligament (PCL) 0.55 BW and 0.65 BW, and medial collateral ligament (MCL) 0.22 BW and 1.10 BW, respectively.

Significant forces throughout the entire exercise were ground reaction forces, quadriiceps and gluteus muscle forces, contact forces (tibio-femoral, patello-femoral and hip), and PCL force. Except gluteus, all these forces reached a maximum during the ascent phase of the jumping. Forces that showed a maximum at the lowest jumping position of 95.4° flexion angle were quadriceps and gluteus muscle forces, PCL force, and tibio-femoral contact force. Forces that were significant only right before take-off and the first phase of landing were gastrocnemius muscle, and MCL. Forces included in the model that showed no significant values were hamstrings and iliacus muscles, anterior cruciate ligament (ACL), and lateral collateral ligament (LCL).

The tibio-femoral contact point moved 6 mm posteriorly on the tibial plateau during the descent phase of jumping reaching a most posterior position. During the ascent phase of jumping that followed, the contact point has been moved about 7 mm
anterioirly from the most posterior position on the tibial plateau. During landing, from the most anterior position reached, the contact point traveled posteriorly on the tibial plateau about 10 mm, then anteriorly about 9 mm.

Comparisons were made to other vertical jump and regular squat studies to add validity to the present knee model.

2D HUMAN LEG ANATOMICAL MODEL

1. Kinematic Data. Musculoskeletal models are developed in order to capture and provide objective criteria for various movements [7]. The present model focuses on the vertical squat jump exercise. There are multiple factors that contribute to the execution of the movement such as muscular coordination, muscular strength, arm swing, arm placement, and others depending on the study, Kim and Kim [20].

   The knee model is composed of three bones, namely, femur and tibia, and patella. The knee is not modeled as a simple revolute joint. The entire motion is observed in a global coordinate system, which is a fixed XY coordinate system. Anatomical data to include articular geometry and insertion points of ligaments and muscles were given local coordinate systems, which are attached to the centers of mass of their respective bodies. OpenSim was used to extract images, model leg6dof9musc.osim, in order better grasp the geometry, anatomy, and the function of femur and tibia, Figs. 1 and 2. Muscle, ligament Fig.3, and contact forces’ components are observed in their local coordinate systems, and later are transformed into components in the global coordinate system. Tibial orientation is given by $\theta_1$, the angle between the tibial longitudinal axis and positive horizontal X-axis, and femoral orientation by $\theta_2$, the
angle between the femoral longitudinal axis and positive horizontal X-axis, Figs. 1 and 2. The overall motion is best described with respect to the tibiofemoral knee flexion angle $\theta_{TF} = \theta_2 - \theta_1$. Patellofemoral flexion angle $\theta_{PF}$ displays a linear behavior with respect to $\theta_{TF}$ [21-23]. Caruntu and Hefzy [21] developed a three dimensional anatomically based dynamic modeling of the human knee to include both knee joints, tibio-femoral and patella-femoral. Their model was a forward dynamics model. They simulated the knee extension exercise and showed that for three different values of quadriceps force, the relationship between the patellofemoral flexion angle $\theta_{PF}$ and tibio-femoral flexion angle $\theta_{TF}$ is as follows

$$\theta_{PF} = \frac{7}{9} \theta_{TF} \tag{1}$$

In present work, we assumed the relationship between $\theta_{PF}$ and $\theta_{TF}$ given by Eq. (1) to hold. The direction of the patellofemoral contact force on the femoral condyle is given by $\theta_{PF}$, Eq. (1). The angle between the patellar tendon and the tibial shaft $\theta_{PT}$, known as PT sagittal plane angle, is a function of the flexion angle as given by DeFrates et al. [24] and Varadarajan et al. [25]

$$\theta_{PT} = \frac{\pi}{9} - \frac{3}{11} \theta_{TF} \tag{2}$$

Figure 4 displays the patellofemoral flexion angle $\theta_{PF}$ and the patellar tendon angle $\theta_{PT}$ as functions of the tibiofemoral flexion angle $\theta_{TF}$, Eqs. (1) and (2).

2. **Knee Articular Surfaces.** Knee femoral articular surface is modeled by two circles, one approximating the patello-femoral contact, and the other one, the tibiofemoral contact. The patello-femoral circle serves as reference for muscle...
insertions, patellar tendon orientation, and patellofemoral contact, Fig. 3. References [26-28] are used to estimate the patellofemoral circle, tibiofemoral circle, and the tibiofemoral contact point at 90° flexion angle, respectively. The radius of the patellofemoral circle is calculated using Yue et al. [26], which reports knee anthropometric data of Chinese, and white men and women. They reported the anteroposterior length of the femoral condyle between the two races. Assuming the human bone structure is related to a person’s height, rather than race, a linear relationship is considered between anteroposterior length of the femoral condyle and the person’s height. The human subject of this present work was 1.75 m tall and had a mass of 84 kg. It was calculated that the subject’s anteroposterior length of the femoral condyle is 7 cm, so the radius of the patello-femoral circle was 3.5 cm [26]. The tibio-femoral circle is used for the tibiofemoral center of rotation, contact point, and ligament insertions. The radius for the tibiofemoral circle was calculated using Granados [27], in which a two-dimensional anatomical knee model was reported. The anatomical surfaces were based on the x-ray of a human subject. For the present work, the x-ray was scaled up to match the anteroposterior length of the subject’s femoral condyle, then the curvature of the posterior side of the femoral condyle was traced using Matlab, and the radius of the tibiofemoral circle was found to be about 2 cm [27].

Finding the proper tibiofemoral center of rotation (COR) was an essential part, since this point is directly related to the ligament insertion points placement and the location of the tibiofemoral contact point. A virtual marker was added in order to adjust the COR to its proper location. The location of COR was determined using the tibiofemoral circle radius, acquired from [27] and Hill et al. [28], where the posterior centres of the femoral condyles were measured during a squat exercise. COR was
placed at a selected distance, anterior to the posterior edge of the tibial plateau. Specifically, it was placed on a direction perpendicular to the tibial plateau and above the tibiofemoral contact point (CTF), Fig 5. At 90° knee flexion angle, the location of CTF was 2 cm from the posterior edge of the tibial plateau [28]. COR was placed 2 cm distance from the tibial plateau on the perpendicular direction to the tibial plateau through CTF, at 90° knee flexion.

The tibiofemoral COR location was then calculated with respect to the intercondylar marker. The intercondylar marker (IM) is a virtual marker located at the midpoint in the $x_2 y_2$ plane between the lateral and medial knee markers, which are denoted as LM and MM, respectively, Fig. 5. At 90° knee flexion, the polar coordinates, with respect to IM of the tibiofemoral COR were calculated. The tibiofemoral COR coordinates were then calculated for all time frames of the exercise.

3. Leg Loads

Muscles. Muscles considered in this work, along with their insertions are as follows: 1) gastrocnemius $F_{gas}$ located between the femoral condyle and the calcaneus through the Achilles tendon, 2) biceps femoris short head $F_{bfs}$ between the fibula head and linea aspera and lateral supracondylar line of femur, 3) biceps femoris long head $F_{bfl}$ between fibula head and ischial tuberosity, 4) rectus femoris $F_{rf}$ between anterior inferior iliac spine and patella base, 5) gluteus $F_g$ between iliac crest and posterior gluteal line and greater trochanter, 6) vasti $F_v$ between femur and patella base, 7) iliacus $F_i$ between the region of the anterior inferior iliac spine and lesser trochanter. Hamstrings muscle force $F_h$, and quadriceps muscle force $F_q$ are given by

$$F_h = F_{bfs} + F_{bfl}, \quad F_q = F_{rf} + F_v$$

(3)

Accepted Manuscript Not Copyedited

Journal of Computational and Nonlinear Dynamics. Received September 27, 2018; Accepted manuscript posted July 18, 2019. doi:10.1115/1.4044246

Copyright (c) 2019 by ASME
Knee Ligaments. There are four major knee ligamentous structures that hold the joint together. Ligaments are bone to bone connective tissues. They limit the relative displacement between femur and tibia. The four ligaments are anterior cruciate ligament (ACL), posterior cruciate ligament (PCL), lateral collateral ligament (LCL), and the medial collateral ligament (MCL), their respective forces are denoted by $F_{\text{ACL}}$, $F_{\text{PCL}}$, $F_{\text{LCL}}$, and $F_{\text{MCL}}$, respectively, Eq. 3. Figures 1 and 2 depict the free body diagrams of tibia and femur for a current moment in time during the exercise.

The locations of the ligament insertion points were considered in the standing position, Shelburne and Pandy [29]. The insertion points were then related to their respective body, i.e. femoral ligament points with respect to the the tibiofemoral center of rotation (COR), and the tibial ligament insertions with respect to the tibial plateau. The ligaments were modeled as non-linear strings, providing force only in tension. The definition for the linear and non-linear regions, slack lengths, and stiffness coefficients of the ligaments were acquired from Ref. [21]. The nonlinear behavior of ligaments given by

$$
(F_i') = \begin{cases} 
0, & \varepsilon^n \leq 0 \\
(k_q^n)(L^n - L_0^n), & 0 < \varepsilon^n < 2\varepsilon_0 \\
(k_l^n)[L^n - (1 + \varepsilon_0)L_0^n], & \varepsilon^n \geq 2\varepsilon_0 
\end{cases}
$$

(4)

where the $\varepsilon^n$, $(k_q^n)$, $(k_l^n)$, $L^n$, $L_0^n$, and $\varepsilon_0$ are the strain, the stiffness coefficients for the quadratic and linear regions, the current length and slack length the $n^{th}$ ligament, and the threshold assumed 0.03, respectively,” Caruntu and Hefzy [21].

Only PCL force was not calculated using Eq. (4). Since the tibiofemoral center of rotation has an unusual behavior, Most et al. [30], the PCL was the only ligament
largely affected. The marker-based approach captures the motion of the transepicondylar axis but not the geometric center axis, which depicts the true motion of the knee, [30] and Kozanek et al. [31]. Due to this limitation, PCL was left to be found through optimization along with muscle forces and contact forces, as shown afterwards. However, PCL insertion points were anatomically based.

*Contact Loads.* Contact force is produced by contact of bones. There are three contact forces considered, namely tibio-femoral contact force, patello-femoral contact force and hip joint contact force. The tibio-femoral contact force consists of two components: contact force parallel to the tibial plateau direction $F_{cx}$, and perpendicular to the tibial plateau $F_{cy}$. The hip contact force components are perpendicular to the femoral longitudinal direction $F_{hx}$, and parallel to the femoral longitudinal direction $F_{hy}$. All muscles and ligaments provide forces in tension, and bone contact only in compression.

There are some assumptions to simplify this model. First, the patellar tendon orientation is given by DeFrate et al. [24], and its force is equal to the quadriceps force given by Eq. (3). Second, although, patella is not included as a third body in the investigation, the patello-femoral contact force $F_{cp}$ is included in the model. Third, $F_{gas}$ and $F_{cy}$ are parallel to the longitudinal axis of tibia, Bolsterlee et al. [32] and Yoshioka et al. [33], respectively, and $F_{cx}$ is zero due to negligible friction on the tibial plateau, Mow et al. [34]. Fourth, the force $F_{h}$ is parallel to the femoral longitudinal axis, Flandry and Hommel [35]. Fifth, the geometrical center of the femoral condyle has a trajectory parallel to the tibial plateau [31].
4. Equations of Motion. The equations of motion are written in the sagittal plane using Newton’s 2nd Law on the X and Y directions, and Euler equation on the Z direction for each body. The equations of 2-D motions of tibia and femur are

\[
\sum F_{Xi} = m_i \times a_{XCi}, \quad \sum F_{Yi} = m_i \times a_{YCi}, \quad \sum M_{Ci} = I_{Ci} \times \alpha_i,
\]

where subscripts are \( i = 1 \) for tibia, \( i = 2 \) for femur, \( C \) for center of mass; \( F_{Xi}, F_{Yi}, M_{Ci}, I_{Ci} \) are forces, moments, and moments of inertia, respectively; \( a_{XCi}, a_{YCi}, \alpha_i \) are linear accelerations of \( C \), and angular accelerations of the corresponding bodies, respectively. Equations (5) can be rewritten as:

\[
R_{Xi} = \left( \sum F_{Xi} \right)_{int} = m_i \times a_{XCi} - \left( \sum F_{Xi} \right)_{ext}, \quad R_{Yi} = \left( \sum F_{Yi} \right)_{int} = m_i \times a_{YCi} - \left( \sum F_{Yi} \right)_{ext}, \quad T_i = \left( \sum M_{Ci} \right)_{int} = I_{Ci} \times \alpha_i - \left( \sum M_{Ci} \right)_{ext}
\]

where \( R_{Xi}, R_{Yi}, T_i \) are the resultant intersegmental (knee and hip) forces and moments, \( \left( \sum F_{Xi} \right)_{ext}, \left( \sum F_{Yi} \right)_{ext}, \left( \sum M_{Ci} \right)_{ext} \) are the external forces on tibia (and foot) and femur, and \( \left( \sum F_{Xi} \right)_{int}, \left( \sum F_{Yi} \right)_{int}, \left( \sum M_{Ci} \right)_{int} \) are the internal forces given by ligament, contact, and muscle forces on the respective body, Bartel et al. [36].

5. Inverse Dynamics. An inverse dynamic model is developed and used in the present work. Finding the muscle forces and internal loads during the exercise is the overall goal [19]. The motion of the subject during the exercise was captured using a Vicon Motion Analysis System. This data was acquired in the Biomechanics Laboratory at the University of Texas Rio Grande Valley. The experimental data was used to calculate the input data for the inverse dynamics model. The input data consisted of
1) coordinates of the femoral and tibial centers of mass, 2) the orientations of tibia and femur, 3) their corresponding accelerations, and 4) ground reaction forces. The first step was to calculate the inter-segmental forces and moments $R_{Xi}$, $R_{Yi}$, $T_{i}$, Eq. (6), for each joint and time frame using the experimental data. Forces were calculated by considering of each segment, iteratively, moving from tibia plus foot, proximally along the kinetic chain, Cleather and Bull [37]. The inverse dynamics model used optimization to solve the system of equations for a large number of unknowns to include muscle forces and contact forces, Seireg and Arvikar [38], Moissenet et al. [39], and Lin et al. [40]. Specifically, the unknowns are nine muscle forces to include $F_{bfl}$, $F_{bfu}$, $F_{d}$, $F_{r}$, $F_{g}$, $F_{v}$, and $F_{gas}$, and $F_{h}$, and $F_{q}$, see Eq. (3), four contact forces $F_{hx}$, $F_{hy}$, $F_{cy}$, and $F_{cp}$, one ligament force $F_{pcl}$, and the location of the tibio-femoral contact point on the tibial plateau. These forces are internal forces having as resultants

$$\left( \sum F_{Xi} \right)_{int}, \left( \sum F_{Yi} \right)_{int}, \left( \sum M_{Ci} \right)_{int}$$

which are the inter-segmental forces and moments in Eq. (6), intersegmental forces that are already calculated from the experimental data. This is an underdetermined system since the number of equations is lower than the number of unknowns. Such system has an infinite number of possible solutions. To find a solution of the inverse dynamics model, the optimization method used an objective function that was minimized under given constraints. Several objective functions can be formulated for minimization of the forces in the muscles, the work done by the muscles, reactions at the joints, moments carried by the ligaments at the joints [38], muscle forces and moments at all joints, Seireg and Arvikar [41], muscle stresses, Czaplicki et al. [42], Crowninshield and Brand [43], and Pierce et
The objective function, [38-40] and Nigg and Herzog [46], to be minimized in the present work includes muscle forces, contact forces and the PCL force,

$$\min f = F_{bfl}^2 + F_{bfs}^2 + F_{rf}^2 + F_v^2 + F_i^2 + F_{gas}^2 + F_{pcl}^2 + F_{cy}^2 + F_{hx}^2 + F_{hy}^2$$

(7)

with the unknowns satisfying the following inequality constraints

$$F_{bfl} \geq 0, F_{rf} \geq 0, F_{cx} = 0, F_{cy} \geq 0, F_{g} \geq 0, F_{i} \geq 0, F_{hx} \geq 0, F_{hy} \geq 0$$

$$F_{vx} \geq 0, F_{v} \geq 0, F_{bfs} \geq 0, F_{gas} \geq 0, F_{cp} \geq 0, F_{pcl} \geq 0, D \geq 0$$

(8)

and equality constraints given by Eqs. (5).

3. EXPERIMENTAL PROTOCOL

1. Biomechanics Motion Instrumentation. The experimental data was gathered using two AMTI force plates and ten Vicon MX T-Series infrared cameras. The AMTI force plates, each measuring 60cm x 60cm, provided the ground reaction forces in $X$, $Y$, and $Z$ directions, moments about the $X$, $Y$, and $Z$ directions, and centers of pressure in the $XZ$-plane. The sampling rate for the AMTI force plates was 1000Hz.

The ten infrared cameras captured the light reflected by the markers as the subject performed the task. The Vicon system recorded all marker locations at a rate of 100Hz, and provided the coordinates of each marker coordinates to be used in the present model. Both sets of equipment are shown in Fig 6.

2. Protocol. The human subject was required to perform a warm-up exercise and dynamic stretching before conducting the test. The warm-up exercise consisted of
walking lunges, high knees, and practice vertical jumps. Once completing the five-minute warm-up routine, the subject was instructed to complete the vertical jump test protocol. The procedure was comprised of five maximal vertical jumps. This is a test based on contact and air time [6].

The vertical jump in this work is a countermovement jump (CMJ) with no arm swing. The CMJ started from the upright standing position and transitioned through instructed phases to replicate the exercise. Two positions are illustrated in Fig. 7. The arms are kept akimbo (arms placed on the hips and elbows faced outward) in order to mitigate any momentum in the CMJ similar to [5]. Additionally, the initial descent phase was performed at a slower pace to further limit any loading. The depth of the squat of jumping was measured between the ground and the hip and measured when the subject’s knee flexion angle was approximately 95° [6]. The subject was instructed to hit the same descent marker before every vertical jump. The subject was asked for a fast ascent phase, jump for a maximum height, and soft landing. A soft landing was accomplished by performing another squat once the feet touch down, in a toe-heel landing progression. If not all the requirements were met, the trial was discarded and repeated after the appropriate rest interval. After every test, completed or failed, the subject waited two minutes before attempting another trial. This was done to avoid any fatigue that might be encountered.

3. Markers. The marker set employed in this study comprises of markers on pelvis, thigh, calf, and foot [5,37]. Reflective markers were placed on bone landmarks of the subject, Fig. 8. Their placements were most distal point on the toe, heel, lateral and medial ankle, lateral and medial knee condyles, tibial tuberosity, hip, and the front
and rear of the pelvis. Shells of four markers were placed on the foot, shank, and thigh. These shells were used when the capture experienced a “gap” in the data. The shells aided the software to calculate the missing position of the marker. Gaps are instances where the marker was not captured, which may be due to marker being covered. The captured data was used for the input data to the knee model.

The segment lengths, centers of mass, radii of gyration, and moments of inertia of tibia and femur were calculated using subject’s mass and height and anthropometric data, Winter [47], De Leva [48], and Zatsiorski et al. [49].

4. Data Filtering. The experimental data was captured and processed using the Vicon Nexus software. Collected data was then exported onto an excel spreadsheet where the marker data and force plate data was compiled. This data was used as input data for the 2D human leg anatomical inverse dynamics model. The collected data (raw data) was filtered using a low-pass, fourth-order, zero-lag Butterworth filter, [47] and Yu et al. [50] with a cutoff frequency of 30 Hz, Weiss et al. [51] and Bisseling and Hof [52], resulting from Residual Analysis [47] of the experimental data.

Linear velocities $\mathbf{v}_{i+1/2}$ and accelerations $\mathbf{a}_{i+1/2}$ of the centers of mass, and angular velocities $\mathbf{\omega}_{i+1/2}$ and accelerations $\mathbf{\alpha}_{i+1/2}$ were calculated halfway between sample times [47]. The frame rate is 100Hz, therefore $\Delta t$ is 0.01s. These accelerations were then used as input data for the equation of motion for the two-dimensional model. The equations of motion were equality constraints in the inverse dynamics model used in this work.
4. NUMERICAL SIMULATIONS

Numerical simulations were conducted using the inverse dynamics model. 1) The vertical jump exercise is compared to other studies with similar procedure, and 2) its descent phase of jumping is also compared with the descent phase of regular squat exercise. The reason for this last comparison is that the descent phase of jumping is similar to some extent to the regular squat. For the vertical jump exercise only jumping and landing are of interest. The standing position and airborne times are not included.

Figures 9 and 11 show the vertical coordinate of the hip joint (greater trochanter marker), and the ground reaction forces during the exercise. In Fig. 9, the vertical coordinate of the hip joint decreases from about 0.89 m to 0.60 m between \( t = 1.40 \) s and \( t = 2.22 \) s, and then the vertical coordinate of the hip increases from 0.60 m to 1.20 m between 2.22 s and 2.90 s. Therefore, the subject jumps 0.3 m above the standing position. During landing, at \( t = 3.26 \) s, the vertical coordinate of the hip reaches a lowest value of 0.80 m, and at \( t = 3.50 \) s the hip reaches 0.88 m.

Figure 10 shows the vertical ground reaction forces (GRF), vertical \( R_y \), and anterior-posterior \( R_x \), on one leg. Vertical force \( R_y \) is significant in this exercise and has bimodal peaks, indicating the toe heel landing style [9]. The largest GRFs are right before the take-off and right after landing. At \( t = 1.00 \) s, \( R_y \) has a value of 0.50 BW, which corresponds to standing, and keeps this value until \( t = 1.27 \) s, and then \( R_y \) decreases to a minimum of 0.34 BW at \( t = 1.40 \) s. Next, \( R_y \) continuously increases to reach a maximum of 1.10 BW at \( t = 2.60 \) s, and then quickly decreases to zero at \( t = 2.66 \) s. The interval of time when the subject is airborne is between \( t = 2.66 \) s and \( t = 3.10 \) s. At \( t = 3.15 \) s, \( R_y \) reaches a maximum of 2.70 BW, then decreases to 1.10 BW and again increases, now to a local maximum of 1.60 BW at \( t = 3.20 \) s. The force \( R_y \)
then decreases to reach a minimum of 0.35 BW at \( t = 3.50 \) s. Then it continues to slowly increase to 0.54, BW and then decrease to 0.50 BW, at \( t = 3.80 \) s and \( t = 4.40 \) s, respectively.

Figure 11 shows the knee flexion angle and the important stages of the exercise marked by vertical lines. The flexion angle reaches a maximum of 95.4° during jumping, and 55.4° during landing. The progression of the exercise is marked by small subject figures on top of the graphs and vertical lines for important transition times. The first vertical line at \( t = 2.22 \) s marks the lowest point (the largest knee flexion angle) of the descent phase of jumping. This is the time of the transition from the descent to the ascent phase of jumping. The second vertical line at \( t = 2.66 \) s shows the instant when the subject takes off and goes airborne. The third vertical line at \( t = 3.10 \) s marks the instant when the subject lands. The progression of the exercise is also marked in Figs. 13-19.

The cycle of the vertical jump exercise is defined as follows. The beginning of the exercise is considered when the vertical ground reaction force reaches its minimum in the descent phase of jumping, specifically when the time \( t = 1.40 \) s, Fig 10. One can see from Fig. 9 that this is the time when the descent phase starts. The end time of this exercise is considered when the vertical ground reaction force reaches its minimum in the landing phase of the exercise, specifically when the time \( t = 3.50 \) s, in Fig. 10. One can see that at this time the subject almost reaches the standing position. From Figs. 9 and 11, one can notice that the time for the lowest squatting position (largest flexion angle) of jumping occurs at \( t = 2.22 \) s, and the time for the lowest squatting position of landing occurs at \( t = 3.27 \) s.
Figure 12 illustrates the angles $\theta_2$ and $\theta_1$ between femoral and tibial longitudinal axes and the positive global horizontal x-axis. The knee flexion angle in Fig. 11 was calculated as $\theta_{TF} = \theta_2 - \theta_1$.

Figure 13 shows the quadriceps, hamstrings, and gastrocnemius muscle force production, where the quadriceps and hamstrings are the combination of components as described in Eq. (3). The quadriceps muscle is the main contributor to the vertical jumping being active throughout the entire exercise. During jumping, it increases from zero reaching 1.10 BW at the lowest descent position of jumping (largest knee flexion angle), decreases to 0.80 BW, and then increases to a maximum of 1.00 BW before the take-off. During landing, in the first part landing, the quadriceps muscle experiences a force of about 2.50 BW. The gastrocnemius muscle force is not significant during the exercise, except right before the take-off when it reaches around 1.40 BW, and in the first part of landing when it reaches 2.80 BW.

Figure 14 depicts the contact forces experienced in the hip and the knee. The tibiofemoral normal contact force $F_{cy}$ is the highest of the five contact force components predicted by the present model. During jumping 1) $F_{cy}$ reaches two maxima, 2.00 BW at the lowest descent position of jumping and 2.10 BW right before the take-off, 2) the hip contact force $F_{hy}$ reaches a local maximum of 0.9 BW during the descent phase, a local minimum of 0.71 BW at the lowest descent position of jumping, and a maximum of 1.60 BW right before the take-off. 3) The patellofemoral contact force $F_{cp}$ has the same pattern as the contact force in the hip, reaching a local maximum of 0.54 BW during the descent phase of jumping, and a maximum of 1.00 BW right before the take-off. During landing, in the first part of it, 1) $F_{cy}$ experiences a
maximum of 5.60 BW, 2) $F_{hy}$ which is parallel to the femoral longitudinal axis reaches a maximum of 2.50 BW and shortly another local maximum of 2.30 BW. $F_{hy}$ experiences a force of 0.23 BW at the end of the exercise, $t = 3.50$ s. 3) In the first part of landing, $F_{cp}$ reaches a maximum of 2.00 BW. Then it decreases to zero by the end of the exercise, $t = 3.50$ s. The force $F_{cp}$ is in agreement with Cleather et al. [5], who reported the same pattern and maxima of 3.20 BW and 3.30 BW during jumping and landing, respectively. Their larger maximum values are due to 1) the subject, male athletes [5] versus a recreational athlete in the present work, 2) jump height, 0.40 m [5] versus 0.30 m in the present work, and 3) different time intervals, jumping time 0.75 s [5] versus 1.26 s in the present work, and landing time of about 0.30 s [5] versus 0.40 s in the present work.

Figure 15 illustrates the motion of the tibiofemoral contact point during the exercise, where $D$ is the tibial distance between the tibiofemoral contact point and the posterior edge of the tibial plateau. During jumping, the contact point moves 6 mm posteriorly on the tibial plateau during the first half of the descent phase of jumping, then 2 mm anteriorly as the subject reaches the lowest position of jumping, 2 mm posteriorly during the first half of the ascent phase, and another 7 mm anteriorly during the second half of the ascent phase. During landing, the contact point travels posteriorly 10 mm from the most anterior position (same location as right before the take-off), then 8 mm anteriorly followed by 4 mm posteriorly, and then another 5 mm anteriorly settling to the same position as when the entire exercise started. The take-off of jumping and the beginning of landing find the tibiofemoral contact point with respect to the tibial plateau in the most anterior positions. The same pattern of the contact point is observed for both jumping and landing. However, landing experiences
a faster change in the location of the contact point. This is due to much shorter landing
time than the jumping time. The distance traveled by the contact point on the tibial
plateau is less than or equal to 10 mm.

Figure 16 illustrates the ligament forces in the knee. The ligament with the
most activity is the PCL. During jumping, the maximum PCL forces are 0.45 BW and
0.54 BW at the lowest position (maximum flexion angle of jumping) and right before
take-off, respectively, and the medial collateral ligament (MCL) is active for a very
short period of time right before the take-off reaching a maximum of 0.22 BW. During
landing, in the first phase of it, the maximum PCL force is 0.65 BW, and the maximum
MCL force is 1.10 BW. Forces in PCL are relatively low when compared to contact
forces and some of the muscle forces. The maximum force in the PCL is much less than
the failure limit for healthy subject PCL which is around 4.5kN [5], i.e. about 5.50 BW.
PCL is in tension when posterior shear occurs [5]. The pattern of PCL force during the
exercise is in good agreement with posterior shear force reported in the literature [5].
Again differences between magnitudes in this work and Ref. [5] are due to 1) the type
of athlete, 2) jump height, and 3) different jumping and landing time intervals, please
see the discussion of Fig. 14. The other two ligaments, anterior cruciate ligament (ACL)
and lateral collateral ligament (LCL) do not show any significant activity during the
vertical squat jump.

Significant forces of the vertical squat jump are shown in Fig. 17, tibiofemoral
normal contact force $F_{cy}$ in the knee, parallel to the femoral longitudinal axis, hip
contact force $F_{hy}$, gluteus muscle force $F_{g}$, quadriceps muscle force $F_{q}$,
gastrocnemius muscle force $F_{gas}$, and PCL force. Gluteus muscle force is a very
important force throughout the exercise. Gluteus muscle has the largest force in the system, 2.12 BW for the lowest position of jumping, and a significant value of 1.13 BW during landing.

From the present data, the greatest magnitudes of the significant forces in this exercise are experienced during jumping, right before the take-off, and right after landing. However, gluteus muscle reaches its maximum at the largest knee flexion angle during jumping. The knee was the joint that experienced most of the loads, contact and supporting muscles.

5. DISCUSSION AND CONCLUSIONS

In this work a sagittal plane, inverse dynamics, model of human knee joint was developed in order to investigate contact, muscle, and ligament forces, and tibio-femoral contact point motion during vertical jump exercise. Experimental data, collected in the Biomechanics Laboratory at the University of Texas Rio Grande Valley, was used for input data for the inverse dynamics model. The vertical jump exercise consists of three phases, namely jumping, airborne subject, and landing. The novelty of this work is related to vertical jump exercise with a larger completion time of the jumping phase and consists of 1) predicting the tibio-femoral, patello-femoral, and hip joint contact forces, and quadriceps and hamstrings muscle forces. This investigation also reports, for the entire duration of the vertical jump exercise, 2) the motion of the tibio-femoral contact point on tibial plateau, 3) gastrocnemius and 4) gluteus muscle forces, and 5) cruciate and collateral ligaments’ forces.
All contact forces, and quadriceps muscle force, experience peak values during the ascent phase of jumping and first part of landing. The largest peak values during these two phases belong to tibio-femoral contact force, jumping 2.10 BW, and landing 5.60 BW. The contact forces for jumping, predicted in this work, have the same pattern, but lower values than data reported in the literature. This is due to larger jumping time in this work. However, the contact forces for landing are in very good agreement with data reported in the literature, since the landing time was similar.

During the vertical jump, the tibiofemoral contact point, Fig. 15, travels posteriorly on the tibial plateau for the descent phases of jumping and landing, 6 mm and 9 mm, respectively. During both ascent phases, before the take-off and the terminal phase of landing, the contact point travels anteriorly, 7 mm and 9 mm, respectively, at the end of the exercise the contact point is back to its original position during standing.

The level of activation of the hamstrings muscle during the exercise is not significant, Fig. 13. This does not contradict data reported in the literature for muscle activation patterns during squat exercise, Slater and Hart [53]. Their experimental electromyography (EMG) data in terms of normalized muscular activity shows that during squat exercise the hamstrings has only a level of activation of 6-7% of its maximum voluntary isometric contraction during the descent phase and only 7-11% during the ascent phase, while the quadriceps reaches 100% of its maximum voluntary isometric contraction. Spagele et al. [7] reported as well a very small level of hamstrings normalized muscle excitation during the exercise.

Gluteus and gastrocnemius muscle forces show different levels of activation, Fig. 17. The gluteus muscle force $F_g$ is activated during the entire vertical jump exercise,
reaching a peak of 2.10 BW in the lowest position of the descent phase of jumping, and about 1.12 BW during landing. The gastrocnemius muscle force is not activated during the entire exercise, except right before the take-off when it reaches a peak of 1.40 BW, and during the first phase of landing when it reaches a peak of 2.80 BW.

PCL and MCL shows significant level of activation during vertical jump exercise. PCL is activated during the entire exercise showing two peaks during jumping, one peak of 0.45 BW at the lowest jumping position and 0.55 BW right before the take-off, and one peak of 0.65 BW during the first phase of landing. MCL shows rather no significant activity during jumping, but shows a peak of about 0.90 BW during the first phase of landing. ACL and LCL show no significant level of activation during the exercise.

1. **Comparisons of Vertical Jump Predictions with Data Reported in the Literature.**

In this section a comparison between present work and data reported in the literature [5,19] is conducted. There are some differences between type of subjects and type of performance during the exercise. Athletic males jumped as high as 0.40 m in Ref. [5], and basketball players jumped as high as 0.50 m in Ref. [19], while in the present work the exercise was performed by a recreational athlete who jumped only 0.30 m. Figure 18 gives details regarding the completion times as well.

In Figs. 18-21, the zero time for all data is the time when the ground reaction forces have a minimum during jumping, since the exercise cycle was defined as the time between the minimum vertical ground reaction force during the descent phase of jumping, and the minimum vertical ground reaction force during landing. Therefore, one is able to compare the same exercise but with different completion times.
Figure 18 displays a comparison with data available in the literature [5,19] of the vertical component of the ground reaction forces $R_y$ produced during the vertical squat jump exercise. In the vertical axis the ground reaction force units are converted into terms of bodyweight (BW). This allows for a proper comparison. The horizontal axis shows time in seconds. In Refs. [5,19] and present work, total completion times are 1.75 s, 1.80 s, and 2.10 s, jumping times 0.75 s, 0.93 s, and 1.26 s, landing times 0.50 s, 0.50 s, and 0.40 s, and airborne times 0.45 s, 0.40 s, and 0.30 s, respectively. Present work has the largest total and jumping completion times, similar landing time, and lowest airborne time. Figure 18 shows a good agreement of this work with data reported in the literature. The patterns are similar, the largest force magnitudes occurred right before the take-off and right after landing. In Refs. [5], [19] and present work, the maximum values of jumping ground reaction forces are 2.70 BW, 2.10 BW, and 1.10 BW, and the maximum values of landing ground reaction forces are 3.10 BW, 3.30 BW, and 2.70 BW, respectively. While, maximum values of landing ground reaction forces are similar to some extent due to similar landing times, the maximum values for jumping ground reaction forces are quite different due to large variations between the jumping times of the three investigations. For shorter jumping completion time, the contact forces have higher values as in Ref. [5], while for longer jumping completion time, the contact forces have lower values as in this work. The maximum height the subject reached during the exercise influenced the ground reaction forces as well.

Figure 19 depicts a comparison of the predicted tibiofemoral normal contact force $F_{cy}$ with data available in the literature. The tibiofemoral contact force parallel
to the tibial plateau $F_{cx}$ is zero, due to very low friction coefficient. The predictions of present investigation regarding tibiofemoral normal contact force $F_{cy}$ are in agreement with data reported in the literature [5, 19]. All show similar patterns during the exercise. The maximum contact force $F_{cy}$ occurs right before the take-off and right after landing. In Refs. [5], [19] and present work, the maximum values of tibiofemoral contact force during jumping are 6.20 BW, 3.70 BW, and 2.10 BW, and the maximum values of tibiofemoral contact force during landing are 6.50 BW, 4.30 BW, and 5.60 BW, respectively. The maximum values of $F_{cy}$ during jumping are different due to 1) type of athlete, 2) completion times, and 3) largest height attained during the exercise, while the maximum values of $F_{cy}$ during landing are similar due to similar landing times.

Figure 20 illustrates a comparison of the resultant hip contact force $F_h$ to data reported in the literature. The components of the hip contact forces were combined

$$F_{hp} = \sqrt{F_{hx}^2 + F_{hy}^2}.$$  

However, the contact force perpendicular to the longitudinal axis of the femur $F_{hx}$ was so minute, that it made little difference. The hip contact forces predicted by the present investigation are in agreement with Ref. [5]. They show the same pattern. The maxima occur right before the take-off and right after landing. The maximum values during jumping are 4.10 BW [5] and 1.60 BW present work, and during landing are 3.70 BW, and 2.50 BW present work. Again, the differences in maximum values are due to 1) type of athlete, 2) completion times, and 3) largest height attained during the exercise.
Figure 21 illustrates a comparison between the quadriceps force $F_q$ of this study and data available in the literature [19]. The resulting data of the two investigations show a similar progression throughout the exercise. Both investigations, [19] and present work, reported similar peaks during jumping 1.50 BW and 1.10 BW, and right after landing 1.70 BW and 2.50 BW, respectively. Present work is in agreement with data reported in the literature. Differences are due to the type of athlete and the way the exercise was performed.

2. Tibio-femoral Contact Point - Comparison of Descent Phases of Vertical Squat Jump and Regular Squat Exercises.

Since the vertical jump exercise is derived from a squat progression, the descent phases of regular squat, and vertical jump can be compared. The ascent phase is not subject of comparison since the vertical jump exercise is performed at a faster rate and higher intensity than the regular squat exercise. Figure 22 depicts the location of the contact point on tibia during the vertical jump in the present work and that of regular squat exercise, Murakami et al. [54]. References [26] and [27] were used to determine the femoral geometry, and Ref. [28] to determine the location of the tibiofemoral contact point at 90°knee flexion. Then calculations of the locations of the tibiofemoral contact point were carried over to the rest of the positions. The progression of the contact point is in good agreement with Ref. [54] for the first half of the descent phase of jumping. For the second half of the descent phase there is a difference. One should mention that Mukarami et al. [54] used healthy males performing squats under periodic x-ray images at a rate of 10 frames per second. They analyzed the in vivo three-dimensional kinematic parameters of subjects' knees,
namely the tibiofemoral flexion angle, anteroposterior (AP) translation, and internal–
external rotation, using serial x-ray images. The model in present work is a two-
dimensional model, so it does not account for the internal external rotation of the
knee, namely the screw home mechanism, which is significant in the second half of
the descent phase of jumping.

3. Conclusions. The present investigation offers new insight regarding internal
forces during the vertical jump exercise, a countermovement jump, Van Hooren and
Zolotarjova [55]. The present work used experimental ground reaction forces as well
as motion analysis data, and predicted the tibio-femoral, patello-femoral, and hip joint
contact forces, as well as the quadriceps and hamstrings muscle forces during vertical
jump exercise. All these forces are in agreement with data reported in the literature.

Moreover, the descent phase of jumping is compared with the descent phase of
regular squat exercise. The comparison shows that the motion of the tibio-femoral
contact point on the tibial plateau is in agreement with data reported in the literature.

An advancement in this study would be to investigate the effects of the arm swing,
change in landing style, or an improvement in the model. Improvement of this model
may include: more refined model of finding the tibiofemoral center of rotation, more
refined model of patellofemoral contact force calculation, anatomical refinement, to
include several fibers for each ligament behavior, and articular cartilage properties to
allow for deformable contact.

4. Limitations. The present model is not entirely an anatomical description of
the human knee. The femoral condyle is modeled as two circles approximating the
tibiofemoral and patellofemoral condylar contact arcs, and the tibial plateau is
modeled as a straight line, as shown in Fig. 3. The present model is a two-dimensional model. Therefore, medial and lateral contact forces cannot be assessed, and internal-external rotation cannot be captured. Moreover, two-dimensional models approximate muscle forces by their projections on the sagittal plane. The contact point, which is related to the tibiofemoral center of rotation, is properly captured. The transepicondylar motion is properly captured though the marker-based approach, but not the geometric center axis described in Refs. [30-31]. The geometric center would better represent the tibiofemoral center of rotation. Therefore, although the insertion points were anatomical, the magnitude of PCL force was calculated through an optimization process. Another limitation of this work is that only one set of experimental data is presented, and therefore there is no comparison between multiple sets of data of the exercise. Also, this work does not investigate the effect of maximum knee flexion angle of jumping on the exercise performance and/or internal forces’ magnitudes. The maximum knee flexion angle in this research is approximately 95°. Cases of deep knee flexion, Caruntu et al. [56], of jumping were not considered here.
REFERENCES


Dumitru I. Caruntu, corresponding author, CND-18-1432


FIGURE CAPTIONS

Figure 1. Free body diagram of femur

Figure 2. Free body diagram of tibia

Figure 3. Knee joint articular geometry model and ligament forces acting on femur

Figure 4. Patellofemoral flexion angle and patellar tendon angle versus tibiofemoral knee flexion angle.

Figure 5. Tibiofemoral center of rotation (COR) calculation at 90°-degree flexion, where TT is the tibial tuberosity marker, and IM is the intercondylar virtual marker calculated as the midpoint between LM and MM, the lateral and medial condyle markers, respectively. Solid lines represent either the tibial plateau and the simplified knee condyle geometry, dashed lines represent the ligaments at that current position; 1, 2, 3, and 4 indicate ACL, PCL, LCL, and MCL, respectively.

Figure 6. Biomechanics Laboratory (Director Caruntu) at University of Texas Rio Grande Valley.

Figure 7. Positions during the vertical jumps

Figure 8. Marker placement.

Figure 9. Vertical coordinate of the hip joint.

Figure 10. Experimental ground reaction forces during the vertical jump exercise.

Figure 11. Experimental flexion angles during the vertical jump exercise.

Figure 12. Experimental upper leg and lower leg angles \( \theta_1 \) and \( \theta_2 \), respectively, with respect to the horizontal axis during the vertical jump exercise.

Figure 13. Quadriceps, hamstrings and gastrocnemius muscle forces during the vertical jump exercise.
**Figure 14.** Contact Forces during the vertical jump exercise, tibiofemoral contact force $F_{cy}$, hip contact forces $F_{hx}$ and $F_{hy}$, and patellofemoral force $F_{cp}$

**Figure 15.** Motion of tibiofemoral contact point during the vertical jump exercise

**Figure 16.** Ligament forces during the vertical jump exercise

**Figure 17.** All significant forces during the vertical jump exercise

**Figure 18.** Vertical ground reaction force $R_y$, comparison for the vertical jump exercise

**Figure 19.** Knee tibio-femoral contact force, comparison for the vertical jump exercise

**Figure 20.** Hip contact force, comparison for the vertical jump exercise

**Figure 21.** Quadriceps force, comparison for the vertical jump exercise

**Figure 22.** Motion of tibiofemoral contact point, comparison for squat data during descent phase of vertical jump exercise. D is the tibial distance from the tibial posterior edge.
Figure 1. Free body diagram of femur

Figure 2. Free body diagram of tibia
Figure 3. Knee joint articular geometry model and ligament forces acting on femur

Figure 4. Patellofemoral flexion angle and patellar tendon angle versus tibiofemoral knee flexion angle.
Figure 5. Tibiofemoral center of rotation (COR) calculation at 90°-degree flexion, where TT is the tibial tuberosity marker, and IM is the intercondylar virtual marker calculated as the midpoint between LM and MM, the lateral and medial condyle markers, respectively. Solid lines represent either the tibial plateau and the simplified knee condyle geometry, dashed lines represent the ligaments at that current position; 1, 2, 3, and 4 indicate ACL, PCL, LCL, and MCL, respectively.

Figure 6. Biomechanics Laboratory (Director Caruntu) at University of Texas Rio Grande Valley.
Figure 7. Positions during the vertical jumps

Figure 8. Marker placement.
Figure 9. Vertical coordinate of the hip joint.
Figure 10. Experimental ground reaction forces during the vertical jump exercise.
Figure 11. Experimental flexion angles during the vertical jump exercise.
Figure 12. Experimental upper leg and lower leg angles \( \theta_1 \) and \( \theta_2 \), respectively, with respect to the horizontal axis during the vertical jump exercise.
Figure 13. Quadriceps, hamstrings and gastrocnemius muscle forces during the vertical jump exercise
Figure 14. Contact Forces during the vertical jump exercise, tibiofemoral contact force $F_{cy}$, hip contact forces $F_{hx}$ and $F_{hy}$, and patellofemoral force $F_{cp}$
**Figure 15.** Motion of tibiofemoral contact point during the vertical jump exercise
Figure 16. Ligament forces during the vertical jump exercise
Figure 17. All significant forces during the vertical jump exercise
**Figure 18.** Vertical ground reaction force $R_y$, comparison for the vertical jump exercise

**Figure 19.** Knee tibio-femoral contact force, comparison for the vertical jump exercise
Figure 20. Hip contact force, comparison for the vertical jump exercise

Figure 21. Quadriceps force, comparison for the vertical jump exercise
Figure 22. Motion of tibiofemoral contact point, comparison for squat data during descent phase of vertical jump exercise. $D$ is the tibial distance from the tibial posterior edge.