The Correlation of Segment Accelerations and Impact Forces With Knee Angle in Jump Landing

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Impact forces and shock deceleration during jumping and running have been associated with various knee injury etiologies. This study investigates the influence of jump height and knee contact angle on peak ground reaction force and segment axial accelerations. Ground reaction force, segment axial acceleration, and knee angles were measured for 6 male subjects during vertical jumping. A simple spring-mass model is used to predict the landing stiffness at impact as a function of (1) jump height, (2) peak impact force, (3) peak tibial axial acceleration, (4) peak thigh axial acceleration, and (5) peak trunk axial acceleration. Using a nonlinear least square fit, a strong ($r = 0.86$) and significant ($p \leq 0.05$) correlation was found between knee contact angle and stiffness calculated using the peak impact force and jump height. The same model also showed that the correlation was strong ($r = 0.81$) and significant ($p \leq 0.05$) between knee contact angle and stiffness calculated from the peak trunk axial accelerations. The correlation was weaker for the peak thigh ($r = 0.71$) and tibial ($r = 0.45$) axial accelerations. Using the peak force but neglecting jump height in the model, produces significantly worse correlation ($r = 0.58$). It was concluded that knee contact angle significantly influences both peak ground reaction forces and segment accelerations. However, owing to the nonlinear relationship, peak forces and segment accelerations change more rapidly at smaller knee flexion angles (i.e., close to full extension) than at greater knee flexion angles.

Key Words: accelerometers, landing ground reaction, jumping height

The forces on the internal muscles, bones, ligaments, and tendons during landing from a jump have been related to various joint pathologies. For example, jump landings have been associated with anterior cruciate ligament (ACL) injuries, various knee tendinopathies, and osteoarthritis (Richards et al., 1996; Hurwitz et al., 2000; and Murphy et al., 2003). However, the exact mechanisms of these injuries are still not known. Several studies have been undertaken to measure and calculate the internal forces in the knee in jump landing (Pflum et al., 2004; Decker et al., 2003; Kernozek et al., 2005). Previous studies have shown that tibial axial accelerations (i.e., along the long axis of the tibia) can be correlated with peak vertical ground reaction forces during jump landing and running (Elvin et al., 2007; Derrick et al., 2004; and Mizrahi & Susak, 1982).
The relationships between segment acceleration, impact force, and knee angle during functional activities have been measured but are not well understood. It has been estimated that 40% of all ACL injuries occur owing to rapid decelerations during landing (Boden et al., 2000; Kirkendall & Garrett, 2000). Increased ground reaction forces and decreased knee flexion angles during contact have been associated with increased incidents of ACL injuries (Yu et al., 2002; Chappell, 2002). Several training programs have been devised to help athletes reduce their ground reaction forces at contact (Myer et al., 2006; Prapavessis & McNair, 1999). However, the actual relationship between the ground reaction force and knee contact angle has not been studied.

Knee contact angles during jump landing can vary from 15° to 55° depending on jump height and level of fatigue (Fagenbaum & Darling, 2003). Peak ground reaction forces can be less than 2 body weights for soft landings, and they can be as high as 14.4 body weights for stiff landings (Denoth, 1986). Furthermore, knee angle plays an important role in joint reaction force and moment profiles and thus the internal forces of soft tissues that bridge the knee joint.

Only a few studies have investigated the effect of jump height on impact force, and these studies are contradictory. Zhang et al. (2000) and Fagenbaum and Darling (2003) showed, in general, that impact forces increase with landing height for drop landing. Elvin et al. (2007), by contrast, showed that when subjects were instructed to jump vertically and touch a marker, there is no correlation between jump height and impact force. The explanation for these contradictory results was that drop jumps tend to be more controlled compared with landings from vertical jump and touch tests (Elvin et al., 2007).

Modern data loggers can store and process sensor data at high sample rates (1,000 samples per second is typical). Previously, Elvin & Elvin (2006) have presented a small, lightweight, and wireless sensor that is unobtrusive and can be used to monitor leg impact accelerations in the field without significantly impeding athletic activity. Peak impact accelerations in jump landing have also been correlated with peak ground reaction forces (Elvin et al., 2007). This study investigates the relationship between impact force, impact segment accelerations, and knee contact angle with the future goal of using similar small-scale wireless accelerometers to better understand the kinematics of jump landing in the field.

In order to investigate more clearly what role external loads may have on internal knee joint structures, it is important to understand the relationship between jump performance characteristics and the associated external loads that occur during the jump and land sequence. In this study, we investigate the relationship between knee contact angle, jump height, peak ground reaction forces, and segment accelerations in landing from a vertical jump. Based on our previous work (Elvin et al., 2007), as opposed to reports of others (Zhang et al., 2000; Fagenbaum & Darling, 2003), we hypothesize that impact forces will decrease with increasing knee flexion; however, the relationship of knee flexion and impact forces will be nonlinear. We also hypothesize that impact axial accelerations will decrease with increasing knee flexion and that the relationship with knee flexion angle will be nonlinear.

**Methods**

**Subjects**

Based on the previous work of Elvin et al. (2007), six recreational male athletes (average age 22.8 ± 8.4 years) participated in the study. All subjects reported no history of orthopedic injury and signed an institution-approved informed consent form.

**Experimental design**

After a self-selected warm-up, the subject was instructed to perform a sequence of six sets of jumps. Each data set was collected over a period of 25 s. Each set of jumps contained between three and seven jumps depending on the self-selected jumping rate of the subject. In order to increase the variability of jump landing forces, the subjects were instructed to perform jumps that fell into six categories, namely, (1) high height jump with soft landing, (2) high height jump with hard landing, (3) medium height jump with soft landing, (4) medium height jump with hard landing, (5) low height jump with soft landing, and (6) low height jump with hard landing. In order to study the difference between hard and soft landings, the subject was asked to land with high contact forces in the hard landings, and low contact forces in the soft landings. The actual height and landing hardness was chosen by the subject.
No instruction was given on the strategy to change landing stiffness. The subject jumped with only his right foot on the force plate; the left foot took off and landed on the adjacent landing platform (located at the same height as the force plate). All jumps were performed with arms crossed over the chest. During the exercise, ground reaction forces were measured under the subject’s right foot using a force plate (AMTI Inc., Watertown, MA) sampling at 1,200 Hz. A seven-camera high-speed (120 frames per second, fps) motion analysis system (Motion Analysis Corp., Santa Rosa, CA) was used to collect the 3-D biomechanical marker data. Twenty-four retro-reflective spherical markers, 25 mm in diameter, were attached securely to the subjects in a standard Helen Hayes configuration (Kadaba et al., 1990): 5th metatarsus (left and right), calcaneus (left and right on the shoe), navicular (left and right), lateral malleolus (left and right), tibial shaft halfway between ankle and knee (left and right), tibial tuberosity (left and right), mid-thigh approximately 15 cm superior to the superior pole of the patella (left and right), femoral epicondyle (left and right), greater trochanter (left and right), pelvis (left and right), space between L5 and S1 vertebrae, acromion (left and right), and vertebra C7 (Figure 1).

Three segment axial accelerations were measured wirelessly using custom-built accelerometer sensor nodes (ZeroPoint Technologies, Johannesburg, South Africa). The uniaxial accelerometer consists of a ±70 g one-axis MEMS accelerometer (ADXL78 by Analog Devices Inc., Norwood, MA), buffer and amplification unit, a microcontroller, and a memory storage bank (Elvin & Elvin, 2006). Calibration of the sensor showed excellent linearity (coefficient of determination $r^2 = 0.9966$). The accelerometers store 64 s of axial accelerations at a sampling rate of 1,000 Hz. The acceleration data were uploaded to a PC after each set of jumps. The lightweight accelerometers were stitched onto tight-fitting elastic belts. Using these belts, the accelerometers were attached to the mid-shaft of the tibia, mid-thigh, and 5th lumbar vertebra (Figure 1). The axis of the shank accelerometers was aligned along the length of the tibia, the thigh accelerometer was aligned along the length of the femur, and the trunk accelerometer was aligned along the length of the spine.

Before and after each test, the subject was instructed to stand fully erect. Biomechanical marker data was collected corresponding to this initial position.

### Data Collection

The collected data are (a) segmental axial accelerations, (b) ground reaction force, and (c) segment displacements. The total number of jumps at each height is summarized in Table 1. Typically, each subject performed 30 jumps, with approximately 5 jumps for each of the prescribed height and landing rigidity.

![Image of accelerometer and biomechanical marker locations](image)

**Table 1** Number of Jumps Performed by Each Subject. The six groups of jumps are high soft (HS), high hard (HH), medium soft (MS), medium hard (MH), low soft (LS), and low hard (LH).

<table>
<thead>
<tr>
<th>Subject</th>
<th>Total number of jumps</th>
<th>Number of jumps of each group (HS/HH/MS/MH/LS/LH)</th>
</tr>
</thead>
<tbody>
<tr>
<td>S.1.</td>
<td>29</td>
<td>4/4/4/4/6/6</td>
</tr>
<tr>
<td>S.2.</td>
<td>30</td>
<td>5/6/4/5/5/5</td>
</tr>
<tr>
<td>S.3.</td>
<td>27</td>
<td>5/4/5/5/4/5</td>
</tr>
<tr>
<td>S.4.</td>
<td>30</td>
<td>4/4/5/5/7/5</td>
</tr>
<tr>
<td>S.5.</td>
<td>33</td>
<td>4/4/6/7/5/7</td>
</tr>
<tr>
<td>S.6.</td>
<td>36</td>
<td>6/6/6/6/6/6</td>
</tr>
</tbody>
</table>
Data Analysis

The marker coordinate data was tracked using commercial software (Motion Analysis Corp., Santa Rosa, CA) and custom Matlab software (Mathworks Inc., Natick, MA). At the start and end of each trial, the subject was instructed to stand erect and biomechanical data was collected in this anatomical position. Euler angle calculations were used to determine the three rotations and translations in a local coordinate system as defined by Grood and Suntay (1983); flexion-extension (sagittal plane motion) was the first rotation matrix. Knee angle was calculated as $\Theta = (\Theta_2 + \Theta_3)/2$. This angle is measured relative to the initial anatomical knee angle; $\Theta = 90^\circ$ is assumed to be full knee extension. The sagittal axis of the knee was taken as the intersection of the line joining the greater trochanter and the lateral epicondyle and the line joining the tibial tuberosity and the lateral malleolus. Accelerations were measured only in the axial direction (Figure 2). There was no filtering of the ground reaction force, biomechanical marker data set, and accelerometer data.

The jump height was calculated directly from the flight time as follows. Height ($H$) is given by $H = gt^2/8$, where $t$ is the total flight time (both up and down) and $g$ is the gravitational acceleration (Torvinen et al., 2003; Linthorne, 2001; Kubo et al., 1999, Lian et al., 1996).

The effect of height on landing impact is not well understood. Assuming the body acts as a 1 degree-of-freedom (df) system (Figure 3), where $m$ is the mass of the body, $k$ is the assumed body stiffness, and $c$ is the body’s damping, then $u$, the displacement of the mass at impact, and $F$, the ground reaction force, are given by the differential equations

$$m\ddot{u} + cu + ku = 0 \quad F = m(\ddot{u} - g) \quad (1)$$

Figure 2 — Measured biomechanical variables. The left figure shows the anatomical position. The right figure shows a typical “soft” landing position at peak impact force, and the corresponding segment accelerations.
Knee Angle in Jump Landing

The initial conditions are as follows:

Velocity at impact: \( \dot{u}(0) = \sqrt{2gh} \) as calculated from energy balance \hspace{1cm} (1a)

Acceleration at impact: \( \ddot{u}(0) = g \) the gravitational constant of acceleration \hspace{1cm} (1b)

The acceleration, \( a(t) \), is given by (Elvin & Elvin, 2006):

\[
a(t) = -e^{-\xi\omega t} \left( \left(\xi^2\omega^2 - \omega_0^2\right)(A \cos \omega_0 t + B \sin \omega_0 t) + 2\xi\omega\omega_0 \left(A \sin \omega_0 t - B \cos \omega_0 t\right) \right)
\]

where \( \xi = \frac{c}{2m\omega} \) is the damping ratio and \( \omega = \sqrt{\frac{k}{m}} \) is the natural frequency, \( \omega_0 = \omega \sqrt{1 - \xi^2} \) is the damped natural frequency for an underdamped system (i.e., where \( \xi < 1 \)), and

\[
A = -2\frac{\xi}{\omega^2} \frac{\dot{u}(0)}{\omega} - \frac{\ddot{u}(0)}{\omega^2} \quad \text{and} \quad B = \frac{\ddot{u}(0)}{\omega_0} \left(1 - 2\xi^2\right) - \frac{\ddot{u}(0)\xi}{\omega_0 \omega_0}
\]

The peak impact force \( (F_{\text{imp}}) \) is given by

\[
F_{\text{imp}} = m(a_{\text{imp}} - g)
\]

where \( a_{\text{imp}} \) is the maximum acceleration from Equation 2.

Assuming that damping \( (c) \) is approximately 0.35 (Derrick et al., 2002; Greene & McMahon, 1979), the only unknown in Equation 2 is the stiffness of the body \( (k) \). In this article, we assume that stiffness \( (k) \) is only a function of knee angle \( (\Theta) \).

The following procedure is used to calculate the optimal stiffness \( (k) \) which best approximates the peak impact force \( (F_{\text{imp}}) \). For a particular jump, the subject's mass \( (m) \), jump height \( (h) \), and an initial stiffness estimate of \( k = 5 \text{kN/m} \) is used in Equations 2 and 3 to calculate the peak impact force. The stiffness \( (k) \) is incrementally increased until it matches the measured force \( (F_{\text{imp}}) \) to within 0.01 kN. This calculated stiffness is referred to as the force-normalized stiffness \( (K_{F}) \) in this study. The normalized stiffness is plotted against knee contact angle \( (\Theta) \).

An analogous procedure is used to estimate the body stiffness using segment accelerations, where the maximum measured segment accelerations are compared with the maximum acceleration \( (a_{\text{imp}}) \) calculated from Equation 2. This calculated stiffness is referred to as the segment acceleration-normalized stiffness \( (K_{A}) \) in this study.

The fitted force-normalized stiffness \( (K_{F}) \) was used to back-calculate the impact forces for a range of knee angles assuming a 10-, 20-, 30-, and 40-cm jump height.
Statistics

Force-normalized stiffness ($K_f$) was correlated with contact knee angle using an exponential nonlinear least square fit of the form

$$K_f = a_f e^{b_f \alpha}$$

where $a_f$ and $b_f$ are curve fit parameters.

The form of Equation 3 is assumed from analysis of measured data and from the Greene and McMahon (1979) study that showed that leg stiffness is a nonlinear function of knee angle. Similarly, segment axial acceleration–normalized stiffness ($K_{sa}$) were correlated with contact knee angle using an exponential nonlinear least square fit of the form

$$K_{sa} = a_{sa} e^{b_{sa} \alpha}$$

where $a_{sa}$ and $b_{sa}$ are curve fit parameters for segment $i$.

Pearson correlation coefficient $r$ was calculated for the best fit. The significance of correlation was assessed using a two-tailed $t$ test with an $\alpha$ level of 0.05. All statistics were calculated using Sigmaplot (Systat Software, Inc. Richmond, CA).

To assess the significance of the jump-height normalization, a new stiffness ($\bar{K}_{FH}$) was calculated using the mean jump height ($\bar{H}$) for the entire set of the subject’s jumps. The same form of nonlinear least squares fit was performed:

$$\bar{K}_{FH} = a_{FH} e^{b_{FH} \alpha}$$

Table 2 - Coefficients of correlation ($r$) for normalized stiffness and knee contact angle ($\alpha$); $K_f$ is the force-normalized stiffness, $K_{as}$ is the shank acceleration–normalized stiffness, $K_{at}$ is the thigh acceleration–normalized stiffness, and $K_{ah}$ is the trunk acceleration–normalized stiffness. Asterisks indicate correlation significance ($\alpha = 0.05$).

<table>
<thead>
<tr>
<th>Subject</th>
<th>$r$ ($\bar{K}_f$)</th>
<th>$r$ ($\bar{K}_{as}$)</th>
<th>$r$ ($\bar{K}_{at}$)</th>
<th>$r$ ($\bar{K}_{ah}$)</th>
</tr>
</thead>
<tbody>
<tr>
<td>S.1</td>
<td>0.97*</td>
<td>0.66*</td>
<td>0.88*</td>
<td>0.91*</td>
</tr>
<tr>
<td>S.2</td>
<td>0.76*</td>
<td>0.22</td>
<td>0.47*</td>
<td>0.79*</td>
</tr>
<tr>
<td>S.3</td>
<td>0.83*</td>
<td>0.42*</td>
<td>0.77*</td>
<td>0.60*</td>
</tr>
<tr>
<td>S.4</td>
<td>0.82*</td>
<td>0.32</td>
<td>0.67*</td>
<td>0.77*</td>
</tr>
<tr>
<td>S.5</td>
<td>0.80*</td>
<td>0.46*</td>
<td>0.68*</td>
<td>0.90*</td>
</tr>
<tr>
<td>S.6</td>
<td>0.95*</td>
<td>0.62*</td>
<td>0.81*</td>
<td>0.91*</td>
</tr>
</tbody>
</table>

Pearson correlation coefficient $r$ was calculated for the best fit. The significance of correlation was assessed using a two-tailed $t$ test with an $\alpha$ level of 0.05. The significance of the jump-height normalization was determined by comparing the coefficients of correlation for $\bar{K}_{FH}$ and $\bar{K}_f$ using the Fisher $Z$ test with $\alpha = 0.05$.

Results

There was a strong ($r = 0.86$) and significant ($p \leq 0.05$) correlation between force-normalized stiffness ($\bar{K}_f$) and knee contact angle for all six subjects (Table 2). For all subjects, the force normalized stiffness increased with knee contact angle (Figure 4a). There was a moderate ($r = 0.45$) correlation between shank acceleration–normalized stiffness ($\bar{K}_{as}$) and knee contact angle for all six subjects (Table 2). This correlation was significant ($p \leq 0.05$) for four of the six subjects. For all subjects, the shank normalized stiffness increased with knee contact angle (Figure 4b).

There was a moderate-to-strong ($r = 0.71$) correlation between thigh acceleration–normalized stiffness ($\bar{K}_{at}$) and knee contact angle for all six subjects (Table 2). This correlation was significant ($p \leq 0.05$) for all subjects. For all subjects, the thigh acceleration normalized stiffness increased with knee contact angle (Figure 4c). There was a strong ($r = 0.81$) correlation between the trunk acceleration–normalized stiffness ($\bar{K}_{ah}$) and knee contact angle for all six subjects (Table 2). This correlation was significant ($p \leq 0.05$) for all six subjects. For all subjects, the trunk acceleration normalized stiffness increased with knee contact angle (Figure 4d).

There was a weaker correlation ($r = 0.58$) between force normalized stiffness when neglecting jump height as compared to when jump-height normalization was not neglected (Table 3). Comparing the coefficients of correlation using the Fischer $Z$ test with $\alpha = 0.05$ showed that for five of the six cases the correlation was significantly better when jump height is taken into consideration. Neglecting jump height also shows a greater scatter in the stiffness-angle plot (Figure 4a). For all subjects, the impact force increases nonlinearly with knee-angle (Figure 5). Furthermore, at a given knee angle, the force increases with jump height (Figure 5).
Discussion

The results of the study demonstrate a strong correlation between force-normalized stiffness and knee angle in jump landings. A strong correlation was also found between trunk acceleration–normalized stiffness and knee contact angle. There was weaker correlation between leg segment acceleration–normalized stiffness and knee contact angle. Neglecting height in the force stiffness normalization was shown to yield a poorer correlation with knee contact angle than when height was included in the normalization process.

The present study shows that impact forces decrease with increasing knee flexion; however, the relationship is nonlinear as opposed to the linear relationship found for running or as expected based on previous reports (Zhang et al., 2000; Fagenbaum & Darling, 2003). Furthermore, the running linear relationship cannot be extrapolated to larger knee flexion angles because this would likely lead to negative predicted peak impact forces. The nonlinear relationship proposed in this study gives a more physically realistic estimate of peak impact forces at larger knee flexion angles. This relationship has not been reported previously when considering the injury potential of jump or drop landings.

Elvin et al. (2007) has previously shown that there is weak correlation between jump height and peak impact forces in unconstrained jumping. The present study also shows a weak linear correlation between jump height and peak impact force (average $r = 0.35$). Zhang et al. (2000), on the other hand, have shown that impact forces increase with increasing drop-landing height. It is possible that during drop landings, the subjects are acutely aware of their landing technique, and are able to keep their leg stiffness within a very narrow range. The present study shows that if knee contact angle can vary, there is an effect of jump height.

It has been shown that knee contact angle plays an important role in determining peak ground reaction forces and trunk axial accelerations. The knee-contact angle plays a smaller role on impact leg accelerations. Several theories have been put forward in the literature to explain the effect of leg acceleration and running knee-contact angle; however, at present, this effect is not fully understood (Derrick, 2004). Derrick (2004) has shown that, in running, the leg acceleration can either increase or decrease with knee contact angle. A wobbling mass model has been proposed to account for these experimental results (Derrick et al., 2002). This study has shown that for jump landing, leg acceleration increases with knee-contact extension angle. However, the scatter is large (e.g., in Figure 4b), and thus leg accelerations could appear to be either increasing or decreasing especially if only the smaller knee contact angles found in running are considered.

The lack of a higher correlation of the given model to the experimentally measured and derived results could be due to (1) measurement errors and (2) an oversimplified model. The measurement errors in the present study have been extensively explained in Elvin et al. (2007) and are likely significantly smaller than the model errors. The measurement error is dominated by the rotation of the shank, thigh, and trunk axes relative to the global coordinate system. Elvin et al. (2007) have shown that neglecting the effect of axis rotation (as in this study) causes a maximum of a 20% error in acceleration data. This error could explain some of the lack of correlation in the accelerometer results. It

### Table 3

<table>
<thead>
<tr>
<th>Subject</th>
<th>$k_{FH}$</th>
<th>$k_F$ vs. $k_{FH}$</th>
</tr>
</thead>
<tbody>
<tr>
<td>S.1</td>
<td>0.87+</td>
<td>&lt;0.01*</td>
</tr>
<tr>
<td>S.2</td>
<td>0.37+</td>
<td>0.03*</td>
</tr>
<tr>
<td>S.3</td>
<td>0.48+</td>
<td>0.02*</td>
</tr>
<tr>
<td>S.4</td>
<td>0.62+</td>
<td>0.11</td>
</tr>
<tr>
<td>S.5</td>
<td>0.33</td>
<td>&lt;0.01*</td>
</tr>
<tr>
<td>S.6</td>
<td>0.78+</td>
<td>&lt;0.01*</td>
</tr>
</tbody>
</table>
Figure 4 — Effect of knee contact angle ($\Theta$) on (a) force normalized stiffness, (b) shank acceleration normalized stiffness, (c) thigh acceleration normalized stiffness, and (d) trunk acceleration normalized stiffness for subject S.4. Circles indicate the experimentally derived stiffness calculations. Lines indicate least-square best fits. Squares in (a) indicate the height normalization is neglected in the stiffness derivation. A knee contact angle of 90° indicates full knee extension.

should be noted that using a uniaxial accelerometer does not allow for the rotation of the acceleration into the global coordinate system.

The model errors in this study are mainly due to the assumption that the total body stiffness and mass can be represented by a single degree of freedom. The total stiffness of the body is assumed to be only a function of knee contact angle. The effects of muscle co-contraction, reflex stiffness, wobbling mass, out-of-plane forces, and hip and ankle angles have all been ignored. A more sophisticated multi-link computer model (e.g., the full body model of Pflum et al., 2004) that accounts for all the above-mentioned variables should be able to accurately predict the entire time history of the ground-reaction forces. However, these models require significantly more measured variables (such as EMG of the major muscle groups), model assumptions (such as the stiffness of various muscle groups, and segment masses), and computer optimization. The assumed critical damping ratio ($\zeta = 0.35$) is not a significant source of error in the model. Changing the critical damping ratio between 0.2 and 0.5—the range reported by Green and McMahon (1979)—does not significantly change the nonlinear fits in any of the results ($p < 0.01$ using the Fischer Z test).
A possible limitation of the study is that knee angle was not specifically controlled. However, the control of knee angle would likely result in unnatural landing kinematics. The maximum ground reaction forces was 12.2 body weights, which is less than the 14.4 body weights previously reported by Denoth (1986). Knee contact angle ranged between 36.9° and 82.3°, which is within 7° of those previously reported by Fagenbaum and Darling (2003).

Another potential limitation is the relatively low numbers of subjects utilized in the present study design. The correlations are generally strong and significant, which suggest adding more subjects or trials may merely increase the level of significance and power without necessarily increasing the general interpretation of the current results.

Further studies need to be performed to ascertain the influence of hip and ankle angles, and muscle co-contraction on peak impact forces. Moreover, the onset of fatigue has to be considered, because it is likely to affect the stiffness of landing and therefore the impact forces. The correlations of landing force, landing accelerations, and knee contact angle with various knee injury etiologies have to be investigated.

Acknowledgments

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References


