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Mechanical Modeling of Tibial Axial Accelerations Following Impulsive Heel Impact

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A fourth order mass/spring/damper (MSD) mechanical model with linear coefficients was used to estimate axial tibial accelerations following impulsive heel impacts. A generic heel pad with constant stiffness was modeled to improve the temporal characteristics of the model. Subjects (*n* = 14) dropped (~5 cm) onto a force platform (3 trials), landing on the right heel pad with leg fully extended at the knee. A uni-axial accelerometer was mounted over the skin on the anterior aspect of the medial tibial condyle inferior to the tibial plateau using a Velcro™ strap (normal preload ~45 N). Model coefficients for stiffness (k_1, k_2) and damping (c_1, c_2) were varied systematically until the minimum difference in peak tibial acceleration (%PTA_{min}) plus maximum rate of tibial acceleration (%RTA_{max}) between the estimated and measured curves was achieved for each trial. Model responses to mean subject and mean group model coefficients were also determined. Subject PTA and RTA magnitudes were reproduced well by the model (%PTA_{min} = 1.4 ± 1.0 %, %RTA_{min} = 2.2 ± 2.7%). Model estimates of PTA were fairly repeatable for a given subject despite generally high variability in the model coefficients, for subjects and for the group (coefficients of variation: CV_{k1} = 57; CV_{k2} = 59; CV_{c1} = 48; CV_{c2} = 85). Differences in estimated parameters increased progressively when subject and group mean coefficients (%PTA_{sub} = 8.4 ± 6.3%, %RTA_{sub} = 18.9 ± 18.6%, and %PTA_{grp} = 19.9 ± 15.2 %, %RTA_{grp} = 30.2 ± 30.2%, respectively) were utilized, suggesting that trial specific calibration of coefficients for each subject is required. Additional model refinement seems warranted in order to account for the large intra-subject variability in coefficients.

Key Words: drop impacts, visco-elastic response, coefficient variability

Introduction

The accumulated effects of impact forces have been shown to contribute to the onset of various microtrauma injuries in the joints of the support limbs of small mammals (Simon et al., 1972; Radin et al., 1973; Radin et al., 1982; Serink et al., 1977) and have also been cited as a contributing factor for a variety of musculoskeletal maladies in humans, including knee and back pain (Light et al., 1980; Voloshin & Wosk, 1981, 1982), and articular cartilage degeneration and osteoarthritis (Buckwalter & Lane, 1997). In order to reduce

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the injurious effects of impact loading in sport, recreational, and occupational activities, a better understanding is needed of how forces are transmitted through the hard tissues (bone) and soft tissues (muscle, fat, connective tissue, etc.) of the body, and how these tissues attenuate peak forces naturally. By accomplishing this, human impact responses can be modeled such that the analysis of a variety of impact scenarios in sport and work activities can be made without additional subject testing.

Attenuation of forces as they are transmitted through the body following impacts has been well documented in humans (Chu et al., 1986; Derrik et al., 1997; Kim et al., 1993; Lafortune et al., 1996b; Light & McLellan, 1977; Light et al., 1980; McGill et al., 1989; Pope et al., 1987; Pope et al., 1997; Wosk & Voloshin, 1981) and highlights the non-rigid properties of biological tissues. The visco-elastic response of the human body to impacts following vertical drops has been modeled using simple mechanical systems comprised of springs and dampers with linear coefficients (Mizrahi & Susak, 1982a, 1982b; Özgüven & Berme, 1988). An advantage of this modeling approach is that the dynamics of low order, low degree of freedom mechanical systems are fairly easily implemented, and can be used to determine the magnitude of the visco-elastic coefficients in vivo. For example, Mizrahi and Susak (1982b) used a two-degree-of-freedom, fourth-order mechanical model to simulate transmitted accelerations to the level of the greater trochanter following impacts from vertical drops of about 5 cm. Actual accelerations were estimated using a pre-loaded, low mass, surface mounted uni-axial accelerometer. Linear coefficients of stiffness and viscosity were determined between the lumped masses of the lower extremity and the rest of the body. The model was reported to provide satisfactory predictions of peak accelerations following transmission through the entire lower extremity. However, a total of 2 subjects were analyzed, thereby limiting the assessment of the within and between subject variability of the model coefficients and the resulting estimates of transmitted peak acceleration.

Therefore, the purposes of this study were to simulate the axial tibial accelerations of subjects following physiological drop impacts using a fourth order mass/spring/damper mechanical model with linear coefficients, and to determine the magnitude and variability of the stiffness and viscosity coefficients required by the model to estimate the in vivo peak tibial response obtained by surface mounted accelerometry.

Methods

Fourteen healthy subjects (22.0 ± 6.3 years; 1.69 ± 0.09 m; 64.9 ± 13.0 kg) participated in this study. Subjects were informed of all aspects of the experiment and signed a consent form prior to participation.

A small uniaxial accelerometer (50 g capacity, Durham Instruments) was mounted to a small aluminum plate (total mass, 18.2 g) and placed over the skin on the anterior aspect of the medial tibial condyle of the right leg of subjects, just inferior to the tibial plateau. The accelerometer was preloaded with a normal force of approximately 45 N using a Velcro™ strap, which encircled the superior portion of the leg below the knee. Once instrumented, subjects were instructed to grasp a metal bar with both hands and hang vertically above a force platform (AMTI model #OR6-5), with the right leg fully extended. The height of the bar was adjusted for each subject such that the drop height (distance between the bottom of the heel of the right foot and the force platform prior to letting go of the bar) was approximately 5 cm. The foot of the landing limb was bare and was dorsiflexed enough to allow for contact to be made directly with the heel pad. Subjects were instructed to maintain the extended position of the landing limb during impact. Three drops were recorded for each subject following sufficient practice with the task to enable subjects to perform the task confidently and with minimum loss of balance. An assistant was always present to help anyone who needed external support. Accelerometer signals and vertical ground reaction forces from the force platform were A/D converted at 2000 Hz and then processed using in-house software.

Pilot tibial acceleration data of living subjects revealed a short latency period that existed immediately following contact of the heel pad with the ground (Figure 2b). During this delay, tibial accelerations increased fairly linearly for between 5 and 8 ms until rising rapidly with respect to the linear trend. This latency was attributed to the compliant deformation of the heel pad prior to calcaneous impact. The time delay was uncorrelated with body mass and was modeled as a linear spring with a generic stiffness (k_{hp} in Figure 1) normalized to body mass. Based on the pilot data, a constant delay of 7 ms was chosen as representative of the response seen for all subjects. This delay allowed the heel pad to deform approximately 7 mm following a drop landing from 5 cm. The change in acceleration and velocity associated with the compression of the heel pad spring was small, from –9.81 m/s^2 and –0.99 m/s just prior to impact (after freefall), to –8.0 m/s^2 and –1.05 m/s following full spring deformation, respectively. These latter values represent the initial conditions of the tibial impact response following heel pad deformation. Following heel pad compression, the spring did not contribute to the tibial response.

At the point of calcaneous impact, a fourth-order, two-degree-of-freedom (DOF), mass/spring/damper mechanical model was used to simulate the movement of the foot/leg segment mass (m_1), and mass of the rest of the body above the knee (m_2; Figure 1). The mass of the foot/leg sement was estimated to be 0.061 M, where M = $m_1 + m_2$ = the total mass of the body (Winter, 1990). Vertical ground reaction forces (F_y) were applied to a massless contact plate m_0. The motions of m_1 and m_2 are described by equations 1 and 2, respectively. Vertical accelerations of m_1 and m_2 are denoted by $\ddot{y}_1$ and $\ddot{y}_2$, and linear stiffness and damping coefficients are represented by k_1 and k_2, and c_1 and c_2, respectively.

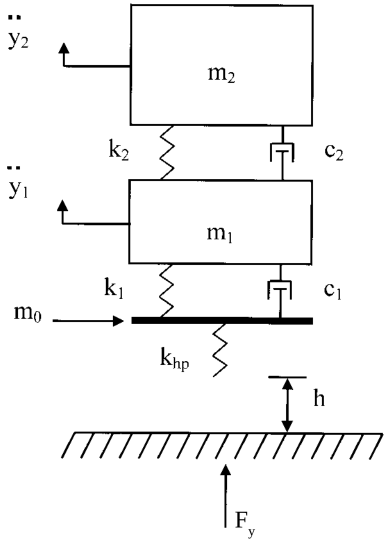

Figure 1 — Fourth-order, two-degree-of-freedom, mass/spring/damper mechanical model of the leg and body used to estimate tibial accelerations. k_{hp} is the stiffness of the generic heel pad.

$$m_1 \cdot \ddot{y}_1 + (c_1 + c_2)\dot{y}_1 + (k_1 + k_2)y_1 - c_2 \cdot \dot{y}_2 - k_2 \cdot y_2 = F_y \quad (1)$$

$$m_2 \cdot \ddot{y}_2 + c_2(\ddot{y}_2 - \dot{y}_1) + k_2(y_2 - y_1) = 0 \quad (2)$$

The magnitude of the vertical ground reaction force (F_y) at any time was estimated from the accelerations and magnitudes of the masses m_1 and m_2 for each subject (equation 3).

$$F_y = [\ddot{y}_1 \cdot m_1 + \ddot{y}_2 \cdot m_2 + (m_1 + m_2)g] / (m_1 + m_2)g \quad (3)$$

The motion of the model was governed by the following initial conditions at the time of calcaneous impact following heel pad deformation. The initial lengths of springs 1 and 2 were set such that the model did not bottom out following the drop and subsequent impact.

$y_1 = 0.01$	Relaxed length of spring 1 (m)
$\dot{y}_1 = -1.05$	Vertical velocity of m_1 after heel pad deformation (m/s)
$\ddot{y} = -8.0$	Vertical acceleration of m_1 after heel pad deformation (m/s^2)
$y_2 = 0.11$	Relaxed length of spring 2 (m)
$\dot{y}_2 = -1.05$	Vertical velocity of m_2 after heel pad deformation (m/s)
$\ddot{y}_2 = -8.0$	Vertical acceleration of m_2 after heel pad deformation (m/s^2)

The values of the linear stiffness coefficients (k_1 and k_2) and viscosity coefficients (c_1 and c_2) were manipulated throughout a range of 10,000 possible combinations (Table 1) until the optimal match between the estimated tibial acceleration and the measured subject tibial acceleration was achieved for each trial. Acceleration curves were compared based on two criterion variables: peak tibial acceleration (PTA; m/s^2) and maximum rate of tibial acceleration (RTA_{max}; g/s). These criteria were chosen based on their injury causing potential. The "optimal" match between the estimated and measured tibial accelerations was defined by that combination of coefficients which resulted in the minimization of the total relative error in PTA and RTA_{max} (i.e., $\%PTA_{min}$ and $\%RTA_{min}$, respectively). The coefficients that resulted in the minimum error ($\%PTA_{min}$ plus $\%RTA_{min}$) were deemed to be the optimal solution for the range of values specified and are referred to as *trial specific coefficients*.

Two secondary analyses were also performed to further test the model. In order to determine the variability associated with using a single set of four coefficients, the model was re-run using the mean trial specific coefficients for each subject (*subject mean coefficients*), and the overall mean of the trial specific coefficients for all subjects (*group mean coefficients*). The relative differences in PTA and RTA_{max} as a result of the use of the

Table 1 Stiffness and Damping Coefficient Constraints Used for Model Simulations of Tibial Accelerations

Constraint	k_1 (kN/m)	k_2 (kN/m)	c_1 (N · s/m)	c_2 (N · s/m)
Minimum	50	5	0	0
Maximum	500	50	1000	2000
Step	50	5	100	200

subject mean coefficients and the group mean coefficients were referred to as $\%PTA_{sub}$ and $\%RTA_{sub}$, and $\%PTA_{grp}$ and $\%RTA_{grp}$, respectively.

Results

A representative vertical ground reaction force and tibial acceleration curve following impact is included in Figure 2a and 2b, respectively. The magnitudes of measured PTAs ranged from 73.3 ± 7.6 m/s^2 to 157.9 ± 8.2 m/s^2 for the subjects studied (Table 2). The mean subject PTA (101.9 m/s^2) corresponds to approximately 10.4 times gravity. Mean peak vertical ground reaction forces (F_y) were 2.68 ± 0.4 times body weight (bw). The range in RTA_{max} was quite large, from a mean of 1022.7 ± 54.9 g/s for subject 11 to 2676.1 ± 136.6 g/s for subject 9.

In general, the within-subject variability associated with both PTA and RTA_{max} values was found to be approximately half what the variability was between subjects (Table 2). Coefficients of variation within subjects were in the order of 10 for most subjects (see means and standard deviations) compared to over 20 for the entire group.

Subject PTA and RTA_{max} magnitudes were reproduced well by the model when the optimal trial specific coefficients were utilized (Table 2 and Figure 3). Relative differences between the measured and estimated curves ranged from 0.4 ± 0.3% to 2.9 ± 1.9% for $\%PTA_{min}$, and 0.5 ± 0.4% to 5.3 ± 8.1% for RTA_{min}. Measured peak vertical ground reaction forces (F_y) were also reproduced fairly well by the model (mean of 2.49 ± 0.51 bw), but were on average slightly more variable than those measured by the force platform (CV = 20 vs. 15).

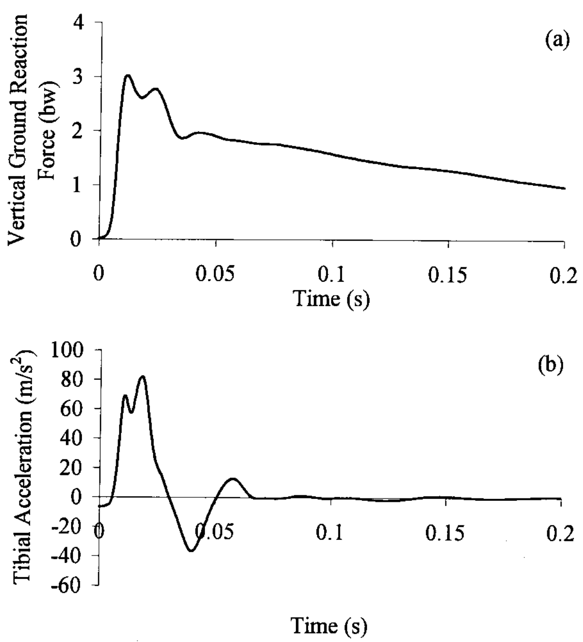

Figure 2 — Measured vertical ground reaction force (a) and associated axial tibial acceleration (b) for a typical subject.

Although model estimated PTA and RTA_{max} values were fairly repeatable for a given subject, the coefficients needed by the model to obtain the optimal match in PTA plus RTA_{max} were much more variable in general. Mean trial specific model coefficients were found to be 173.8 ± 99.0 kN/m, 24.0 ± 14.2 kN/m, 307 ± 146 N · s/m, and 681 ± 582 N · s/m for k_1, k_2, c_1, and c_2, respectively. The group mean trial specific coefficients were also more variable than the group mean acceleration variables—larger by a factor of between two and three.

In order to determine how sensitive the PTA and RTA_{max} outputs of the model were to the coefficients used, separate analyses were performed using the subject mean and group mean coefficients applied to the data for each trial. Relative differences in the estimated parameters (PTA and RTA_{max}) increased progressively when subject and group mean coefficients were utilized ($\%PTA_{sub}$ = 8.4 ± 6.3%; $\%RTA_{sub}$ = 18.9 ± 18.6%; $\%PTA_{grp}$ = 19.9 ± 15.2%; $\%RTA_{grp}$ = 30.2 ± 30.2 %, respectively). These differences were between 4 and 21 times greater than those that resulted from the model analyses utilizing coefficients specific to each trial.

Discussion

The mean subject peak tibial acceleration in this study was 10.4 times gravity (g). Comparable values of 9.1 g and 11.2 g were reported by Lafortune et al. (1995) from bone mounted and surface mounted accelerometers during running, and 9.4 g from a surface mounted accelerometer during pendulum impacts with the legs of subjects maintained in extension (Lafortune et al., 1996a). The magnitudes of the measured peak vertical ground reaction forces following the 5 cm drops also fell within the range of two to three times body weight that is typically seen in running (Cavagnagh & Lafortune, 1980). These results indicated that drop landings, as described in this study, seem to be reasonable for inducing physiological tibial impact responses, at least in terms of the peak magnitude of the applied forces and the peak axial tibial acceleration.

However, the drop impact method likely contributed to the variability seen in subject PTAs and model coefficients. A 5 cm block was used as a spacer, while the height of the bar was adjusted for each subject to accommodate subjects in full stretch. Although the exact height of each drop was not determined, vertical movements were noted visually, and trials were repeated if sizeable displacements from 5 cm were detected. Since the initial conditions of velocity in the simulations were assumed to be those consistent with a drop of 5 cm, any deviation from this during subject testing would be a source of varia-

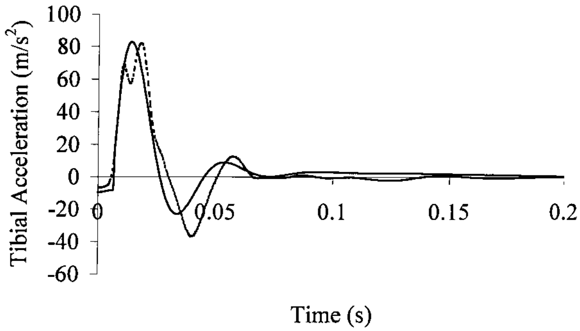

Figure 3 — Measured (subject -----) and estimated (model ——) axial tibial accelerations for a typical subject (same subject as Figure 2).

Table 2 Mean (*SD*) Measured (Subject) and Estimated (Model) Variables for the Three Drop Trials

S	Measured peak variables PTA (m/s^2)	Measured peak variables RTA_{max} (g/s)	Measured peak variables Fy_{max} (bw)	Estimated peak variables PTA (m/s^2)	Estimated peak variables RTA_{max} (g/s)	Estimated peak variables Fy_{max} (bw)	Model k_1 (kN/m)	Model k_2 (kN/m)
1	73.3 (7.6)	1836.8 (181.7)	2.80 (0.20)	73.4 (8.0)	1832.4 (125.8)	2.33 (0.21)	116.7 (28.9)	23.3 (5.8)
2	109.3 (9.5)	1671.8 (244.9)	2.68 (0.22)	110.6 (8.6)	1691.1 (238.0)	2.56 (0.44)	133.3 (28.9)	23.3 (18.9)
3	122.9 (10.0)	1763.3 (156.4)	3.11 (0.25)	121.2 (7.8)	1687.7 (238.0)	2.92 (0.72)	183.3 (57.7)	33.3 (24.7)
4	81.9 (11.7)	1396.6 (137.5)	2.65 (0.18)	80.8 (10.8)	1378.5 (180.8)	2.58 (0.95)	150.0 (100.0)	33.3 (24.7)
5	97.0 (17.9)	1209.7 (266.7)	2.33 (0.28)	97.9 (19.6)	1209.8 (262.9)	2.25 (0.67)	216.7 (175.6)	16.7 (20.2)
6	96.4 (14.9)	2005.2 (123.7)	2.68 (0.15)	96.4 (15.0)	2064.4 (91.1)	2.17 (0.49)	166.7 (28.9)	16.7 (12.6)
7	103.6 (12.7)	2009.1 (336.0)	2.31 (0.16)	103.6 (13.6)	2030.1 (366.4)	2.34 (0.99)	133.3 (57.7)	18.3 (23.1)
8	112.2 (15.3)	2075.0 (397.2)	3.25 (0.32)	112.7 (15.5)	2109.8 (390.7)	2.29 (0.78)	200.0 (86.6)	20.0 (26.0)
9	157.9 (8.2)	2676.1 (136.6)	3.12 (0.22)	158.1 (11.3)	2695.3 (186.0)	2.67 (0.79)	266.7 (160.7)	23.3 (23.6)
10	100.3 (6.4)	1939.3 (128.4)	2.62 (0.16)	99.6 (6.0)	1949.2 (117.5)	2.85 (0.90)	233.3 (115.5)	23.3 (23.6)
11	83.4 (4.4)	1022.7 (54.9)	2.22 (0.09)	84.7 (4.6)	1010.4 (45.2)	2.57 (0.55)	116.7 (76.4)	25.0 (18.0)
12	80.8 (4.3)	1323.5 (127.4)	2.17 (0.05)	80.8 (4.7)	1302.1 (105.8)	2.11 (0.55)	100.0 (0)	16.7 (20.2)
13	82.4 (16.7)	2100.7 (446.9)	2.48 (0.27)	82.1 (13.7)	2055.4 (491.2)	2.64 (0.90)	116.7 (76.4)	38.3 (16.1)
14	125.3 (16.2)	1554.5 (279.6)	3.08 (0.28)	127.6 (18.5)	1557.5 (270.5)	2.54 (0.57)	300.0 (150.0)	38.3 (16.1)
Group	101.9 (23.7)	1756.0 (444.7)	2.68 (0.40)	102.1 (24.1)	1755.3 (455.7)	2.49 (0.51)	173.8 (99.0)	24.0 (14.2)
CV	23	25	15	24	26	20	57	59

Note. Model outputs are for the coefficient combination which produced the optimal match to subject PTA plus RTA_{max}. Relative differences are provided for trial specific ($\%PTA_{min}$, $\%RTA_{min}$), subject mean ($\%PTA_{sub}$, $\%RTA_{sub}$), and group mean ($\%PTA_{grp}$, $\%RTA_{grp}$) coefficient analyses

coefficients		Relative differences measured versus estimated					
c^1 (N·s/m)	c^2 (N·s/m)	$\%PTA_{min}$ (%)	$\%RTA_{min}$ (%)	$\%PTA_{sub}$ (%)	$\%RTA_{sub}$ (%)	$\%PTA_{grp}$ (%)	$\%RTA_{grp}$ (%)
133	733	0.4	2.0	7.1	8.8	35.9	13.4
(58)	(416)	(0.3)	(1.7)	(4.5)	(6.3)	(13.2)	(11.2)
367	200	1.2	1.9	6.0	12.3	9.8	25.3
(58)	(200)	(1.2)	(2.4)	(4.8)	(6.7)	(7.8)	(17.8)
400	533	1.6	5.3	6.9	8.3	14.1	16.2
(0)	(577)	(2.3)	(8.1)	(1.9)	(2.1)	(7.2)	(8.9)
233	867	1.4	3.9	11.6	32.6	27.8	48.6
(115)	(902)	(1.9)	(0.6)	(7.6)	(13.7)	(20.0)	(15.3)
367	1000	1.9	0.5	15.0	59.7	13.8	77.1
(153)	(1000)	(0.2)	(0.4)	(10.6)	(37.9)	(10.1)	(42.1)
267	800	0.2	5.1	11.9	4.2	13.3	4.4
(58)	(346)	(0.1)	(1.8)	(7.3)	(4.5)	(14.4)	(4.6)
233	467	2.0	1.7	8.1	16.6	10.8	11.1
(58)	(503)	(1.4)	(0.9)	(6.4)	(11.1)	(5.9)	(8.4)
333	733	1.3	2.1	10.1	17.0	11.9	14.4
(58)	(416)	(0.3)	(2.3)	(8.7)	(16.6)	(3.7)	(9.8)
500	400	1.4	1.7	3.9	13.7	35.3	22.6
(100)	(693)	(1.0)	(1.7)	(2.41)	(6.0)	(3.4)	(4.1)
500	400	1.5	0.5	4.3	18.1	4.3	5.8
(100)	(693)	(0.7)	(0.8)	(5.3)	(7.7)	(5.0)	(2.9)
267	600	1.7	1.2	3.9	16.6	27.8	97.2
(58)	(721)	(1.0)	(1.3)	(2.2)	(6.3)	(6.7)	(10.6)
233	200	0.7	1.7	4.3	8.0	7.9	41.0
(58)	(0)	(0.4)	(1.3)	(1.7)	(2.8)	(6.0)	(14.3)
600	867	2.9	2.5	13.8	30.0	41.5	13.5
(100)	(611)	(1.9)	(3.9)	(7.9)	(25.3)	(26.4)	(15.5)
600	867	1.9	0.6	10.8	19.4	24.2	32.5
(100)	(611)	(1.4)	(0.6)	(6.0)	(15.3)	(9.6)	(22.9)
307	681	1.4	2.2	8.4	18.9	19.9	30.2
(146)	(582)	(1.0)	(2.7)	(6.3)	(18.6)	(15.2)	(30.2)
48	85						

tion in the model coefficients. Since these deviations were likely very small, this was not considered to be a major source of within-subject variation. Measurements were also not made to confirm that the landing limb was kept vertically extended during impact. However, dropping from a height of only 5 cm does not allow much deviation from the vertical to occur once the subjects release. If deviations were observed, trials were rerun. Slight deviations from the vertical would reduce the magnitudes of the measured peak accelerations relative to true vertical and would likely contribute to the variability seen in the peak tibial accelerations for a given individual. During pendulum impacts, peak tibial accelerations have been shown to increase as the knee angle increases when the orientation of the leg is kept perpendicular to the impact surface (Lafortune et al., 1996a). Since the model used in this study neglects the possible rotational degrees of freedom during impact, it would fit coefficients to the measured waveforms, thereby underestimating the accelerations that would exist if the legs were truly vertical. The authors agree that more controlled limb position at impact, such as with a pendulum-type design (e.g., Lafortune & Lake, 1995; Kim et al., 1993) would likely reduce the variability of estimated peak accelerations and model coefficients by providing more consistent initial conditions.

Non-invasive measurement of the impact response of the leg using skin mounted accelerometers is prone to artifact caused by skin movement relative to the underlying bone (Kim et al., 1993; Lafortune et al., 1995; Morris, 1973; Saha & Lakes, 1977; Ziegert & Lewis, 1979). This relative movement depends on factors including accelerometer mass (Valiant et al., 1987; Ziegert & Lewis, 1979), degree of pre-loading (Saha & Lakes, 1977; Valiant et al., 1987), and the amount of soft tissue existing between bone and transducer (Saha & Lakes, 1977). In the current study, normal preloads of approximately 45 N were applied to push the transducer closer to the bone. Loading much beyond this was too uncomfortable for subjects and interfered with blood circulation to the landing foot. The location used for mounting the accelerometer, on the antero-medial aspect of the tibial condyle, was also chosen to minimize the movement artifact, since the skin here is relatively thin. Despite these limitations, low mass surface mounted accelerometers that have been sufficiently preloaded, have been used successfully in the non-invasive recording of tibial peak accelerations in living subjects (Mizrahi & Susak, 1982b; Lafortune & Lake, 1995; Valiant et al., 1987; Wosk & Voloshin, 1981; Ziegert & Lewis, 1979). Also, given the invasive alternative of mounting the accelerometer to pins imbedded into bone, estimates obtained from surface mounted transducers are required of any study involving more than just a few human subjects. It must be emphasized that the method for determining coefficients outlined here was non-invasive, and regardless of whether the subject data is collected using surface or bone mounted accelerometers, the model will optimally match it by varying the coefficients as described. The authors acknowledge that small errors in the surface mounted signals compared to bone are likely, but the quantification of this error was not the purpose of this study. Future development of the model will include the determination of a transfer function which can be used to adjust accelerations from surface mounted transducers to be more representative of the motion of the underlying bone. Techniques such as this have been reported previously (e.g., Katazahi & Griffin, 1995; Kim et al., 1993; Lafortune et al., 1995), but ultimately require knowledge of the true motion of the underlying bone, and therefore invasive methodology.

The tibial accelerations of all subjects tested exhibited a relatively flat toe region that lasted between 5 and 8 ms after heel pad impact with the force platform. Pilot work indicated that the simulated springs in the model had to be very stiff in order to match the initial peaks and maximum rates of subject tibial accelerations. This resulted in a rise in the model estimated acceleration immediately after impact, which was considerably more rapid than exhibited by the toe region of subject responses. It was apparent that the model needed an additional element in order to successfully simulate both the magnitudes and temporal characteristics of subjects. The short delay just after impact was attributed to the considerable deformation of the heel pad, which occurs during foot strikes such as in running (De Clercq et al., 1994), and was modeled as a generic heel pad with a constant stiffness normalized to body weight. Assuming drops of 5 cm, the velocity at impact was 0.99 m/s. At this impact velocity, the heel pad deformed approximately 7 mm in 7 ms, a time period that was between the 5 ms to 8 ms range documented for the group of subjects studied. A heel pad deformation of 7 mm is slightly lower than previously reported mean estimates of 8.8 mm and 9 mm during barefoot impacts with a pendulum (Lafortune et al., 1996a), and during running (De Clercq et al., 1994), respectively. Since the exact drop height was not known in the current study, the actual impact velocities of subjects could have been less than the model estimated 0.99 m/s if they dropped from less than 5 cm. If this did occur, it would have resulted in reduced delay following impact, and hence, reduced heel pad deformation. If a longer time delay of 9.5 ms is assumed, the heel pad model used in this study estimates an associated heel pad deformation of 9.3 mm. A mechanical system incorporating a separate foot segment with a deformable foot/ground interface of higher order may further improve simulations of the measured response seen here. Regardless, the simple generic heel pad modeled here was able to account for much of the time delay in the recorded tibial responses and the slight alterations to the acceleration and velocity that occurred between the times of heel pad and calcaneous impact.

Comparison of leg stiffness coefficients to those from other lumped parameter models in the literature is difficult as a result of the numerous differences in model attributes and the variety of impact methodologies utilized. The wide range of approaches used in the literature is similarly reflected in the range of reported leg stiffness estimates. Using similar models, Mizrahi and Susak (1982b) and Özgüven and Berme (1988) reported stiffness estimates that ranged from a mean of 5.32 ± 0.36 kN/m to a mean of 142.7 ± 10.5 kN/m for single subjects, respectively. These represent values that are between about 32 and 1.2 times less than the group mean estimate for k_1 reported here. Some of this discrepancy for the Mizrahi and Susak (1982b) estimates may be accounted for by the fact that they were for the entire lower extremity (to the level of the greater trochanter) and not just the leg. He et al. (1991) showed leg stiffness values in the order of 10 kN/m for a variety of speeds and across fractional levels of simulated gravitational loading. The model used by He et al. (1991) lumped all body mass at the body center of gravity and considered the entire landing limb as a single linear spring with no damping (McMahon & Cheng, 1990). Stiffness estimates of Lafortune et al. (1996a) resulting from pendulum impacts were 28.7 kN/m for straight legs, approximately six times smaller than the mean value of 173.8 kN/m estimated here, despite the fact that peak shank accelerations were of comparable magnitude (10.2 g).

The within-subject variability of the model coefficients was high despite fairly repeatable peak accelerations in most cases. This suggests that even for 1 subject doing the same task several times, consistent estimates of stiffness and viscosity are not likely using the current methodology, making it difficult to choose one value for any of the coefficients that will result in an accurate representation of a subject's general response. This is reinforced by the results of the two additional analyses which clearly showed that the relative differences in PTA and RTA_{max} between measured and estimated curves progressively increased when subject mean and then group mean coefficients were used for each trial. In order to improve model estimates using subject mean coefficients, model refinement with additional elements seems warranted.

The model coefficients in this study were linear and represented the lumped stiffness and damping properties of all leg tissues including muscle and bone, articular cartilage, skin, fat, and the connective tissue binding all structures together. Although the mechanical properties of individual biological tissues are not represented well by linear coefficients, they were used in this study for the two lumped masses to compare the results to previously reported models of similar composition. The inability of the model without the simulated heel pad to follow the temporal aspects of the tibial accelerations just after impact indicated that a linear coefficient model with so few elements was inadequate in terms of an overall simulation of reality. The lumped nature of the model also limits understanding of the interaction between the tissues and their individual capacities to attenuate or transmit force and also to withstand injury. However, with respect to injury prevention in a global sense, being able to reduce the magnitude and rate of applied forces transmitted to structures within the knee, for example, has considerable value. Although the current model cannot provide loading information for specific tissues, it does provide fairly repeatable estimates of peak tibial acceleration and maximum rates of tibial acceleration.

The relative contribution of bone and soft tissues of the leg to the attenuation of impact forces has been addressed previously in rabbits (Paul et al., 1978) and with amputated human lower limbs (Cornelissen et al., 1986). However, quantification of the relative effects of wobbling mass (Gruber et al., 1987; Gruber et al., 1998) in living subjects is needed. Recently, Gruber et al. (1998) showed that biomechanical modeling of segment impacts using traditional rigid segment model assumptions resulted in incorrect internal torques and forces compared to a model that incorporated wobbling masses and visco-elastic connections. Rigid segment analysis has also been shown to overestimate the magnitude of transmitted peak forces through the trunk when the applied forces are impulsive in nature (McGill et al., 1989). But, widespread incorporation of non-rigid elements into existing biomechanical models is currently limited by a number of practical methodological considerations that include, but are not restricted to: an inability to easily determine the relative mass of segment tissues (e.g., muscle, bone) for living individuals; an inability to accurately measure, non-invasively, the true motion of body segments and their tissues in vivo without corresponding skin motion artifact; and an inability to determine, non-invasively, the visco-elastic parameters that describe the nature of the physical connection between tissue components or segments in vivo (Trujillo & Busby, 1990).

Summary and Conclusions

A fourth order mechanical system similar to Mizrahi and Susak (1982b) was presented, and simulations of tibial axial accelerations of subjects following drop impacts were made. Stiffness and viscosity coefficients were varied until the optimal match between the measured and the estimated accelerations was obtained. The variability of coefficients and the corresponding peak tibial accelerations and maximum rates of tibial acceleration was discussed. The sensitivity of the model outputs was assessed using subject mean and group mean coefficients for all trials.

The reported drop landing method resulted in physiologically reasonable peak tibial accelerations and ground reaction forces, but may contribute to the variability in estimated visco-elastic coefficients because drop height and leg orientation were not monitored. The low order mechanical model used in this study sufficiently reproduced the initial peak accelerations and maximum rates of acceleration measured following subject impact; however, a simple generic model of the heel pad was needed to account for the temporal characteristics of the subject measured tibial accelerations following impact. Estimated model coefficients for a subject were generally trial dependent because of their associated high variability. Therefore, coefficients determined for a given trial are not necessarily representative of a given subject or group of subjects. Additional model elements seem to be needed in order to reduce coefficient variability for a given subject.

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