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The Influences of Impact Interface, Muscle Activity, and Knee Angle on Impact Forces and Tibial and Femoral Accelerations Occurring After External Impacts

Wolfgang Potthast, Gert-Peter Brüggemann, Arne Lundberg, and Anton Arndt

The purpose of this study was to quantify relative contributions of impact interface, muscle activity, and knee angle to the magnitudes of tibial and femoral accelerations occurring after external impacts. Impacts were initiated with a pneumatically driven impacter under the heels of four volunteers. Impact forces were quantified with a force sensor. Segmental accelerations were measured with bone mounted accelerometers. Experimental interventions were hard and soft shock interfaces, different knee angles (0°, 20°, 40° knee flexion), and muscular preactivation (0%, 30%, 60% of maximal voluntary contraction) of gastrocnemii, hamstrings, and quadriceps. Greater knee flexion led to lower impact forces and higher tibial accelerations. Increased muscular activation led to higher forces and lower tibial accelerations. The softer of the two shock interfaces under study reduced both parameters. The effects on accelerations and forces through the activation and knee angle changes were greater than the effect of interface variations. The hardness of the two shock interfaces explained less than 10% of the variance of accelerations and impact forces, whereas knee angle changes explained 25–29%, and preactivation changes explained 35–48% of the variances. It can be concluded that muscle force and knee joint angle have greater effects in comparison with interface hardness on the severity of shocks on the lower leg.

Keywords: shock reduction, lower extremity mechanics, impact forces

Several authors have described impact forces and the related transient shock waves as one key factor in the development of typical injuries occurring in running and walking (Clement et al., 1981; Cook et al., 1990; James et al., 1978; Johnson, 1990; Milgrom et al., 1985; Milgrom et al., 1992; Schwellnus et al., 1990; Simkin et al., 1989; Tooms et al., 1987; van Mechelen, 1992; Whittle, 1999; Wosk & Voloshin, 1981). Others could not find a positive relationship between the magnitude of impact forces and injury rates (Bahlsen, 1989; Nigg et al., 1995) and some studies have even shown adaptation effects of biological material as a response to impact loading in running (Kersting, 1997; Kersting & Brüggemann, 1999; Kersting et al., 2006; Fredericson et al., 2007; Middleton et al., 2008).

Possible mechanisms to modulate the magnitude of the impact force and the shock wave along the skeleton are variations in the hardness of the shock interface (Lafortune et al., 1996a), changes in joint angles (Lafortune et al., 1996a; McMahon et al., 1987; Ratcliffe & Holt, 1997) and changes in muscle force (Bobbert et al., 1992; Denoth 1986; Gerritsen et al., 1995). However, shoes with softer midsoles worn during running did not systematically reduce the impact load when compared with harder midsoles (Clarke et al. 1985; Nigg & Liu, 1999; Nigg et al., 1987; Kersting & Brüggemann, 1999). Differences in running style—for example, variations in initial knee angle—have been found to modify impact forces and the related shock wave (Bobbert et al., 1991; McMahon et al., 1987). Shorten (2002) and Shorten and Mientjes (2003) attributed the high-frequency component of the impact force curve (frequencies > 12 Hz) to the collision of the foot and lower leg with the ground. The magnitude of this high-frequency component occurred later and was lower when soft midsoles were used. The low-frequency component (frequencies ≤ 8 Hz) was attributed to the leg spring and was not influenced by midsole hardness.

The stiffness of the leg spring depends partially on the knee joint angle (Denoth, 1986; Bobbert et al., 1991) and the force generated by the muscles crossing the knee (Denoth, 1986; Bobbert et al., 1992). It can therefore be assumed that not only the isolated knee angle but also the muscle force, or the muscle activity for a given joint angle, have an influence on the shock transmission through the knee joint (Bobbert et al., 1992; Ratcliffe & Holt, 1997; Lafortune et al., 1994, 1996a, 1996b). Due to their interdependence it is, however, difficult to analyze the role of musculature and initial knee angle in

Potthast and Brüggemann are with the Institute for Biomechanics and Orthopaedics, German Sport University, Cologne, Germany. Lundberg is with the Karolinska Institute, Karolinska University Hospital, Huddinge, Sweden. Arndt is with the Karolinska Institute, Karolinska University Hospital, Huddinge, Sweden, and the Swedish School of Sport and Health Sciences, Stockholm, Sweden.

locomotion individually (Bobbert et al., 1992; Lafortune et al., 1996a, 1996b; Lafortune & Lake, 1995).

The contributions of impact interface, knee joint angle, and muscle activity on the magnitude of impacts and segmental shocks have not been quantified independently. The purpose of this study was to quantify the relative contributions of impact interface, muscle activity, and knee angle to the magnitudes of tibial and femoral accelerations occurring after external impacts.

Methods

During human locomotion it is not possible to independently quantify the influences of knee angle, muscle activity, and hardness of shock interface on impact magnitude. Therefore the experimental design was restricted to controlled external impacts under the heel. The experimental interventions consisted of controlled changes in knee angle, muscle activation conditions, and impact interface during the induced impacts. Three knee angle conditions were chosen: 0°, 20°, and 40°. In each knee angle position, three isometric muscle contraction conditions were defined: 0%, 30%, and 60% of previously performed maximal voluntary contractions (MVCs). Two different cushioning materials at the plantar impact interface were compared in all contraction conditions at the 40° knee angle.

Five healthy male volunteers agreed to participate as subjects in this study. The data of one subject did not satisfy the inclusion criteria (see below). Therefore, the data of four subjects (age: 35, 36, 37, and 47 years; height: 178, 189, 183, and 173 cm; weight: 85, 88, 76, and 70 kg respectively for subjects 1–4; running distance: 15–60 km/week) were included in the study. The subjects were informed of the aims and risks involved in the study and they provided written consent to participate. The study was approved by the local ethical committee of Huddinge University Hospital, in Sweden.

Two triaxial lightweight piezoelectric accelerometers (piezotron, 8694M1, Kistler Instrumente AG, Winterthur, Switzerland, mass < 0.0025 kg) were used to determine the tibial and femoral axial accelerations. Data from the mediolateral and anterioposterior axes were not analyzed in this study. To avoid artifacts due to soft tissue movements, the accelerometers were mounted on the protruding ends of intracortical pins (3 mm Apex, Stryker, Geneva, Switzerland). These were inserted under local anesthetic in the right medial tibial condyle about 7 cm below the knee joint space and the medial femoral condyle about 2.5 cm above the knee joint space and distal to the insertion of the vastus medialis muscle. The proximal part of the tibia was chosen for insertion to facilitate the accurate determination of shock transmission over the knee in another study.

Since exact alignment of the accelerometer axis to the relevant segment's axis was not possible, a reference measurement was conducted to mathematically adjust the orientation of the accelerometer data to the segment axis. The tibial axial axis was defined as a line from the midpoint between the two malleoli to the midpoint between the medial and lateral aspects of the knee joint space. The femoral axis was defined as a line from the midpoint between medial and lateral aspects of the knee joint space to a point 5 cm medial to the greater trochanter. Five external reflective markers (12 mm) were fixed at the relevant landmarks for the reference measurement. Two reflective markers (5 mm) were attached to straight plastic tubing positioned on the accelerometer axis most closely oriented along the segment axis. Five infrared cameras (Proreflex system, Qualisys, Gothenburg, Sweden) recorded the coordinates of the segment and sensor markers in the laboratory system for 3 s at a sampling frequency of 240 Hz. Knowing the orientation of the accelerometers and leg segments in the laboratory systems permitted calculation of the sensor axis relative to the segment axis and therefore determination of the resulting accelerations along the respective segment. Furthermore, the acceleration component resulting from gravity could be subtracted (Lafortune & Hennig, 1990).

Experiment

After pin insertion, the subjects were positioned supine while the upper body and the pelvis were firmly strapped into a custom-built adjustable seat. The shank and the thigh were not strapped to the seat. By tilting the seat, the setup allowed for knee angle positions of 0°, 20°, and 40°, while maintaining a constant hip angle of 40° in each knee angle condition.

An electronic custom-built goniometer (Förster, German Sport University, Cologne, Germany) was attached laterally to measure the knee angle. The goniometer axis was aligned with the flexion-extension axis of the knee. The hip angle was controlled manually with an optical goniometer and the lower leg was oriented horizontally, which was controlled with a spirit level.

The foot was fixed to a custom-built pneumatically driven impacter device (Küsel, German Sport University, Cologne, Germany). Impacts were applied under the heel with the circular contact surface (radius 5 cm) of the impact cylinder covered with a cushioning material. The direction of the impact was aligned with the longitudinal axis of the tibia. To determine the influence of materials with different elastic moduli at the impact interface, two 1-cm-thick ethylene vinyl acetate (EVA) materials were chosen. The harder material had an elastic modulus of 1.5 MPa (Asker C, shore 60) and the softer material (soft) 0.9 MPa (Asker C, shore 30). The subject's forefoot was fastened with a strap to the forefoot sole of a running sandal (Plainsrunner, Nike Inc., Portland, USA). The plantar surface of the rearfoot was freely accessible for the impacter so that shocks could be initiated under the heel. The distance between the contact surface of the impacter and the plantar surface of the heel was 1.5 cm (controlled visually with a scale) before the impact was initiated. A piezoelectric force sensor (type 902A, Kistler Instruments AG, Winterthur, Switzerland) integrated into the impact cylinder allowed quantification of the applied forces (sampling rate: 1000 Hz). The impact phase was defined by the force-time-history. The start of the impact phase, and thereby the beginning of the time window of the acceleration analysis, was set at the instance (t_{start}) when the applied force exceeded 15 N. The impact phase ended (t_{end}) at the time of the first local minimum after the impact maximum (Figure 1).

After one trial for familiarization, in each experimental knee angle position one MVC was recorded for all analyzed muscles. The MVC signals were used as a reference for the muscle activity during the impact trials as well as a trigger to initiate the impacts (see below). The maximum of the rectified and smoothed (100 points moving median) MVC signals for each muscle at each knee angle condition was calculated to obtain reference values for assessing the muscle activity during the impact trials.

Activity of muscles crossing the knee was recorded with bipolar electromyography (EMG) lead-offs with preamplification (analog band pass filter 10–500 Hz, Biovision, Werheim, Germany) and self-adhesive surface electrodes (blue sensor, Medicotest, Ballerup, Denmark). EMG data (sampling frequency 1000 Hz) was collected from the following muscles: vastus medialis (VM), vastus lateralis (VL), gastrocnemius medialis (GM), gastrocnemius lateralis (GL) (Figure 2), and semitendinosus (ST). The EMG signals of GM and VM were used as input for the impact initiation. This input produced a software-initiated impulse, which opened the valve of the impacter device that led to the cylinder actuation. The software impulse was initiated when the activities of both reference muscles were simultaneously at 0%, 30%, or 60% of the MVC value, depending upon the particular activation level required for the relevant experimental condition. Deviations of ±10% were tolerated in both muscles. An initiation condition was fulfilled when the mean value of the EMG rectified signal over 100 ms was within the described tolerance interval.

During the trials, the subjects had to contract the analyzed muscles either 0%, 30%, or 60% of MVC while observing the activity level of the two reference muscles online on a biofeedback system. The impact was only initiated if the activity levels of both muscles were simultaneously within those intervals. Subjects had one to three practice trials for familiarization with the procedure.

Ten impacts were initiated in each of the nine different combinations of knee angle conditions and muscle activity levels.

Signal Processing

The EMG signals of the five muscles under study (GM, GL, VM, VL, and ST) were analyzed to investigate the influence of muscle activity. The raw signals were rectified and smoothed (100 points moving median). To allow comparisons between subjects, these signals were scaled to the previously recorded MVC data. For each trial, the integrals over time (iEMG; the time interval from 50 ms before impact to the instant of impact) of these data sets were used to assess the muscle activity.

The maximal knee angles (α_{max}) and the maximal knee angle changes ($\Delta\alpha_{max}$) during the impact phase were calculated. The external force accelerating the leg segments was assessed using the force maximum, F_{max}.

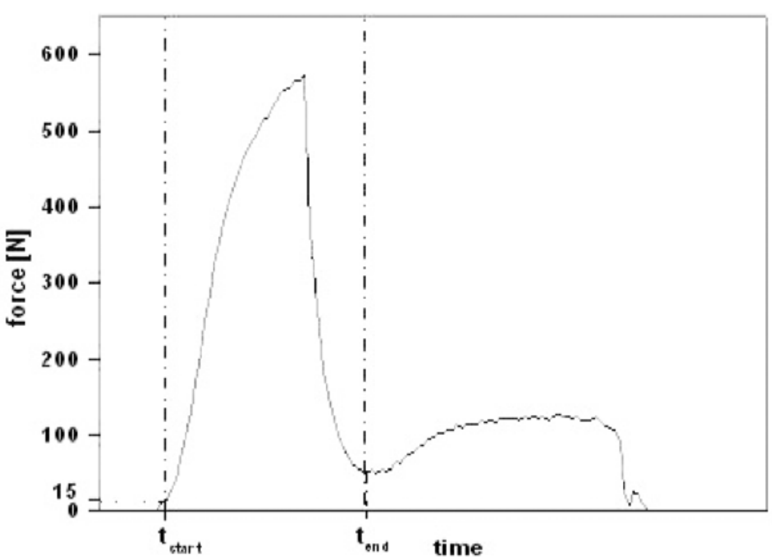

Figure 1 — Determination of the impact phase from t_{start} to t_{end}. The impact phase started when the reaction force exceeded 15 N and ended at the first local minimum after the force maximum.

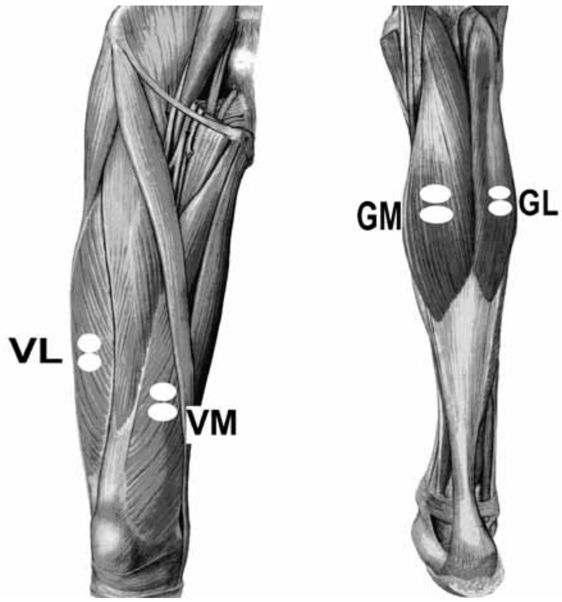

Figure 2 — Placement of the bipolar electrodes for electromyography. White ellipses indicate the location of the electrodes on the vastus lateralis (VL), vastus medialis (VM), gastrocnemius medialis (GM), and gastrocnemius lateralis (GL).

The natural frequency of the experimental instrumentation was above 150 Hz so that an adequate digital filter algorithm could be applied to separate the segment acceleration from the pin vibrations (Potthast et al. 2002). A 22nd-order Butterworth algorithm was used.

Inclusion Criteria

Three criteria were used to identify nonvalid trials. (1) Trials were excluded if the time between the initial forward movement of the impacter cylinder and the first contact with the heel (t_{start}) differed by more than two standard deviations from the mean value over all trials under the specific experimental condition. In those cases the distance between the heel and the impacter could not have been exactly 1.5 cm, which had only been visually controlled before the impact initiation. (2) A trial was also excluded from further analysis if the related F_{max} value differed by more than two standard deviations from the mean value over all trials under the specific experimental condition. In those cases the impact could not have been initiated centrally under the heel. (3) The final exclusion criterion was based upon an analysis of the oscillating frequency of the intracortical pin to identify possible loosening within the bone. The frequency spectra of the nonfiltered acceleration signals was determined using a fast Fourier transformation algorithm from $t = t_{start} - 511$ ms until $t = t_{start} + 512$ ms. A loosening of the pin in the bone would lead to a left shift of the local amplitude minimum in the frequency domain, which separates the natural frequency of the experimental instrumentation from the acceleration of the leg segments (Potthast et al., 2002). Trials were excluded if the local minimum was below 100 Hz.

Statistics

A multivariate ANOVA with post hoc tests (Student-Newman-Keuls) was used for the identification of significant differences between the variables knee angle, muscle contraction level, and impact interface. The level of significance was 5% ($p \leq .05$). The eta squared value obtained by the ANOVA was interpreted as the proportion of the variability in the dependent variables that is accounted for by the variation in the independent variables.

Results

Table 1 summarizes mean values and standard deviations of all experimental conditions.

The average peak force (F_{max}) for all subjects and testing conditions was 491 ± 129 N. Increasing knee angles at constant contraction levels led to F_{max} reductions between 9 N (from 0° to 20° knee angle at 60% contraction level) and 158 N (0° to 40° at 30%). The differences were all significant ($p < .05$) except for the comparison between 0° and 20° at 60% contraction level. Figure 3 shows the mean values for F_{max} for all conditions.

Increasing contraction levels led to increasing F_{max} values in all knee angle conditions. The only knee angle at which the force increase was not significant was the 0° position between 30% and 60% contraction level ($p = .83$). A significant reduction ($p < .05$) in F_{max} resulting from the softer shock interface was found only in the 30% contraction level (81 N). Figure 4 shows representative tibial and femoral acceleration time histories of one subject in one particular testing condition. Both segments were accelerated proximally by the impact. The average

Table 1 Mean values and standard deviations of peak forces (F_{max}), and peak tibial (ACC_{tib}) and peak femoral (ACC_{fem}) accelerations for the different experimental conditions. Significant differences are indicated in figures and text.

Knee Angle and Interface	F_{max} (N)			ACC_{tib} (g)			ACC_{fem} (g)		
	0%	30%	60%	0%	30%	60%	0%	30%	60%
0°; Hard	475 ± 64	619 ± 33	643 ± 147	2.7 ± 0.6	2.2 ± 0.3	2.0 ± 0.4	2.0 ± 0.6	1.7 ± 0.3	1.5 ± 0.3
20°; Hard	414 ± 48	512 ± 69	633 ± 68	3.6 ± 1.2	3.1 ± 0.8	2.2 ± 0.6	2.0 ± 0.5	1.7 ± 0.5	1.3 ± 0.4
40°; Hard	347 ± 29	461 ± 38	554 ± 96	4.0 ± 0.7	3.6 ± 0.8	3.2 ± 0.3	1.1 ± 0.6	1.4 ± 0.9	1.7 ± 1.4
40°; Soft	334 ± 37	380 ± 77	520 ± 121	4.0 ± 1.0	2.8 ± 0.6	2.1 ± 0.6	0.8 ± 0.5	1.0 ± 0.7	1.1 ± 0.8

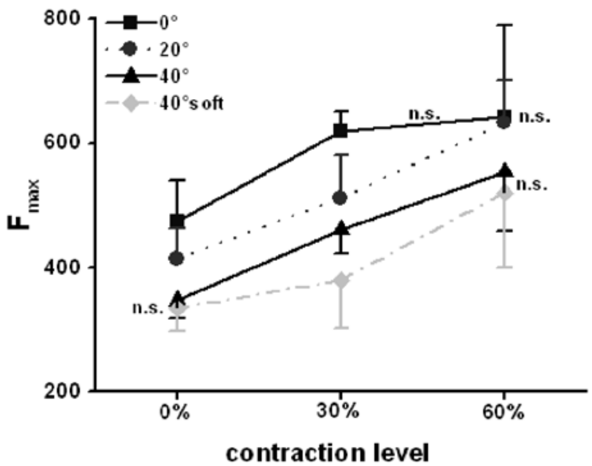

Figure 3 — Mean peak reaction forces (F_{max}) of all subjects for all testing conditions. Nonsignificant differences ($p \geq .05$) are indicated by *n.s.* Statistical comparisons with the soft condition were carried out only within the 40° angle position.

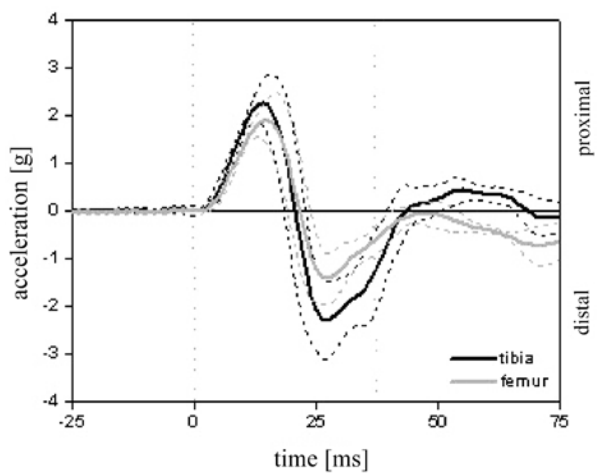

Figure 4 — Representative acceleration time histories for the tibial and femoral acceleration (ACC_{tib}, ACC_{fem}). Mean and standard deviation curves for one subject at 20° knee angle at the 30% contraction level.

maximal acceleration was 2.9 ± 1.0 g at the tibia (ACC_{tib}) and 1.4 ± 0.8 g at the femur (ACC_{fem}).

Figure 5 shows the mean accelerations for the tibia (ACC_{tib}) and for the femur (ACC_{fem}) for all experimental conditions. Increasing knee angles led to higher accelerations at the tibia (significant except for 0% between 20° and 40° and 60% between 0° and 20°), which was not the case for the femur. At 0% contraction, ACC_{fem} was approximately 1 g lower at 40° than at 0° and 20°. No significant differences were found between knee angle conditions under 30% and 60% contraction levels. Tibial acceleration decreased with increasing contraction levels (significant except for at 0° between 30% and 60%), which was also seen at the femur under the 0° and 20° knee angle conditions (significant except for at 0° between 30% and 60%). No significant effect of muscular contraction on ACC_{fem} could be identified under the 40° knee angle. The softer interface did not lead to significant acceleration changes under the 0% contraction condition for either segment. In the 30% and 60% contraction levels, the acceleration peaks were reduced significantly at the tibia (0.7 and 0.8 g) and at the femur (0.4 and 0.6 g) using the softer interface.

The multivariate ANOVA revealed the contributions of the different independent variables on the variance of the force and segmental accelerations (Figure 6). Approximately 77% of the variance of F_{max} could be explained by the experimental variations of the knee angle, contraction level, and shock interface. For ACC_{tib}, about 73% of the variance could be explained and for ACC_{fem}, only 9%. The experimental variation of the muscular contraction had the biggest influence on the variance of F_{max} (48%) and ACC_{tib} (35%). The knee angle changes explained 25% (F_{max}) and 29% (ACC_{tib}), whereas the interface intervention explained only 4% and 9% respectively. At the femur the interface explained about 5% of the variance of the acceleration, the knee angle about 3%, and the contraction level only about 1%.

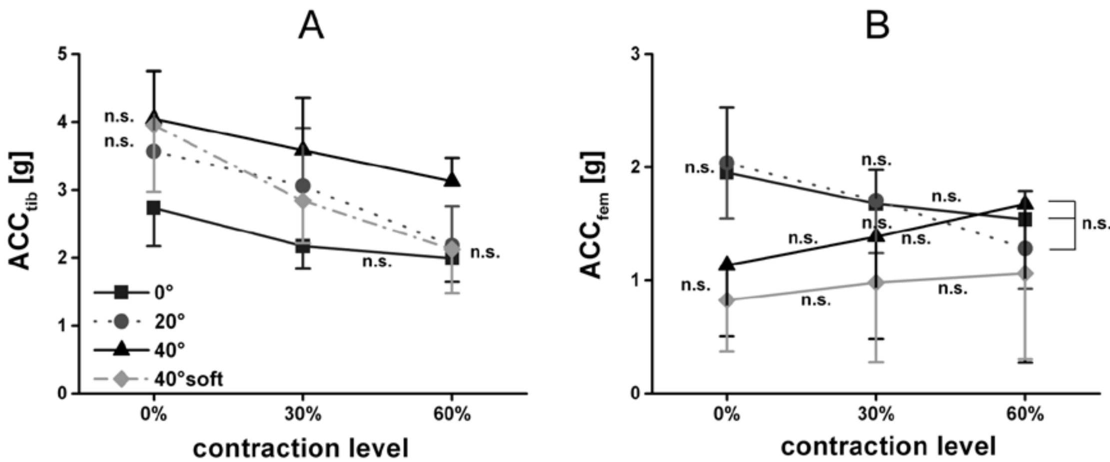

Figure 5 — Mean peak accelerations at the tibia (A) and femur (B) of all subjects for all testing conditions. Nonsignificant differences ($p \geq .05$) are indicated by *n.s.* Statistical comparisons with the soft condition were carried out only within the 40° angle position. Note different scaling of the vertical axes.

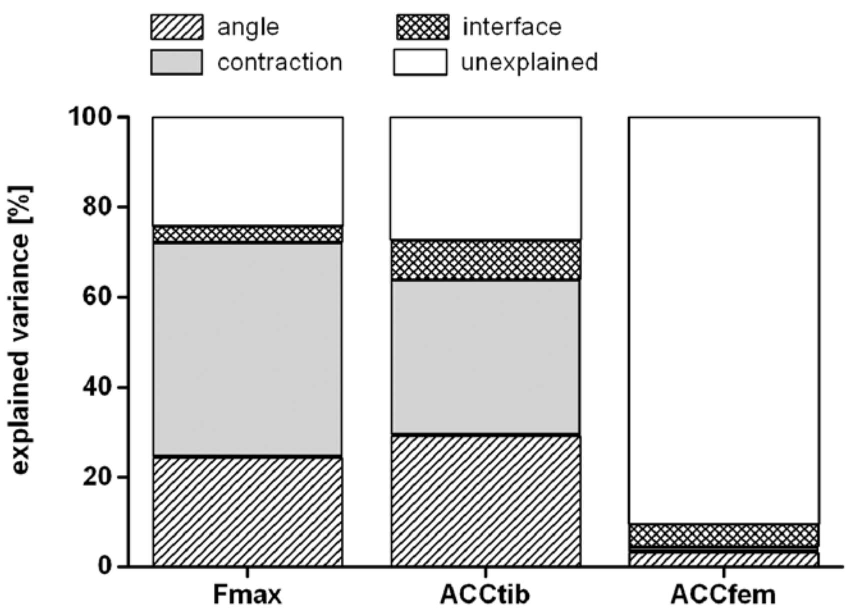

Figure 6 — Contributions of the independent variables on the variance of F_{max}, ACC_{tib}, and ACC_{fem} as determined by eta squared.

Discussion

The purpose of this study was to determine the relative contributions of the collision interface, muscle activity, and knee angle on tibial and femoral accelerations and impact forces after external impacts. Impacts were initiated under the heels of four volunteers and impact forces as well as skeletal accelerations of the tibia and femur were measured. Due to the invasiveness of the study, subject numbers were limited, which must be taken into consideration when interpreting the results.

The study shows that the three analyzed mechanisms (knee flexion angle, contraction of muscles surrounding the knee, impact interface) can modulate the magnitude of the impact under controlled conditions independently from each other. Higher muscular contraction as well as a more extended knee increase the stiffness of the limb and therefore generate a stiffer abutment for the impacter. This consequently leads to higher impact forces (see Figure 3). If the knee angle and the muscle contraction stay constant, a softer interface can also reduce the impact force. However, as Figure 3 suggests, that the interface has less effect on the force than muscular contraction and knee joint angle. The effect of decreasing impact force with higher knee angles is in accordance with results reported by Lafortune et al. (1996a). However, they found a bigger effect on reaction force through interface variations than through changes in knee angle. The lack of control of muscle activity in their human pendulum experiments might contribute to the explanation of the differences to the current study. Also, the studies of Shorten (2002) and Shorten and Mientjes (2003)—as well as other running-related studies, which indicate no systematic effect of midsole hardness on impact magnitude (Clarke et al. 1985; Nigg & Liu, 1999; Nigg et al. 1987; Kersting & Brüggemann 1999)—do not contradict the results of the current study. Due to their possible interdependence it is impossible to measure the influence of midsole hardness, knee joint angle, and muscle activity independently from each other in running directly in an experimental study (see Bobbert et al., 1991). Effects caused by one parameter (e.g., midsole hardness) could be balanced or blurred by an inverse effect of one or two other parameters (knee angle and/or muscle activity). The controlled experimental interventions of knee angle, muscle activity, and shock interface used in this study enable identification of the effect of those variables independently.

Peak forces (429 ± 129 N) as well as tibial (2.9 ± 1.0 *g*) and femoral peak accelerations (1.6 ± 0.7 *g*) found in this study mirror the magnitudes of peak forces (Lafortune & Hennig, 1992; Light et al., 1980), peak accelerations at tibia (Lafortune & Hennig, 1992; Light et al., 1980), and femur (Wosk & Voloshin, 1981) occurring at the heel strike in walking. Future studies with higher impact magnitudes might be necessary to explore the identified effects more elaborately.

The accelerations at the tibia decreased due to the stronger abutment caused by the stiffer coupling of tibia and femur. A given impact at the distal end of the lower leg leads to higher acceleration if the coupling at the proximal end is weak or, in other words, if the abutment is soft and vice versa. In the extended knee positions and at higher muscle contractions, the abutment is stiffer compared with the more flexed knee positions and low contractions. In general, the tibia acceleration decreased with a more extended knee, higher muscle forces, and a softer interface. Similar effects on tibia acceleration through joint angle interventions in running have been observed previously (McMahon et al., 1987). Except for the muscle contraction effect, the results of the current study are in accordance with those reported by Lafortune et al. (1996a). The present study indicates that the softer interface does not have a dominant effect on the tibial peak acceleration under the low- and medium-contraction conditions. The interface effect appears to be as big as the knee angle effect only under the highest activation level (see Figure 5 A).

The accelerations at the femur were about 45% lower than at the tibia. Unlike the tibia accelerations, the femur accelerations showed less systematic responses to the experimental interventions (see Figure 5 B). Apparently the influences of the experimental interventions were weakened at the knee level. This deduction is supported by the results of the ANOVA (see Figure 6): Whereas 77% of the variance of F_{max} and 73% of the variance of ACC_{tib} could be explained by the experimental interventions, only 9% of the variance of ACC_{fem} have been explained whereas 91% stayed unexplained at that segment. Bobbert et al. (1992) showed that the high-frequency component of the vertical ground reaction force is associated with the vertical acceleration of the lower leg and foot whereas it corresponds less with the acceleration of the upper leg. Therefore, it is no contradiction that variations of the independent variables influence F_{max} and ACC_{tib} but not ACC_{fem} systematically.

The contribution of the shock interface on the variation of the variables F_{max} and ACC_{tib} respectively is below 10%. Since knee joint angle and muscle force explain 64–73% of the variation of F_{max} and ACC_{tib}, they play a dominant role in shock modulation compared with the hardness of the interface. Especially the role of the muscle force, explaining 48% of the variance of F_{max} and 35% of ACC_{tib}, should be of major interest. It is widely established that the high-frequency contents of stress and strain determine the tissue response in biological structures (Mosley, 2000). The present study suggests that especially the forces applied by the muscles crossing the knee have a bigger influence than the hardness of the shock interface. That means that changes in muscle activity as chosen in the current study affect tissue responses due to high-frequency loading to a greater extent than the shock interface. The increments in muscle activity of 30% from maximal voluntary contractions are within the range of changes in preactivation, measured with surface EMG in different impact-related sport activities (Morey-Klapsing et al., 2007; Moritz & Farley, 2004; Besier et al., 2003). In addition, Herzog and Longino (2007) underlined the possible role of weak muscles in the development of joint degeneration.

In summary, the current study shows that muscular preactivation and knee angle have a more dominant effect over the shock on the lower leg than the hardness of the shock interface. This suggests that the development of adaptation effects or injuries depend more on joint coupling parameters, such as muscle force and joint angle, than on the shock interface, as long as those parameters are not interdependent. Therefore, interfaces between foot and ground (e.g., footwear and insoles) should not disturb or hinder muscle function.

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