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ORIGINAL RESEARCH

Musculoskeletal Attenuation of Impact Shock in Response to Knee Angle Manipulation

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Shock waves resulting from the foot-ground impact are attenuated by biological tissues within the body. It has been suggested that the primary site for shock attenuation is the knee joint. The purpose of this study was to determine if knee flexion affects the filtering characteristics of the musculoskeletal system in response to impacts. Impacts were delivered to 10 participants during inline skating on a treadmill at 2.0 m/s. Four knee angle conditions (0, 10, 20, and 30 degrees) were investigated using real-time visual feedback of motion capture data. Shock attenuation between the leg and head was determined using accelerometry. The cutoff frequency of the body was determined by progressive filtering of the leg acceleration until differences between head acceleration and filtered leg acceleration were minimized. A nonlinear increase in shock attenuation ($p < .001$) and a nonlinear decrease in the cutoff frequency of the body ($p < .001$) were observed as the knee became more flexed. These results suggest that the knee joint acts as a low-pass filter allowing greater shock attenuation with increased knee flexion. Flexing the knee may shift the shock-attenuating responsibilities away from passive biological tissue toward active muscular contraction.

Keywords: inline skating, lower extremity, injury prevention, accelerometry, effective mass

The foot–ground contact that occurs during locomotion and landing results in an impact force that propagates a shock wave through the musculoskeletal system (Shorten & Winslow, 1992; Hamill et al., 1995). The severity of the shock wave can be characterized by the peak magnitude and frequency of the acceleration resulting from impact. Before reaching the head, the magnitude and frequency of the shock wave is decreased passively by the heel fatpad, skin, bone, ligament, and tendon (Paul et al., 1978), and actively by eccentric muscular contraction (Derrick et al., 1998; Winter, 1983). The response of biological tissue to impact shock can be either positive or negative. Intermittent bouts of high impact loading will improve bone integrity through functional adaptation (Turner & Robling, 2003). Conversely, repetitive impact loading has been implicated in the development of overuse injury and degenerative disease (Voloshin & Wosk, 1981, 1982).

Lower-extremity geometry at contact plays an important role in shock attenuation (Derrick, 2004). When landing in a more extended position, the line of action of the impact force vector tends to pass through the lower-extremity joint centers. In this scenario, the shock wave travels a direct route through the skeleton; a minimal amount of active attenuation occurs and increased loads are placed upon passive tissue. In contrast, landing in a more flexed position tends to orient the impact force vector away from the joint centers and thus eccentric muscular contraction absorbs a significant portion of the shock wave. This also decouples the lower-extremity segments so that accelerations of the more distal segments are greater than accelerations of the proximal segments and trunk.

Due to their relative size and strength, the quadriceps muscle that crosses the knee joint has the highest potential to assist in the attenuation of shock. Previous studies have observed a positive relationship between knee flexion angle at contact and shock attenuation during running (Derrick et al., 2002; McMahon et al., 1987; Derrick, 2004). Unfortunately, voluntary changes in knee contact angle may affect distal and proximal joint kinematics, muscle activation patterns, and impact severity. Therefore, it is difficult to assess the individual contribution of knee contact angle to shock attenuation during running.

To the best of our knowledge, only one study has investigated the effects of a systematic change in knee contact angle on shock attenuation. Lafortune et al. (1996) examined shock transmission in response to knee angle manipulation using a human pendulum device in which participants laid supine on a suspended canvas bed. In both the time and frequency domain, an increase in knee flexion at impact corresponded to an increase in shock

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attenuation. The drawbacks of this experimental design are that participants are not weight bearing and do not require visual-vestibular feedback to maintain stability. Visual-vestibular stabilization has been implicated as a key control variable in response to impact shock (Pozzo et al., 1989, 1991), and the additional muscle activity, as well as intrajoint and spinal intradiscal pressures, associated with bearing weight could have an effect on shock attenuation (Magnusson et al., 1993). The gliding phase of inline skating provides a controlled experimental platform, in which knee angle can be systematically manipulated during impact and participants are required to bear weight and maintain postural stability in a similar fashion to normal human locomotion.

The observed reduction in the magnitude and frequency of acceleration between the leg and head during impact activity has lead researchers to speculate that biological structures of the musculoskeletal system behave as a low-pass filter (Hamill et al., 1995; Mercer et al., 2003; Shorten & Winslow, 1992). In this scenario, the filtering coefficients and therefore the cut-off frequency would depend on the dominant structures attenuating the shock. Stiffer, passive structures such as bone may transmit higher frequencies than active, more compliant tissue (i.e., muscle). A more flexed knee during impact would be associated with increased stretching of the elastic components in the muscle–tendon complex and increased muscular energy absorption (Derrick, 2004). Therefore, the amount of knee flexion during impact might alter the cutoff frequency of the body.

The purpose of this study was to determine if knee flexion affects the filtering characteristics of the body in response to impacts. It was hypothesized that increased knee flexion would decrease the cutoff frequency resulting in increased shock attenuation. A better understanding of how the human body counters the potential deleterious effects of shock transmission will aid in the prevention of musculoskeletal injury caused by impacts. This same information may also prove useful for developing strategies to maximize the osteogenic response associated with impacts.

Methods

Participants

Four females and six males were recruited for this study (age 25.6 ± 7.6 years, height 171.1 ± 9.4 cm, mass 69.2 ± 9.8 kg). Participants had been injury free for at least three months before the study. Written informed consent was obtained from each participant and the study was approved by the Institutional Review Board. All participants had prior experience with inline skating.

Data Collection

Upon visiting the laboratory, participants were outfitted with commercially available inline skates. Retroreflective markers were placed on anatomical landmarks of the trunk and right lower extremity. Markers were adhered to the toe, heel, and lateral malleolus on the exterior of the inline skate, and to the anterior calf, lateral femoral epicondyle, anterior thigh, left and right greater trochanter, and the left and right acromion process of the participant. A uniaxial piezoelectric accelerometer (PCB Piezoelectronics, Model353B, Depew, NY) was mounted to the lateral aspect of the inline skate just above the malleolus marker. A second identical accelerometer was mounted to the forehead (see Hamill et al., 1995). Both accelerometers were encased in Petro wax and fixed to a hard plastic backing before being adhered with double-sided tape. Athletic tape was used to secure the accelerometer to the skate, and an elastic band tightened to the participants comfort was used to secure the accelerometer to the head (Figure 1).

Participants were asked to stand on a treadmill as the belt was increased to an arbitrarily chosen speed of 2.0 $m \cdot s^{-1}$. Participants were given adequate time to get used to the treadmill belt moving underneath their skates. Once comfortable with the treadmill the protocol was initiated. A static motion capture trial was collected to determine sagittal joint angles when the participant stood in the anatomical position. All motion capture data

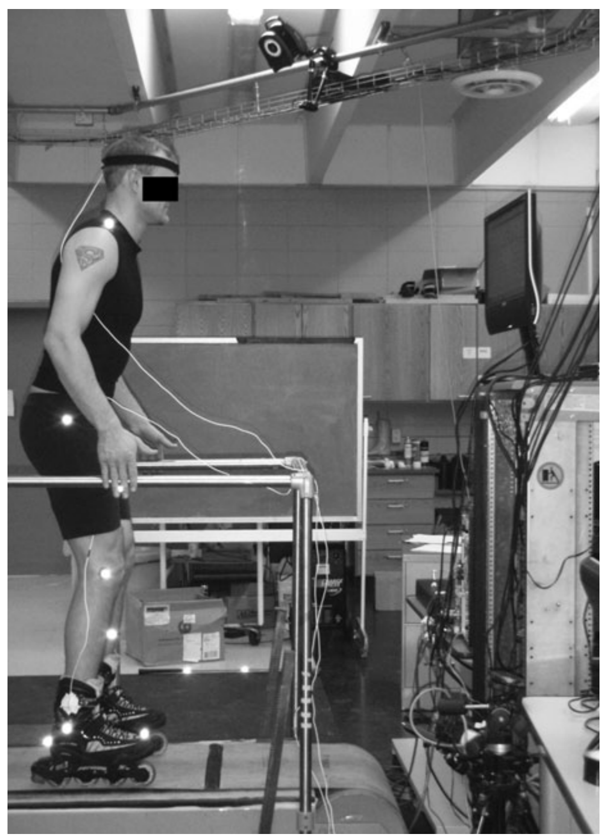

Figure 1 — Experimental setup displaying the location of retroreflective markers, leg and head mounted accelerometers, small bump (black strip on belt), and real-time feedback monitor.

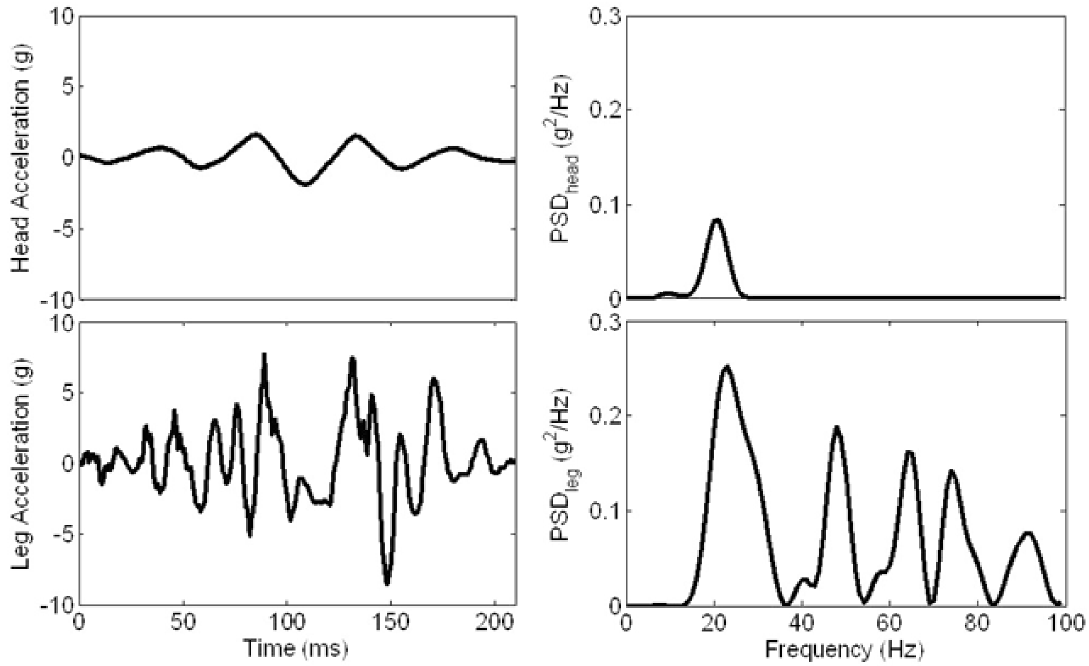

Figure 2 — Representative trial displaying leg and head accelerations in both the time and frequency domain.

were collected with eight optical cameras (Vicon MX, Vicon, Centennial, CO) placed in a circular configuration around the treadmill. A small bump approximately 3 mm in height and spanning the width of the treadmill belt was secured to the treadmill surface with duct tape (Figure 1). The treadmill was started, and impacts were delivered to the participant each time the treadmill belt completed one full rotation. Unlike gait, which illustrates a single impact associated with foot strike, there was a series of impacts delivered to the musculoskeletal system as the inline skate passed over the bump (Figure 2). Head accelerations consistently illustrated four main impacts associated with each time a wheel contacted the bump. Leg accelerations were less consistent, illustrating the four main impacts, superimposed with secondary impacts associated with the wheels recontacting the treadmill surface, and vibratory accelerations most likely associated with resonance of the rigid skate following each impact.

Four sagittal knee angle conditions were investigated, corresponding to 0, 10, 20, and 30° of knee flexion. Here, 0° refers to the anatomical position. The order of knee angle condition was counter-balanced across participants. Knee angle manipulation was accomplished through real-time visual feedback of motion capture data using Vicon Nexus software (Vicon, Centennial, CO). A red line, indicative of the current sagittal plane knee angle was displayed on a monitor in front of the participant (Figure 1). Participants were asked to maintain the desired knee angle throughout the entire condition. Each condition consisted of 10 delivered impact series. Accelerometer data were passed through the Vicon MX Control A/D board and collected concurrently with kinematic data using Vicon Nexus software. Kinematic and accelerometer data were collected at sampling frequencies of 160 and 1600 Hz, respectively. Hand railings were provided for safety and to help counter the resulting inline skate–treadmill frictional forces. Participants were asked to only put medially oriented loading on the railings when necessary, so as not to affect the transmission of the vertically oriented impact.

Data Processing

The raw motion capture and accelerometer data were exported to MatLab 7.0.4 (Mathworks, Natick, MA) for processing. Vectors defining the foot, shank, thigh, and trunk were used to calculate sagittal ankle, knee, and hip angle to verify knee angle manipulations and investigate its influence on the ankle and hip. All ankle, knee, and hip angles were referenced to the static motion capture trial to make angles 0° in the anatomical position.

Peak acceleration and attenuation were analyzed in the frequency domain (see Hamill et al., 1995). Briefly, the impact series for leg and head accelerations were extracted from the acceleration profiles. Any mean and linear trends in the data were removed. Data were transformed to the frequency domain using a Fast-Fourier transform to determine the power spectral density at frequencies from 0 to the Nyquist frequency. Powers and frequencies were interpolated so that each frequency bin was equal to 1.0 Hz.

Preliminary visual inspection of the power spectral density curves for leg (PSD_{leg}) and head (PSD_{head}) accelerations illustrated major power between 15 and 35 Hz. This power represented the four main impacts occurring each time a wheel contacted the bump. PSD_{leg}

also illustrated higher frequency power associated with the secondary impacts and vibrations observed in the time domain (Figure 2). This higher frequency power was deemed irrelevant to the current study because it was not representative of the four main impacts and was not observed in PSD_{head}—indicating these frequencies were always attenuated before reaching the head regardless of knee angle condition. The integral of power for frequencies between 15 and 35 Hz was extracted for a measure of leg (intLeg) and head (intHead) impact magnitude. A transfer function describing the gain or attenuation between PSD_{leg} and PSD_{head} was calculated using the following formula:

$$\text{Transfer Function} = 10 \log_{10} (PSD_{head} / PSD_{leg})$$

where the transfer function is the gain (positive value) or attenuation (negative value) in decibels. The transfer function values for frequencies between 15 and 35 Hz were converted to a linear scale, averaged, and converted back to decibels for a measure of impact attenuation (ATT).

To determine the low-pass filtering characteristics of the body, acceleration data were analyzed in the time domain. A cross-correlation of the leg and head acceleration impact profiles was performed. The head acceleration profile was shifted in the time domain by the lag resulting in the strongest correlation. This lag indicated the time it took the shock wave to traverse the musculoskeletal system. Progressive low-pass filtering (4th-order zero-lag Butterworth) of the leg impact profile was then performed in 1-Hz increments until the difference in maximum and minimum peak head and leg acceleration were minimized. The cutoff frequency that minimized the difference between peak head and leg acceleration in the time domain was chosen as the low-pass filter cutoff frequency (FREQC) of the body (Figure 3).

Statistical Analysis

All statistical analyses were performed in SPSS v.17.0.1 (SPSS Inc., Chicago, IL) following screening for normality and outliers. Differences in ankle angle, knee angle, hip angle, intLeg, intHead, ATT, and FREQC were examined using a one-way repeated-measures ANOVA (four knee contact angle conditions) with least significant difference (LSD) post hoc comparisons. Polynomial regression was used to determine if linear or quadratic trends explained the relationships between dependent variables and knee angle condition. Where appropriate, the F-statistic was used to determine if the increase in r^2 from a linear to a quadratic trend was significant. The criterion alpha level was set to 0.05 for all statistical tests. Cohen's d effect sizes for ATT and FREQC were calculated between conditions to help guide the interpretation of meaningful findings (Cohen, 1992).

Results

Knee angle was successfully manipulated during the study ($p < .001$), resulting in an approximate 10° difference between each condition (Figure 4). On average, the experimentally measured knee angles exhibited 2.3 ± 1.5° more flexion than the target knee angle, but for simplicity, conditions will be referred to as conditions 0, 10, 20, and 30° for the remainder of this paper. Angles were reasonably maintained throughout each impact series, with the mean range of motion being less than 0.3°. Knee angle manipulation also influenced ankle ($p < .001$) and hip ($p = .002$) angles. Ankle flexion increased with changes in knee angle (Figure 4). Hip flexion for condition 30° was greater than conditions 0 through 20°, and hip flexion for condition 20° was greater than conditions 0 and 10° (Figure 4).

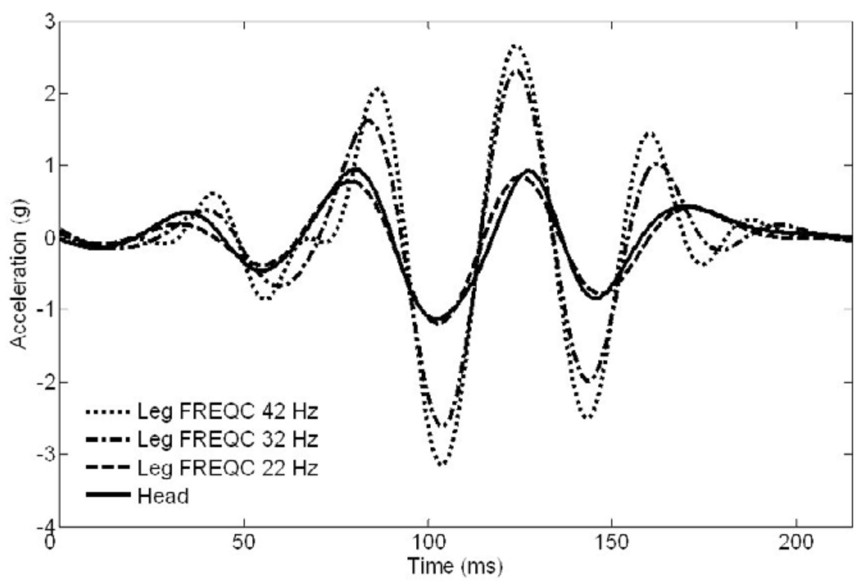

Figure 3 — Representation of progressive filtering procedure to obtain FREQC.

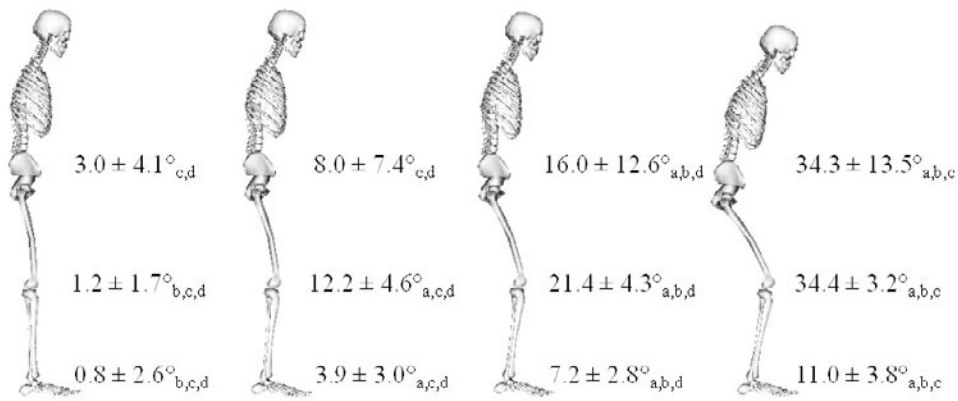

Figure 4 — Mean (*SD*) hip, knee, and ankle angles for each condition (0–30° left-right). Subscripts a, b, c, and d denote significant difference from condition 0, 10, 20, and 30°, respectively ($p < .05$).

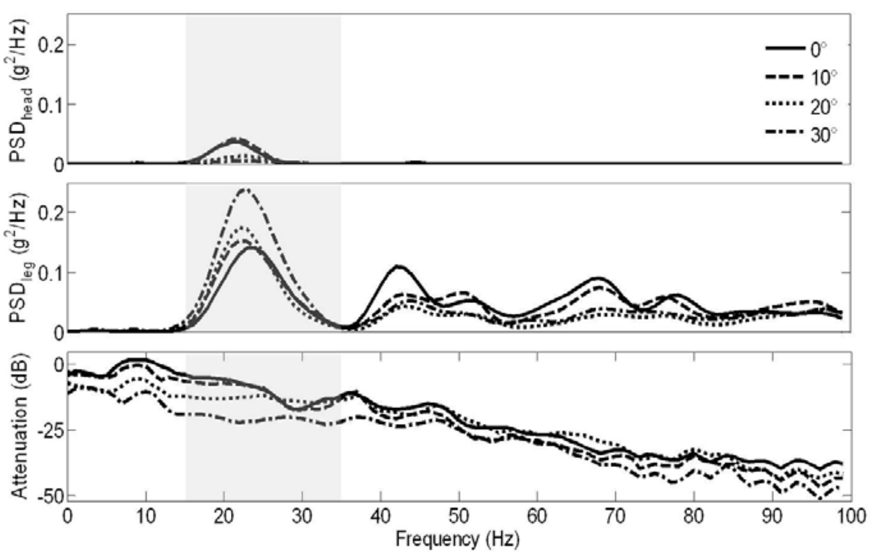

Figure 5 — Group ensemble power spectral densities and transfer functions for each knee angle condition. Gray shading illustrates the primary impact frequencies from 15 to 35 Hz.

Figure 5 illustrates the group ensemble power spectral densities and transfer functions for each knee angle condition. Knee angle manipulation influenced intLeg ($p = .025$) and intHead ($p = .007$). The intLeg for condition 30° was greater than conditions 0 through 20°, and intLeg during condition 20° was greater than condition 10° (Table 1). In addition, intHead during condition 30° was greater than conditions 0 through 20°, and intHead during condition 20° was greater than condition 0° (Table 1). Significant linear and quadratic relationships were observed between knee angle and intLeg (linear $r^2 = .33$, $p < .001$; quadratic $r^2 = .40$, $p < .001$), as well as knee angle and intHead (linear $r^2 = .28$, $p = .001$; quadratic $r^2 = .29$, $p = .003$). The additional variance explained by the quadratic relationship was significant for intLeg ($p = .022$), but not for intHead ($p = .674$) (Figure 6).

ATT tended to increase with knee flexion angle ($p < .001$). Condition 30° attenuated more impact than conditions 0 through 20° and condition 20° attenuated more impact than condition 0° (Table 1). The FREQC tended to decrease with knee angle ($p < .001$). Condition 30° had a lower FREQC than conditions 0 through 20°, and condition 20° had a lower FREQC than conditions 0 and 10° (Table 1). Significant linear and quadratic relationships were observed between knee angle and ATT (linear $r^2 = .50$, $p < .001$; quadratic $r^2 = .55$, $p < .001$), as well as knee angle and FREQC (linear $r^2 = .63$, $p < .001$; quadratic $r^2 = .66$, $p < .001$). The additional variance explained by quadratic relationships were significant (ATT, $p = .030$; FREQC, $p = .039$) (Figure 7). The effect size estimates presented in Table 2, suggest that the observed significant differences for ATT and FREQC are substantially meaningful.

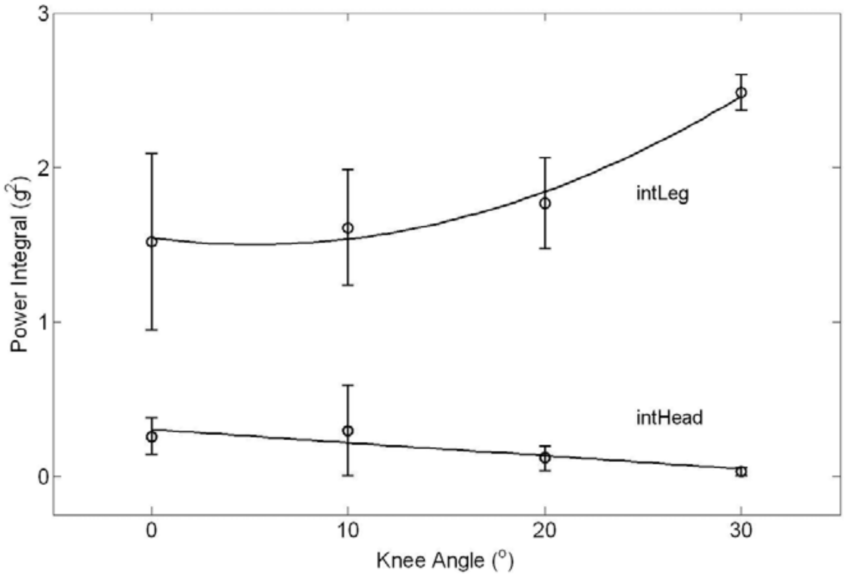

Figure 6 — A quadratic fit best explained variance in the integral of leg (intLeg), but not head (intHead) power with changes in knee angle.

Table 1 Mean intLeg, intHead, ATT, and FREQC for the four knee angle conditions

	Condition 0°	Condition 10°	Condition 20°	Condition 30°
intLeg (g^2)	1.516 ± 0.569	1.608 ± 0.373	1.767 ± 0.294[d]	2.485 ± 0.114[a]
intHead (g^2)	0.254 ± 0.119	0.293 ± 0.292	0.116 ± 0.078[c]	0.029 ± 0.025[a]
ATT (dB)	–7.8 ± 4.2	–8.2 ± 3.1	–12.7 ± 5.2[c]	–20.1 ± 3.2[a]
FREQC (Hz)	21.8 ± 2.5	20.1 ± 3.4	17.0 ± 3.0[b]	12.8 ± 2.4[a]

[a]Significantly different from conditions 0, 10, and 20°.
[b]Significantly different from conditions 0 and 10°.
[c]Significantly different from condition 0°.
[d]Significantly different from condition 10° ($p < .05$).

Discussion

The purpose of this study was to determine if knee angle manipulation affects the filtering characteristics of the body in response to impacts. We hypothesized that increasing knee angle would result in an increase in ATT and a corresponding decrease in FREQC. Our hypothesis was supported by the results of this study. Attenuation became progressively greater and FREQC became progressively lower between knee angles of 10–30°. No differences in ATT and FREQC were observed between knee angles of 0 and 10°, but this was most likely due to the nonlinear relationship observed between these variables and the amount of knee flexion.

If the resulting changes in ankle and hip angles between conditions were to bring about changes in the orientation of the accelerometers it is possible that the differences in attenuation would reflect changes in the sensitive axis of the accelerometers rather than "true" shock attenuation. It is unlikely that the changes in hip angle brought about substantial changes in the orientation of the head accelerometer as participants needed to keep their eyes fixed on the monitor in front of them to maintain the appropriate knee angle. However, due to changes in ankle angle, the orientation of the leg accelerometer went through a range of approximately 10° between knee angle conditions of 0 and 30°. We investigated the influence that a 10° rotation in the sensitive axis of the accelerometer had on acceleration using an Exeter impact testing machine (Exeter Research, Inc, Exeter, NH). A 10° rotation from vertical resulted in a change in impact acceleration from 11.1 to 10.9 g. Assuming the acceleration occurred at a single frequency, this difference would correspond to an attenuation of –0.16 dB. Our differences in attenuation between conditions 0 and 30° were large enough that the reduction in acceleration from a 10° rotation would be negligible, and if anything make our measures more conservative. We can thus conclude that our measurements provide a meaningful and accurate comparison of shock attenuation across knee angle conditions.

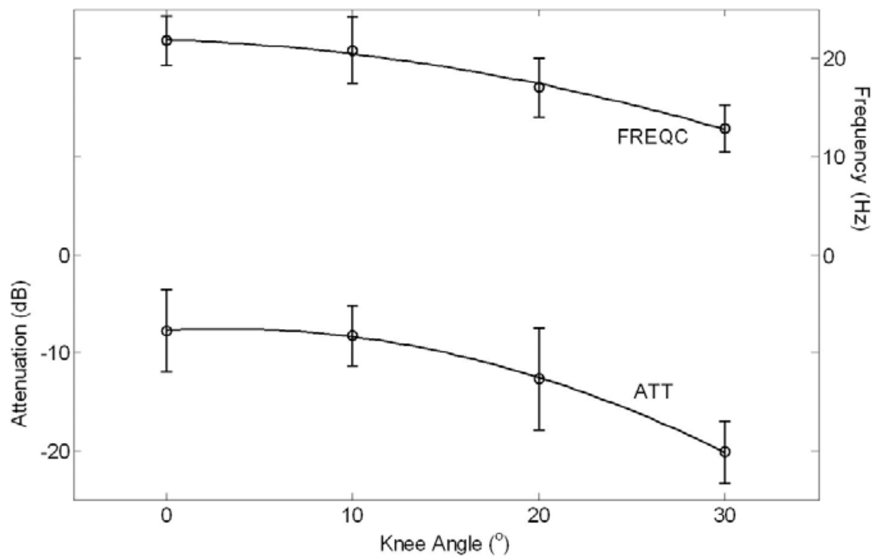

Figure 7 — A quadratic fit best explained variance in attenuation (ATT) and cut-off frequency (FREQC) with changes in knee angle.

Table 2 Cohen's *d* effect sizes between conditions for ATT and FREQC

Comparison	ATT	FREQC
Condition 0° – Condition 10°	0.1	0.6
Condition 0° – Condition 20°	1.2	1.6
Condition 0° – Condition 30°	3.1	3.2
Condition 10° – Condition 20°	1.1	1.0
Condition 10° – Condition 30°	3.1	2.5
Condition 20° – Condition 30°	1.8	1.6

Increasing knee flexion at contact may shift the shock attenuating responsibilities away from passive biological structures and toward active muscular contraction. Passive structures such as bone and cartilage are relatively stiff and the resulting tissue deformation in response to applied load is small (Armstrong et al., 1979; Lanyon et al., 1975). This does not allow for considerable energy absorption as evidenced by the higher FREQC observed at lower knee flexion angles. Muscle is more compliant than bone and cartilage and the elongation that takes place during eccentric muscular activity serves to absorb energy (Garrett et al., 1987), thereby removing more high frequency shock before it reaches the head. A change in knee flexion angle from 0 to 30° decreased the cutoff frequency of the body approximately 41%. This information may be applicable to therapeutic interventions for bone and muscular strength, such as whole body vibration (Torvinen et al., 2003). For example, deleterious frequencies associated with shock or vibration resonance may be avoided by manipulating knee angle, while still allowing for the transmission of safe anabolic frequencies (Rubin et al., 2003).

The nonlinear trend in ATT and FREQC with changes in knee flexion is difficult to explain. Derrick et al. (1998) suggested that shock attenuation at the knee can be partly explained by the moment arm (r_k) between the line of action of the ground reaction force vector and the knee joint center. For a given force vector, a larger r_k must be countered by a larger knee extensor moment (note the numerical accuracy of this model falls apart as we move proximally up the kinetic chain (Winter, 1990), but it provides a convenient conceptual framework to think about joint moments). The greater knee extensor muscle activity could lead to more energy absorption at the knee joint. However, changes to r_k would be expected to increase proportionately to the cosine of the leg angle, which is linear within the range of knee flexion angles that we investigated. It is possible that changes in knee angle were accompanied by a nonlinear change in the orientation of the ground reaction force vector, which ultimately would change r_k in a nonlinear fashion. Further study allowing for the direct assessment of ground reaction force would be necessary to verify this assumption. In addition, the amount of ankle and hip flexion were found to increase positively with knee flexion. These changes may have influenced the total lower-extremity flexion in a manner that increased the shock absorbing capacity of the lower extremity in a nonlinear fashion. For now, the issue of nonlinearity must remain a topic for future investigation.

The differences observed in ATT and FREQC among conditions were due to changes in both head and leg shock. This is contrary to human running in which head shock stays relatively constant and differences in leg shock are primarily responsible for changes in ATT (Hamill et al., 1995, Mercer et al., 2003). One argument for this phenomenon is that during locomotion, kinematic adjustments are made in an attempt to keep the head stable. Minimizing head movement allows for a more stable view of the visual field (Pozzo et al., 1991) and it provides the vestibular system with a steady gravitational reference (Pozzo et al., 1989). In the current study, kinematics were constrained and therefore participants were not able to choose a preferred posture that reduced head shock. It would be interesting to note the preferred geometric alignment during an inline skating impact protocol. It is doubtful that head shock would serve as an optimizing criterion in this scenario, as large increases in knee angle would necessitate strenuous muscle activation and considerable increases in oxygen uptake (McMahon et al., 1987). Previous research suggests that participants choose preferred kinematic strategies that optimize metabolic cost, rather than impact severity (Hamill et al., 1995).

The fact that intLeg tended to increase with knee angle, but the magnitude and frequency of shock reaching the head tended to decrease warrants further discussion. This result is best explained by the concept of effective mass (Derrick, 2004). Effective mass can be defined as the portion of body mass needed to model an impact if a single mass particle were used rather than a system of rotating and deforming segments. Effective mass is highly dependent on lower-extremity geometry at contact. Denoth (1986) estimated effective mass as a function of knee angle using a combination of experimental and modeling techniques during walking, running, and landing. For a 65-kg participant, an increase in knee flexion from 5 to 20° resulted in an effective mass change from 11 to 5 kg. Keeping all other parameters constant (impact velocity, joint stiffness, etc), a lower effective mass causes an increase in impact acceleration because the lower mass is easier to accelerate. Lafortune et al. (1996) also observed a positive relationship between leg acceleration and the amount of knee flexion at impact using a human pendulum device. Increases in impact acceleration were accompanied by corresponding decreases in impact force and should therefore not be considered an increased injury risk.

Extrapolation of these results to locomotion should be done with caution. Impact frequencies during running tend to occur between 12 and 20 Hz (Shorten & Winslow, 1992), and spectral analysis of the inline skating impacts revealed considerable power between 15 and 35 Hz. The different input frequencies to the musculoskeletal system could be associated with different muscle activation strategies, or "muscle tuning" responses (Wakeling et al., 2001). Running is also a more dynamic activity than our inline skating protocol; the increased lower-extremity joint excursion during impact in running would be associated with greater transfer of impact energy into rotational energy of the segments. The cutoff frequencies determined as a function of knee angle are therefore specific to this activity, and would likely be lower during locomotion. In our study, an accelerometer was mounted to the side of the inline skate. Such a rigid attachment is not available during running and in this circumstance accelerometers are commonly mounted to the distal leg (Valiant, 1990). Impact shock measured via leg-mounted accelerometers is subject to skin movement artifact. Although we anticipated differences between ATT during inline skating and the running literature, our values are in close agreement. The mean ATT during this study ranged from –7.8 to –20.1 dB, while ATT during running has been reported to range from –7.4 to –14.2 dB across a wide range of conditions (Derrick et al., 2002; Mercer et al., 2003).

In conclusion, we used an inline skating impact protocol to investigate the effects of knee flexion on shock attenuation. Our results suggest that the knee joint acts as a low-pass filter with a variable cut-off frequency that is dependent upon the amount of flexion. We believe that increasing knee flexion angle may shift the shock attenuation responsibilities away from passive tissues and toward active muscle tissue. Attenuation by active muscle tissue allows fewer of the high frequencies present in the impact to be transmitted to the head and thus improves shock attenuation.

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