THE INFLUENCE OF EFFECTIVE MASS ON IMPACT FORCE AND ACCELERATION

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Accelerometry is often used as a means to quantify the osteogenic or injury potential of impacts. This paper uses a series of four experiments to demonstrate theoretically, mechanically, and experimentally that increasing the effective mass of an impact can lead to an increase in impact force with a corresponding decrease in acceleration. The four experiments included: 1) mass spring models, 2) shoe impact testing, 3) cadaver impact simulation, and 4) an *in vivo* study manipulating knee angle during running. Results were consistent with the aim, illustrating a limitation for the use of accelerometers for impact assessment. In order to appropriately interpret the results from accelerometry it is necessary to quantify the effective mass of the impact. Failure to account for the influence of effective mass can lead to erroneous conclusions about impact severity.

KEY WORDS: running, landing, injury.

INTRODUCTION:

The impact force that occurs between the foot and the ground during locomotion and landing results in an impact acceleration that is transmitted up the musculoskeletal system. This impact acceleration is commonly measured by an accelerometer mounted to the distal leg. Researchers are interested in impact acceleration because it is assumed to play a role in the promotion of bone strength and the etiology of overuse injury.

Intuition suggests that high impact acceleration corresponds to high impact force, but this is not necessarily the case if the effective mass of the impact is reduced (Derrick, 2004). If researchers continue to use impact acceleration as a measure of osteogenic or injury potential it is important to understand the limitations of this technique so that accurate interpretations of the results can be made. The purpose of this study was to demonstrate theoretically, mechanically, and experimentally that increasing mass can result in an increase in impact force and a corresponding decrease in acceleration. This concept is illustrated through a series of four experiments.

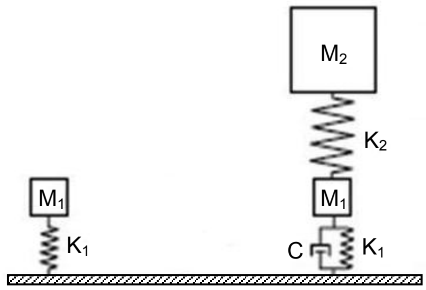

Figure 1: Schematic diagram of mass-spring models

METHODS:

Experiment 1: Vertical impacts and running ground reaction forces were modeled using a simple mass-spring system and a mass-spring-damper system, respectively (Figure 1). Impact velocity was held constant at -0.95 m/s. The simulations were run three times each with a different mass element (M_1) of 6.5, 8.5, and 10.5 kg. These mass elements correspond to effective masses typically seen during normal human running (Denoth, 1986). In the mass-spring-damper model total mass (M_1+M_2) was held constant at 70 kg. All other constants and initial conditions were taken from Derrick et al. (2000). The resulting forces and accelerations of M_1 were compared between mass conditions.

Experiment 2: An Exeter impact testing system (Exeter Research, Inc, Exeter, NH) was used to deliver impacts to the heel of a commercially available neutral running shoe. The missile head was dropped a distance of 5 cm with an impact velocity of approximately -0.95 m/s. Three testing sessions were performed each with a different impact mass (6.5, 8.5, and 10.5 kg). Each testing session consisted of 10 pre-impacts and 5 measured impacts. The force and acceleration experienced by the missile head was analyzed.

Experiment 3: Using the Exeter impact testing system, impacts were delivered to the heels of five human lower extremities with tibia/fibula osteotomy 20 cm above the malleoli. A uni-axial piezoelectric accelerometer (PCB Piezoelectronics, Model 353B, Depew, NY) was mounted to the skin of the distal anteriomedial face of the leg. Pro-flex elastic athletic tape was wrapped around the shank to minimize skin movement artifact. The Exeter impact testing system was situated over a force platform and a 3/8 in. aluminum rod was used to connect the proximal end of the tibia to a custom made missile head at the distal end of the impacting shaft. Foot angle was controlled prior to impact by applying tension to the anterior tibialis tendon.
Each cadaver extremity was dropped six times in each of two conditions, consisting of two different masses. Average impact masses were 8.77 ± 0.18 kg and 10.77 ± 0.18 kg. A drop height of 5 cm was used to approximate a contact velocity similar to Experiment 2. Accelerometer data were smoothed using a low-pass Butterworth filter with a cutoff frequency of 60 Hz. Peak forces and accelerations were compared between conditions using paired t-tests with a criterion alpha level set to 0.05.

Experiment 4: Five males and five females (age 25.3 ± 6.5 yrs; mass 68.6 ± 8.0 kg) ran off a platform (22.5 cm) at 2.7 ± 0.4 m/s. Three conditions were performed including: 1) normal running off the platform (NPR), 2) running of the platform with exaggerated knee flexion (FPR), and 3) running of the platform with exaggerated knee extension (EPR). Ten trails per condition were completed for each subject. An accelerometer was mounted to the distal anteriomedial leg of each subject. Lower extremity kinematics, ground reaction forces, and leg accelerations were collected concurrently. Ground reaction force curves were used to estimate the effective mass of the impact during each condition. The impact force curve was extracted from the ground reaction force curve (Derrick et al., 2005; Figure 2), and effective mass was calculated using the linear impulse-momentum relationship:

$$m_e = \frac{1}{\Delta v}\int_{t_1}^{t_2} F dt$$

where m_e is the effective mass of the impact, F is integrand of the linear impulse from foot contact (t_1) to peak impact (t_2), and Δv is the change in velocity from t_1 to t_2 estimated from the heel marker. Knee angle at contact, peak impact force, leg acceleration, and effective mass were compared between conditions using a repeated measures ANOVA with Tukey's *post hoc* tests. The criterion alpha level was set to 0.05.

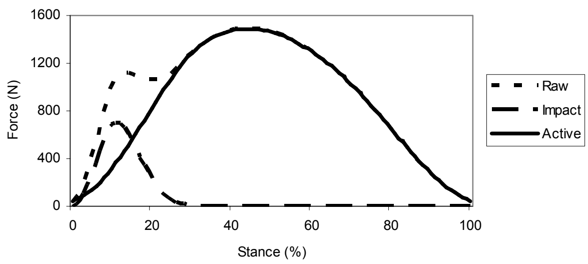

Figure 2: Extracted impact force curve.

RESULTS:

Experiment 1: Increasing M_1 resulted in an increase in impact force and a corresponding decrease in impact acceleration (Figure 3 and 4).

Experiments 2 & 3: Increasing the impact mass on the Exeter impact machine resulted in an increase in impact force and a corresponding decrease in impact acceleration during both the shoe tests and the cadaver tests (Figure 5 and 6).
For the shoe tests, increasing impact mass by 2 kg resulted in a 170 N increase in impact force and a 0.5 g reduction in acceleration (Table 1). For the cadaver tests, increasing impact mass by 2 kg resulted in a 170 N increase in impact force and a 1.0 g reduction in

acceleration (Table 2). The changes in impact force and acceleration were significant ($p<0.05$) for the cadaveric simulation.

Experiment 4: The three running conditions off the platform successfully manipulated knee angle at contact. On average subjects had 7° more flexion during FPR and 3° less flexion during EPR when compared to NPR (Table 3). Although not significant, a general trend in effective mass change with manipulation of knee angle at contact was observed. On average, an increase in knee flexion corresponded to a lower effective mass. Impact force was highest during EPR, but no differences were found between FPR and NPR. No differences in leg acceleration were found; however, an increase in average impact force was associated with a corresponding decrease in average leg acceleration.

Table 3. Impact force and acceleration (1SD) for shoe impact tests.

Mass (kg)	Force (N)	Acceleration (g)
6.5	697.2 (0.7)	10.9 (<0.01)
8.5	867.0 (6.1)	10.4 (<0.01)
10.5	1032.7 (7.8)	10.0 (<0.01)

Table 4. Impact force and acceleration (1SD) for cadaver impact tests.

Mass (kg)	Force (N)	Acceleration (g)
8.77	1195.4 (455.8)	7.8 (2.8)
10.77	1370.7 (543.7)	6.9 (2.9)

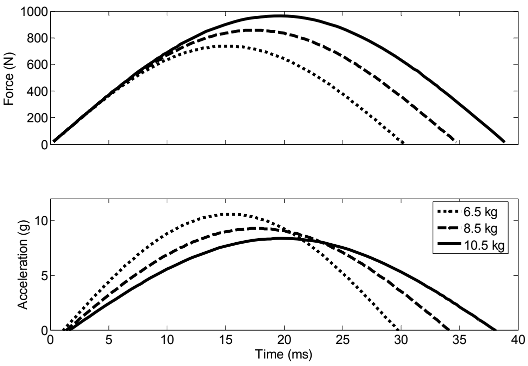

Figure 3: Impact force and acceleration for simple mass-spring model

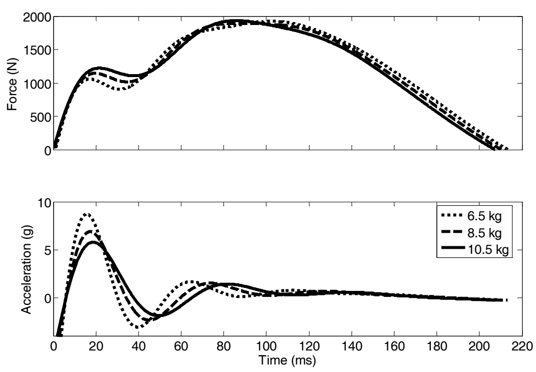

Figure 4: Ground reaction force and M1 accelerations for mass-spring-damper model

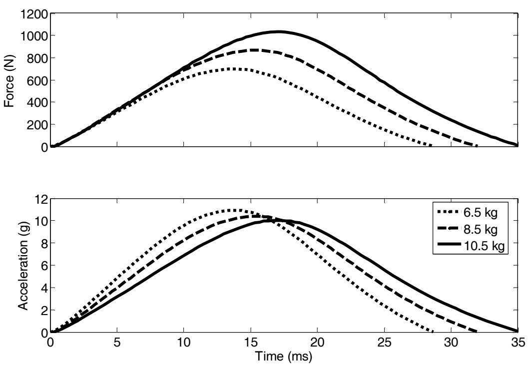

Figure 5: Impact force and acceleration for shoe impact tests

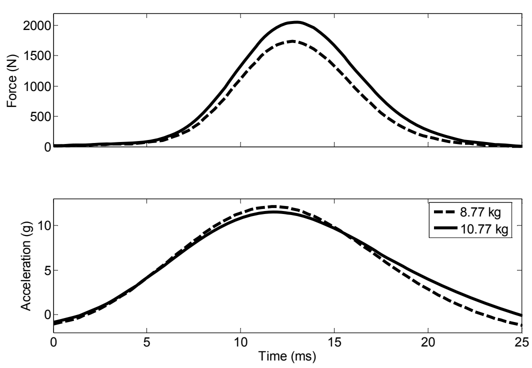

Figure 6: Impact force and acceleration for cadaver impact tests

Table 5. Knee angle at contact, impact force, leg acceleration, and effective mass for running

Condition	Knee Angle (°)	Effective Mass (kg)	Impact Force (N)	Leg Acceleration (g)
FPR	20.5 (5.2) [ne]	8.5 (5.4)	1671.8 (467.6) [e]	16.3 (5.7)
NPR	13.6 (4.5) [fe]	10.0 (4.3)	1825.5 (351.9) [e]	13.9 (5.8)
EPR	10.6 (3.7) [nf]	11.7 (5.2)	2111.3 (514.7) [nf]	13.5 (4.7)

[n]different from NPR; [f]different from FPR; [e]different from EPR; $p<0.05$

DISCUSSION:

The purpose of this study was to demonstrate that increasing the effective mass of an impact can result in an increase in impact force and a corresponding decrease in acceleration. This relationship was illustrated theoretically with mass-spring models, mechanically with a shoe impact tester, and experimentally with cadaveric impact simulation. Our *in vivo* study design only partially verified this theory, but the trends in mean values supported our aim. The impulse-momentum relationship is also mediated by impact velocity; if differences in velocity existed between conditions this could attenuate changes in impact force and acceleration due to effective mass. Nevertheless, significant increases in impact force were observed with no corresponding increase in impact acceleration. This demonstrates a limitation of accelerometry for the assessment of impact severity.

We propose that the osteogenic and injury potential of an impact acceleration is dependent upon the amount of mass being accelerated. For example, unpublished data from our lab found that clapping the hands produced accelerations on upwards of 200 g's. This is about 30 times greater than leg accelerations during running, but it is difficult to argue that the potential for injury is higher for clapping than it is for running.

CONCLUSION:

The results of this study have implications for clinicians and researchers wanting to use impact acceleration as a means to quantify impact severity. When making comparisons between conditions in which changes in lower leg geometry at impact occur the effective mass of the impact must be known in order to appropriately interpret the results. A more extended knee at contact may correspond to a higher effective mass, and while this higher mass results in an increase in impact force, it is harder to accelerate and can result in a decrease in impact acceleration.

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