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Are maximum ground forces and leg compression in phase? A test of the classical spring mass model of running gaits

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Are maximum ground forces and leg compression in phase?

A test of the classical spring mass model of running gaits

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Abstract:

The mechanical understanding of human running has classically been described as a spring-mass system, with subsequent models predicting the movements of the body's center of mass and the forces applied by the leg against the ground. A central requirement of any spring system is the phasic relationship between the length of the elastic elements and the forces applied to these structures. Specifically, elastic elements compress under load and extend as the load is released. We tested whether this model applies to individuals with specialization for extreme performance in human gait. Recent work from elite level sprint runners suggest that their patterns of force application differ from those used during slow speed running, and similarly differ between individuals capable of high speed running and those that are not. We measured force application and center of mass movements in collegiate sprinters (n=7; top speed 10.1 ± 0.7 m s^{-1}) and recreational runners (n=9; top speed 8.4 ± 0.1m s^{-1}) as they ran on an instrumented force treadmill at speeds spanning each individual's range. Between these groups we found sprinters applied greater stance average forces at common speeds (mean difference = 11 ± 0.2%) and used an asymmetrical pattern of force application to do so when running at speeds great than 7.0 m s^{-1}. Further at speeds greater than this threshold peak force application preceded minimum center of mass height by 13±1% when expressed relative to the duration of foot-ground contact. This result produced force-length relationships, a method to describe the elastic properties of the leg, that were unique among terrestrial species displaying increased compression of the leg despite lesser levels of force application. We conclude sprint runners use novel gait mechanics to obtain increased whole-body performance rather than a reliance on the storage and release of elastic energy, classically documented at low speeds and for recreational runners.

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Chapter 1 -- Introduction:

The legs of human runners alternate between periods of foot-ground contact where the limbs support the body's weight, and aerial periods where the legs are repositioned in anticipation of the next foot-ground contact. During stance, the majority of the force is oriented vertically against the ground by the leg to counteract the earth's gravity (Cavagna *et al.*, 1977; Weyand *et al.*, 2000; Weyand *et al.*, 2010). These forces compress the leg, and yielding at the joints results in the downward travel of the center of mass.

The classical spring-mass model has been developed to describe the motion of a runner's center of mass. The model treats the center of mass as a point bouncing on a massless linear spring (Bickhan, 1989; McMahon & Cheng, 1990). Most iterations of these models use simple harmonic motion and expect a sinusoidal trajectory of center of mass displacement with the minimum height occurring roughly at mid-stance and in phase with the peak of leg force application. In the second half of stance phase, the center of mass reverses direction and moves upward as the leg extends and the body enters the aerial phase (fig. 1) (Blickhan, 1989; McMahon 1990, Farley *et al.*, 1990). At slow speeds (3-6 m s^{-1}), the spring-mass model provides a very adequate prediction of the center of mass displacement and the ground reaction force waveforms (McMahon & Cheng 1990; Blickhan, 1989; He, 1991, Farley and Gonzalez 1996; Clark et al 2014).

The center of mass movement patterns allow elastic strain-energy to be stored during the first half of stance and recovered in the second half of stance phase by the muscle-tendon units (Alexander, 1988; Roberts *et al.*, 1998; Cavagna 2006). These energy transfers reduce the amount of mechanical energy that must be met through muscular activity and thus, most of the investigations leading to this understanding have been conducted at relatively slow running speeds where both the mechanical and the metabolic inputs can be readily measured. However, recent results from high-speed running suggest this classical understanding of leg function may not apply across the entire range of human speeds (Clark *et al.*, 2014). Specifically, elite athletes sprinting on a force treadmill have produced ground reaction force waveforms that are asymmetrical from 3.5 m s^{-1} to the top speed of the athlete (Clark *et al.*, 2014). This asymmetry is due to much greater forces applied in the first half of stance than those during the second half. This deviation from the predictions of the spring-mass model provide an opportunity to test the classical understanding of leg function and evaluate whether gait dynamics or the biological properties of the leg contribute to the limits human running.

These recent studies have utilized specialized high-speed force treadmills to capture the ground reaction force waveforms over multiple footfalls at the same speed as highly trained runners approach the limits of human running performance (Bundle *et al.*, 2015). Because earlier studies necessarily relied on single foot-fall data and potentially non-steady speed running, the previous literature describing the movements of the center of mass has been inconsistent. The substantial variability in these results comes from differing protocols, as well as inconsistent force measurements from multiple trials using a single footfall on a force plate. This technique increases the variability present in the running speeds at which these subjects strike the force plate, an event highly dependent on speed, and typically introduce whole-body accelerations or decelerations throughout the instrumented zone. These single footfall data sets preclude statistical inference and generation of means and variances. Despite these limitations previous data indicate a roughly negative relationship between the speed of the runner and center of mass displacement.

Reductions in the deflections experienced by the center of mass as speed increases are indicative of a change in the material properties of the leg. Classical material engineering theory defines stiffness as the stress applied over the strain of the material (McMahon, 1990). In the case of the leg, the stress is measured as the vertical ground reaction force (F_z) without an area term. Yield of the limb or the change in the height of the center of mass is considered to be the strain (Δy) (McMahon, 1990). This quantity (K_{vert}) is considered vertical stiffness, or alternatively the intrinsic stiffness of the limb and is calculated as:

$$K_{vert} = \frac{F_{z\ max}}{\Delta y} \qquad \text{eq. (1)}$$

Because ground reaction forces increase at faster speeds as center of mass deflections decrease, vertical stiffness of the leg also increases with speed (Munro *et al.*, 1987; Weyand *et al.*, 2000; Stafilidis & Arampatzis, 2006; Morin *et al.*, 2006, Weyand *et al.*, 2010). The limitations of this value are it assumes the peak force application and the minimal center of mass displacements are in phase. This value accounts for whole system stiffness, not the stiffness of the leg or any particular joint.

The alteration of running speed, muscular force application, center of mass dynamics and the intrinsic stiffness of the limb are controlled by the central nervous system. Under these circumstances neural control of muscle force can occur in two ways: through the more forceful contraction of the musculature or via pre-activation of the active muscles (Moritz & Farley, 2005). Throughout the stance phase, force application is thought to be governed by the classical Henneman properties of the muscle (Mero and Komi 1986). Therefore, as a runner increases speed, the increase in force application occurs through a similar increase in neuromuscular activity in the leg (Mero & Komi, 1986). Neuromuscular activation begins in the milliseconds leading up to ground contact and remain active throughout stance phase (Mero & Komi, 1986). Pre-activation likely stiffens the muscle-tendon units by reducing the slack within the sarcomere and may lessen the yield of the limb upon impact with the ground, and is the most commonly accepted mechanism used to increase stiffness of the limb during terrestrial gait (Bosco *et al.*, 1987, Heglund & Cavagna, 1985; Finni *et al.*, 2001; Moritz & Farley, 2004).

In order to develop a more thorough understanding of running gaits, previous studies have manipulated the way in which subjects come into contact with the ground. For instance, McMahon and colleagues (1987) changed the mechanics of the leg during stance phase by having subjects 'Groucho' run with deep knee flexion (McMahon *et al.*, 1987). The ground reaction force waveforms produced by Groucho running at slow speeds also deviate from the spring-mass model (McMahon *et al.*, 1987). The deviation in the Groucho running ground reaction force waveforms are indicative of a normal heel strike followed by a prolonged stance phase due to deep flexion of the knee; neither element is well characterized by the spring-mass modeling. Another way the mechanics of the leg have been altered has been to change stride frequency at a set speed (Gonzalez & Farley, 1995). By changing stride frequency at a common speed, these investigators manipulated the material properties of the leg by inducing altered force application and displacement of the center of mass. At decreased stride frequencies, the force application occurred over a greater period of time, and allowed for greater center of mass displacements. While at the increased stride frequencies the force application, center of mass displacement and contact time decreased as well. Finally, in barefoot running different runners apply ground forces differently against the ground depending on which of three portions of the foot struck the ground first (Liebermann *et al.*, 2010). These studies have begun to show the limits of the spring-mass considerations, and the dynamic interaction between ground force application, center of mass displacement and functional changes to the material properties of the limb in response to a more or less common motor task.

Thus, the purpose of our study was to examine of whether the asymmetrical force application present in the gait of individuals with the capability for high speed running led to similarly asymmetrical movements of the body's center of mass, i.e., are the force application and center of mass displacements in phase or not? To address this question, we measured ground reaction forces across the entire running speed range and evaluated the fraction of stance at which the maximum forces were applied and the minimum leg lengths were observed. We further analyzed the force-length relationship of the leg during stance to estimate the relative contributions of the active and passive mechanisms involved in transferring force from the body's muscular and skeletal elements to the environment. Based upon the extensive reliance on elastic mechanisms in biological movement (Alexander, 1992; Biewener *et al.*, 1998; Cavagna *et al.*, 1977, Roberts *et al.*, 1997; Tobalske *et al.*, 2003) and nearly 50 years of empirical results on human gait dynamics, we expected the rate of leg compression to be similar to that of force application, in the two groups studied: sprint and recreational runners. This maintains one of the basic requirements of the running gait: a phasic relationship between force and displacement of the center of mass. We subsequently expected that the ground force – leg length workloops would display classical elastic properties, contain relatively little hysteresis (Biewener *et al.*, 1998), and indicate a greater reliance on passive mechanisms to apply the necessary levels of ground force.

Chapter 2 – Thesis Requirements

Statement of the Problem:

It is known that the speed at which an athlete runs is dependent on ground reaction force application (Weyand *et al.*, 2000). We also know in order for running to energetically efficient, there must be a way to conserve energy throughout gait, and this is done through the loading and unloading of the muscle tendon units which act as springs (Alexander 1988, McMahon 1990). The review of literature does not show how humans apply force against the ground across the range of speeds and how this effects the trajectory and total displacement of the center of mass, intrinsic stiffness of the leg and the material properties of the limb. In recent years, it has been shown sprint runners apply forces in a different manner than other groups, including athletes non-sprinters and recreationally trained runners (Clark and Weyand, 2014).

Significance of the Study:

This study will investigate how speed affects ground reaction force application, intrinsic stiffness of the leg, and center of mass dynamics of a runner. We used two groups, sprint runners and recreationally trained runners to examine the differences in the ways each group strikes the ground. What makes this study unique is: 1) study of gait mechanics, ground reaction force application and the material properties of the limb from slow speeds to the top speed of the subject on an instrumented high-speed treadmill, and; 2) How changing the way in which a subject strikes the ground changes the gait of the subject and consequently the function of the leg spring system.

The recreational group were also asked to change the way in which they struck the ground in order to elicit similar ground force application patterns akin to sprint runners. They did this by striking the ground as hard as possible at three speeds, 3, 5, and 7 m s^{-1}. They were also asked to strike the ground as soft a possible at these same speeds.

Research Hypothesis:

<u>Hypothesis 1:</u> The spring-mass model will not accurately represent the center of mass displacements and the ground reaction forces applied by the subjects when their foot contacts the ground at faster speeds. The application of force against the ground in the fastest sprint subjects will occur around 30% of contact time, and the maximal displacement of the center of mass will also occur around this time during stance (Clark *et al.*, 2014). The elastic properties of the limb will also be maintained by the recreational runners; however their peak ground reaction force application and minimum center of mass displacements will occur at approximately 45% of stance phase.

<u>Hypothesis 2:</u> The hard footstrikes will show the shortcomings of the spring-mass model. The application of forces throughout the stance phase for hard footstrike will be more like the footstrike patterns of sprinters. The maximal center of mass displacements will occur in phase with the peak ground reaction forces. The soft footstrikes will align more with the spring-mass model in the application of forces and calculated center of mass displacements. The ground reaction force waveform will have peak force application around 50% of ground contact time and the peak center of mass displacement will occur at the same time.

Limitations and Delimitations:

The analysis process is predicated upon the collection of ground reaction force waveforms from a variety of subjects and have the sample include high level sprinters. The sample did not include elite sprinters who have shown the greatest asymmetry in their waveforms at high speed. We studied a recreational and collegiate athlete population, because they had differing ground force application patterns and were readily available for testing. This allowed for the analysis of a variety of footstrikes. We only had recreational athletes complete hard and soft footstrikes at the speeds of 3, 5, and 7 m s^{-1}.

We did not use 3-D high speed video to measure the gait kinematics of the subjects. This tool would have provided us with data about the function of the individual joints and another measure of displacement of the center of mass. We also did not calculate the stiffness of the leg; we calculated the stiffness of the entire system, k_{vert}. There are many ways to measure leg stiffness, and it is inconsistent across studies. Vertical stiffness offers a whole system measurement, there were not any individual joint stiffness's included in this work, nor is leg stiffness used as a measure of stiffness. We also did not use electromyography to measure the neural control of the limb throughout gait. This does not allow for us to examine the pre-activation of the muscle or how the increase in speed relates to the Henneman size principle.

Rationale of the Study:

There has not been a comprehensive study to measure how ground reaction forces, gait kinematics, and center of mass dynamics change as speed is increased from slow to top speed. This study tested the limits of the spring-mass model and how well it represents the center of mass displacement and material properties of the limb at high speeds and footstrikes.

Chapter 3 -- Methods:

Subjects:

Nine recreational (M_b = 78.8 ± 1.3 kg; mean ± SE) and seven sprint trained M_b = 81.2 ± 1.3 kg) male subjects provided their written informed consent in accordance with the Institutional Review Board at the University of Montana.

Treadmill data collection:

Subjects wore a safety harness to suspend them above the treadmill in the event of a fall. At the beginning of each trial, subjects lowered themselves onto the moving treadmill belt by transferring their weight from the handrails to the tread. The subjects were encouraged to take as many weight-assisted steps as necessary to maintain balance; typically, this transfer required fewer than 6 steps. At lesser speeds, trials involved up to 30 seconds of continuous running, whereas trials approaching the subject's top speed were as brief as 8 consecutive steps, i.e. roughly 3 seconds. Subjects were allowed to select the rest period between trials.

The top speeds of the subjects were determined by an incremental test to failure. Trials were considered successful if the subject achieved eight consecutive footfalls without more than 20 cm of forward or backward movement during the trial (Weyand *et al.*, 2000). The testing was typically completed over two visits to the laboratory. The first day of the protocol consisted of slow running, generally less than 7 m s^{-1}. Following a self-selected warmup, testing began at 3.0 m s^{-1} and was increased for subsequent trials by 0.5 m s^{-1} up to 7 m s^{-1}. The second day of the protocol required the subjects to run to their top speed. As the subjects approached their top speed, each trial's speed was increased by 0.2 m s^{-1} until subjects could no longer successfully complete a trial. A few of the subjects completed their top speed test on subsequent visits, (3^{rd} or

4th), to laboratory due to taxing personal training schedules or pre-existing cases of muscle soreness.

Ground reaction force data was collected on a custom high-speed force treadmill as described in detail by Bundle and colleagues (2015). The data was collected during foot-ground contact by the four load cells located at the corners beneath the treadmill bed. The force measures from the load cells were amplified (MiniAmp MSA-6) and digitized (Digidata 1322A, Axon Instruments Inc.) to computer at 2000 Hz. Signals were conditioned with a 40 Hz low-pass zero-lag 6 pole Butterworth filter in a custom Matlab application. The per trial ground force waveforms were normalized with respect to body weight providing force in multiples of the body's weight (xW_b) .

Contact time (s): The foot-ground contact times were determined from the continuous period during which the vertical treadmill reaction force exceeded 50 N. Reported values are the pooled means of the individual trials administered at a particular speed. The measures representing each individual trial were obtained from the analysis of eight consecutive steps.

Stance-average forces (N & xWb)*:* The stance-average vertical ground reaction forces were the mean value of the ground force waveform during the period of foot-ground contact.

Center of mass displacement (Δ CoM): The vertical center of mass displacements were calculated by twice integrating the ground reaction forces from the eight-step record using the method of Cavagna (1975). The specific trial means were generated from the collection of these eight waveforms.

Stress-Strain Properties of the Leg: The measures of center of mass displacement and ground force application were used to calculate the leg's vertical stiffness (k_{vert}) as described in equation 1. The shape of and area within a stress-strain curve, or workloop, describes the functional material properties of the leg, indicating whether elastic properties dominate or whether energy is being absorbed from or released to the environment (Josephson, 1985). We evaluated the ground force – leg length relationship during the loading and unloading phases of stance. We considered the area within the resulting workloop to represent the mechanical work done by the active musculature (Josephson, 1985; Farley *et al.*, 1990) throughout the stance phase.

Harmonic Model: The simple harmonic model produced an estimate of the forces applied against the ground given the measured values of contact time, aerial time (the period between successive stance phases), and the mass of the subject. These force waveforms were also integrated twice to determine the center of mass displacement (Cavgana, 1975).

Hard vs Soft Footstrikes:

A subset (n = 6) of the recreationally trained subjects completed six additional trials at speeds of 3, 5, and 7 m s^{-1}. Before each trial, they were instructed to either strike the ground as hard or as softly as possible. Each trial consisted of at least 8 steps at each speed; if there was any drift by the subject on the treadmill during the trial the trial was attempted again.

Statistical analysis:

We evaluated the between-group, sprinter vs. recreational comparisons for top speed running, contact time, stance average vertical ground reaction forces, center of mass displacements, vertical stiffness, the fraction of ground contact when peak vertical ground reaction forces occurred, and the fraction of ground contact when maximal displacement of the center of mass occurred using two-tailed independent t-tests. The hard and soft footstrike data were tested for significant differences between conditions and normal footstrike data with the aforementioned variables using a paired two-tailed t-test. The *a priori* p-value for significance was set to an alpha level of 0.05. The data are reported as means with standard error.

Chapter 4 – Results

Gait mechanics as a function of speed:

For all the subjects in this study, the ability to run faster was predicated on an increase in the application of ground reaction forces during shorter ground contact times (fig. 3). At slower speeds, there was a greater difference in contact times between groups than at faster speeds; the largest difference with the recreational runners in ground contact time was 0.02s at 3.0 m s^{-1}, $p = 0.008$. For the speeds above 5.5 m s^{-1}, there was no significant difference in contact time between sprinters and recreational runners, $p > 0.05$ (fig. 3, A) at common speeds. The slight increase at the fastest speeds occurred because of a decrease in sample size.

The stance averaged forces with respect to body weight (xW_b) were distinctly different between the groups; sprinters struck the ground with more force throughout stance than non-sprinters with an average difference of 11±0.2% more force applied at each speed, for each speed from 3 m s^{-1} to 8.5 m s^{-1} stance average ground reaction forces were significantly different, $p < 0.05$. At 9.0 m s^{-1} there was not a significant difference between groups, $p = 0.82$, due to the small sample size of recreational runners, and the difference between the two groups was 0.03 xW_b or 1.7% (fig 3, B). The increases in force application remained approximately linear across the range of speeds for both groups: for recreational runners, $y = 0.08x + 1.29$, R^2=0.92, and for sprinters, $0.07x+1.57$, R^2=0.93.

The maximal center of mass displacement decreased as speed increased for both groups until approximately 8.0 m s^{-1}. The sprint group's center of mass displacements were 9±0.2% larger than the center of mass displacements of the recreational group's across the range of speeds. At speeds faster than 8.0 m s^{-1}, the center of mass displacements remained approximately constant for each group at 2.0 cm of displacement at each speed (fig 3,C).

The vertical stiffness of the limb increased with speed for each group. The sprint group's limbs were not significantly stiffer than the recreational groups across the range of speeds, $p > 0.05$ for all speeds, the difference between groups was less than 12 kN m^{-1} at common speeds. The relationship between stiffness and speed was not linear; there appears to be two different slopes associated with vertical stiffness, the first from 3.0 to 7.5 m s^{-1} and second from 7.5 m s^{-1} to top speed (fig 3, D). The increase in stiffness after 7.5 m s^{-1} is notable because after this point, the center of mass displacement remains approximately constant and stance average forces did not change slope.

The recreational runners did not show the phasic decoupling of the ground reaction forces and calculated center of mass displacements the sprint groups showed (fig 4). The application of force generally occurred in phase. The average difference between peak force application and peak center of mass displacement for the recreational group was 2±0.2% of contact time. The sprint group applied forces against the ground earlier in stance than the recreational group by a minimum of 2% of ground contact time across the range of speeds. At the speed of 7.0 m s^{-1} the sprint group began to apply ground reaction forces earlier in stance. From 3.0 to 7.0 m s^{-1} the average difference between the peak force application and the peak displacement was 2±0.1%, similarly to the recreational group, suggesting a spring like function of the limb. At speeds above 7.0 m s^{-1} phasic decoupling of peak ground reaction forces and center of mass displacements occurred; the average percent difference between peak force and peak displacement for these speeds was 12±0.2%.

Top speed kinetics:

The top speeds of this experiment ranged from 9.7 to 10.6 m s^{-1} for the sprint group and from 7.7 to 9.2 m s^{-1} for the recreational group. The average top speed for the sprint group was 10.1±0.7 m s^{-1} and for the recreational group was 8.4±0.1 m s^{-1}, Δ=18%, $p > 0.001$ (table 1). The sprint group spent 16% less time on the ground than the recreational group, and applied significantly more force, 0.31 xW_b than the recreationally trained group, $p = 0.002$. The vertical deflection of the center of mass in the sprint group at top speed was 15% less than the recreational group, $p = 0.09$. This indicated sprinters had a greater intrinsic stiffness than the recreationally trained group (table 1). The difference in vertical stiffness at the top speeds of the subjects differed by an average of 45 kN m^{-1}, or 34%, $p < 0.001$. The application of forces against the ground were different between the groups as well: the sprint runners applied peak forces 11% earlier in contact time than the recreational group, $p = 0.009$. The peak center of mass displacements occurred at 43 ±1% contact time for the sprint group, and 46±1% contact time for the recreational group, $p = 0.02$.

Sprint Athlete Workloops:

Workloops were used to show the force - displacement relationship between the predicted harmonic model force and displacement curves and those from the data (fig 5). The workloops for this representative sprint subject shows two distinct stiffnesses of the limb during the loading and unloading during stance, particularly at top speeds. The stiffness of the limb during the loading phase was greater, due to the slope of the curve, than both the stiffness of the limb predicted by the model and the measured unloading phase of the limb (fig 3). The simple harmonic model does not account for the rapid loading of the limb during the first half of stance phase, nor does the model account for variable stiffness of the limb throughout stance phase.

Hard and Soft Footstrikes:

The hard footstrikes were distinctly asymmetrical, and consist of the rapid loading of the limb during the first instances of ground contact (fig 6). The soft footfalls were approximately sinusoidal and more readily match the prediction of the simple harmonic model for the normal footfall at each speed. There was a slight deviation in the 7.0 m s^{-1} trial with a more rapid loading of the limb than in previous soft footstrike trials (fig 6, C). The center of mass displacements of the hard footstrikes occurred after the peak force had been applied in all three cases. The soft footstrike's center of mass displacement occurred at approximately the same time as peak ground reaction force application as predicted by the classical spring-mass model.

Hard and Soft Footstrike Kinetics:

The hard footstrike contact times was significantly shorter than soft footstrike, $p < 0.001$ at each speed (table 2). The difference between the contact times of hard footstrikes and normal footstrikes, at the speeds of 3.0 m s^{-1}, p=0.03, 5 m s^{-1} p =0.05, and 7 m s^{-1} p =0.09. The differences between the contact time of hard and soft footstrikes follow the same pattern, however the difference in contact time is 0.03s, 3.0 m s^{-1}, p=0.01, 5 m s^{-1} p =0.01, and 7 m s^{-1} p =0.01 (table 2). Both footstrikes' appear to converge with the normal footstrike contact time (fig 7, panel A).

It follows that hard footstrikes had greater stance average ground reaction forces than the soft and normal footstrikes; these differences were significant at 3.0 and 5.0 m s^{-1}, hard 3 m s(table 2). At 7.0 m s^{-1}, there is not a significant difference between the stance average forces, of any of the footstrikes. (table 2). The percent difference between the stance average ground reaction forces for the hard footstrikes is only 7±0.8% between 3.0 and 7.0 m s^{-1} while the soft footstrikes show a 14±0.4% difference between the average force applied at the same speeds. The soft footstrikes were shown to not be statistically different from the normal footstrikes at any of the three speeds with respect to stance average ground reaction forces. The stance average ground reaction forces also appeared to be converging to the normal footstrike stance average force as speed increased for both types of footstrikes.

The center of mass displacements for the hard footstrikes at 3.0 m s^{-1} were greater than and statistically different from soft p = 0.65, and normal footstrikes, p=0.64 (table 2). At the speeds of 5.0 and 7.0 m s^{-1}, the vertical center of mass displacements of the hard footstrikes were significantly less than both the soft, 5 m s^{-1} p= 0.04, 7 m s^{-1} p = 0.01 and normal footstrikes, 5 m s^{-1}, p =0.001, 7 m s^{-1}, p = 0.001. The center of mass displacements of the soft footstrikes were almost identical to those of the normal footstrike data, with a 36±0.8% change in center of mass displacements for soft footstrikes, a 38±0.7% change for normal footstrikes and an 43±1.3% change for hard footstrikes.

The vertical stiffness of the limb at 3.0 m s^{-1} for the hard and soft footstrikes was not significantly different than the normal footstrike $p > 0.05$ (table 2). For the speeds above 3.0 m s^{-1}, the hard foot falls were significantly than the normal at the speeds of 5 m s^{-1} p =0.02, and 7 m s^{-1} p =0.03, and for all speeds the vertical stiffness of the hard foot strike was significantly different than the soft footstrikes 3.0 m s^{-1}, p=0.04, 5 m s^{-1} p =0.01, and 7 m s^{-1} p =0.01. The soft footstrikes were not significantly different than normal footstrikes at each speed, 3.0 m s^{-1}, p=0.87, 5 m s^{-1} p =0.47, and 7 m s^{-1} p =0.30; with the greatest difference between the vertical stiffness's of the two different footstrikes occurring at 7.0 m s^{-1}, Δ=11 kN m^{-1}. From 3.0 to 7.0

m s^{-1}, the hard footstrikes increased the vertical stiffness of the limbs by 48±1.0%, the soft footstrikes by 46±0.7% and the normal footstrikes by 41±0.6%.

The maximal displacements of the center of mass remained relatively constant across the range of speeds and footstrikes, with all of the peak displacements occurring between 44 - 47% of contact time. The differences in force application were more pronounced: the hard footstrikes show a phasic decoupling between the peak ground reaction forces and the maximal displacement of the center of mass, as observed with the sprint group, at 3.0 m s^{-1}, Δ=4.4%, at 5.0 m s^{-1}, Δ=8.2% and at 7.0 m s^{-1}, Δ=10%. The soft footstrike's peak ground reaction forces occurred closer to the midpoint of stance phase than the hard footstrike, suggesting a more spring-like function of the limb (fig. 8, A and B). The force and center of mass displacements were not significantly different at any speed for the soft foot strikes, 3 m s^{-1} $p = 0.79$, 5 m s^{-1} $p = 0.19$ and 7 m s^{-1} $p=0.86$. Significant differences were seen with the hard foot strikes, 3 m s^{-1} $p = 0.04$, 5 m s^{-1} $p = 0.04$ and 7 m s^{-1} $p=0.01$.

Footstrike workloops:

The soft footstrikes maintained the spring-like function of the leg were better as speed increases (fig 9 panel A). Less work was required of the muscle as evidenced by the area between the loading and unloading portions of the curve. The overall stiffness of the system for soft footstrikes was shown to be greater than those predicted by the simple harmonic model at the speed of 5.0 and 7.0 m s^{-1}. These workloops deviated from the simple harmonic model with more rapid loading of the limb. The loading of the limb deviated from the spring-mass model and the soft footstrikes by having two different slopes and greater stiffness across the range of speeds. The unloading of the limb had a single stiffness much like the soft footstrikes, which was greater than those predicted by the harmonic model.

Chapter 5 – Discussion

We tested wether the classical understanding of the leg functioning as spring to capture and release elastic energy during running is valid in individuals capable of high-speed performances. Our results do not support this expectation. We evaluated whether the movements of the body's center of mass occur in phase with the application of ground reaction force by the leg. They do not. At speeds above 7 m s^{-1}, the group of sprint trained subjects exhibited the assymetrical ground force waveforms (Fig. 2) that have been recently identified in Olympic and national caliber sprinters by Clark and colleagues (2014). Our data inidicate that at these speeds, the peak ground reaction forces occurred at 30% of the stance phase (Fig. 4); in contrast the temporal loaction of the minimum center of mass height was essentially constant across the range of speeds occuring close to 46% of the stance phase. The phasic decoupling of force application by the leg spring and displacements of the center of mass are indicative of a breakdown of the spring-like function of the leg.

We subsequently analyzed the ground force – leg length workloops to evaluate whether these kinematic alterations were accompanied by variance in the active vs passive requirements of gait. These data indicate that sprinters (Fig 5), and individuals endeavoring to strike the ground with as much force as possible (Fig. 9) deviate from the classical expectations of elasticity (Fig 1). For both groups, the workloops contain considerable hysteresis indicating a requirement for muscular contribution that is absent in the gait of non-trained individuals at all speeds and for sprint specialists at lesser speeds. These data indicate that sprint runners must provide mechanical power, in addition to very high levels of force application in extermely brief periods to achieve high speed performances.

The hysteresis observed in the force – length workloops (Fig 5 & 9) indicates energy that is absorbed or released depending on the direction that the workloop turns as it completes its trajectory (Joesphson, 1985). Because the conventions of the muscular- and gait-related literature vary with respect to the expression of length change, it is not evident from our figures whether work is being provided to the environment or absorbed from it. When the workloops were analyzed prior to the rectification step common for gait analysis the workloops traveled in a counter-clockwise rotation indicating the generation of muscular power (Josephson 1985). The workloops we observed were unconventional in their shape due to the different slopes of the loading and unloading segments; this produced a loop with a figure 8 shape. This adds uncertainty to the interpretation of these data, and either represents the addition of net power, or a requirment for instantaneous power which may be transferred across the limb via bi-articulate muscles (Kuo & Donavan, 2005 ; McGowan *et al.*, 2013). The data indicate that throughout the stance phase work is done by and on the system. However, our use of ground force application to determine the kinematics of the center of mass does not provide the detail necessary to evaluate this possibility.

These are the first data to show the movements of the COM during normal gait do not conform to the classical spring-mass characteristics. Furthermore, these data show energy cannot be passively returned in the second half of stance (fig 5 & 9). This study also provided further evidence that sprinters are different from both recreational and longer-distance runners because they apply more force at all speeds. Additionally, sprinters at speeds greater than 7.0 m s^{-1} apply ground reaction forces earlier in stance, leading to an asymmetrical ground reaction force waveform. This caused hysteresis in the force-length relationship and required energy input. We were able to illicit similar limb function when individuals were asked to strike the ground hard, as sprinters do. The results indicate a further element of difficulty and selection that must be overcome for individuals interested in performing at the elite level.

We studided the function of the limb with respect to classical spring mass model. We have shown sprint runners will strike the ground in a manner that reduces the amount of elastic energy retrun in gait. The systemic properties of the limb and running gait have been described by this study, given the complexities of the muscloskeletal system we are unsure of the details and were limted by the force only approach we took. However, the evidence presented showss the limitations of the classical spring mass model. The next step is to determine the neuromuscular control of the limb at faster speeds, and the specific muscular work done by each joint. Thus enhancing our understanding of the function of the limb at lower levels of biological organization, and the overall control of gait.

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Figure Captions

Figure 1: Force and displacements of the spring-mass model. Panel A shows the classical spring-mass model superimposed on the human runner's limb, showing the initiation of contact and loading and unloading of the limb. Panel B: simple harmonic model predications of ground reaction forces applied and the trajectory of the displacement of the center of mass of a sprinter running at 10.6 m s^{-1} Panel C: The predicted stress-strain curve of the simple harmonic model for the waveforms in panel B, showing minimal hysteresis between the loading and unloading of the limb.

Figure 2: The 8-step average ground reaction force waveforms of a sprint athlete in the top panel and the corresponding calculated center of mass displacements at the speeds of 3.0 m s^{-1} (gray) and 10.6 m s^{-1} (black).

Figure 3: Comparison between the gait kinetics of sprint athletes and recreational runners across the range of speeds each group was capable of. Panel A shows the contact time of stance phase for each group. Panel B shows the stance average ground reaction forces. Panel C shows the maximal absolute calculated center of mass displacement during stance phase. Panel D shows the vertical stiffness of the limb.

Figure 4: Temporal differences in force application between recreational runners and sprinters as a percent of contact time. In the top panel, the recreational runners' peak ground reaction force application and the maximal displacement of the center of mass occurs at approximately the same time during the stance phase. In the bottom panel: sprinters apply forces against the ground differently, by applying peak ground reaction forces well before the maximal displacement of the center of mass at 7.0 m s^{-1} is 11% and at top speed the difference is 12%.

Figure 5: The stress-strain curve of a sprinter compared to the stress strain curve of the simple harmonic model. The red lines represent the results of the simple harmonic model for this runner at each of the three speeds, matched with peak ground reaction forces. The blue lines are data from the sprinter: the compression of the leg is represented by the thicker blue line, and the thinner blue line shows the period of the stance phase in which the leg is extending.

Figure 6: The spring-mass model in comparison to hard and soft footstrike ground reaction forces, calculated center of mass displacements and the predication of the harmonic model. Panel A has the waveforms for 3.0 m s^{-1}, panel B are 5 m s^{-1} waveforms and panel C are 7 m s^{-1} waveforms.

Figure 7: Comparison between recreational runners' gait kinetics striking the ground as hard and as softly as possible at the speeds of 3, 5, and 7 m s^{-1} compared to the normal footstrike data. Panel A shows the contact time of stance phase for each footstrike. Panel B shows the stance average ground reaction forces. Panel C shows the maximal absolute calculated center of mass displacement during stance phase. Panel D shows the vertical stiffness of the limb.

Figure 8: The differences between hard, soft and normal footstrikes with the range of speeds. Panel A shows the center of mass displacements for the different footstrikes. Panel B shows peak force application as a percent of contact time.

Figure 9: The stress-strain workloops for both soft and hard footstrikes in comparison to the prediction of the simple harmonic model for a normal footstrike at each speed for a representative subject. Panel A: shows the workloops for soft footstrikes and Panel B shows the workloops for the hard footstrikes and the simple harmonic model.

Table 1:

	Sprinter (n=5)	Recreational (n=9)
Top Speed (m s^{-1})	10.1±0.7	8.4±0.1
Tc (s)	0.10±0.00*	0.12±0.01
F_{avg} (xW_b)	2.21±0.12*	1.90±0.02
Δ COM (cm)	1.96±0.21*	2.21±0.03
Kvert (kN/m)	153±2*	108±3
P_{Fmax}	0.30±0.01*	0.41±0.01
P_{Zmin}	0.43±0.00*	0.46±0.00

Table 1: shows the differences in kinetics from the top speeds of the sprinter and the top speeds of the recreational runners. * indicates a significant difference between the two groups $p < 0.05$.

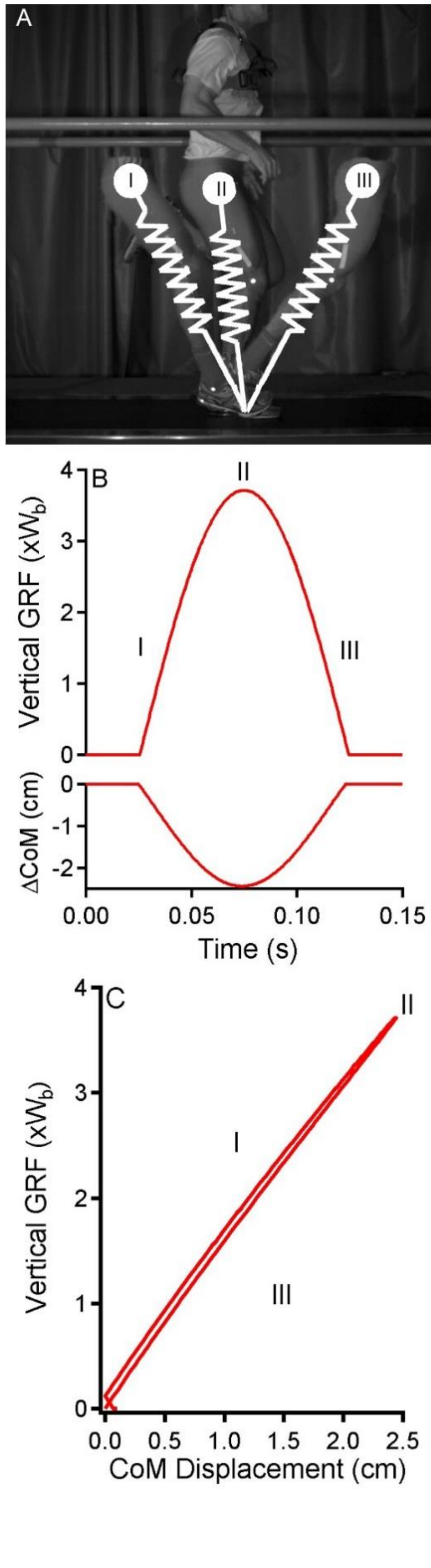
Table 2:

	Hard 3 m s^{-1}	Soft 3 m s^{-1}	Hard 5 m s^{-1}	Soft 5 m s^{-1}	Hard 7 m s^{-1}	Soft 7 m s^{-1}
Tc (s)	0.21±0.00*,†	0.24±0.00*,†	0.15±0.00*,†	0.17±0.00*	0.13±0.00*	0.13±0.00*
Stance Avg. GRF (xW_b)	1.81±0.0*,†	1.46±0.00*	1.92±0.02*,†	1.74±0.006*	1.98±0.03	1.92±0.02
Δ CoM (cm)	6.1±0.2	5.9±0.1	3.6±0.1*,†	4.1±0.1*	2.4±0.0*,†	2.7±0.1*
K_{vert}(kN m^{-1})	39.1±1.2*	33.1±0.9*	68.5±1.8*,†	56.59±1.8*	113.4±4.3*,†	92.5±3.4*
F_{peak} (% Tc)	40.5±0.0*	45.6±0.0*,†	37.1±0.01*,†	48.3±0.004*	34.2±0.01*,†	46.4±0.0*
CoM_{min} (% Tc)	44.2±0.0	45.7±0.0†	45.3±0.0*	47.4±0.0*	44.4±0.0*,†	46.0±0.0*

Table 2: Gait kinetics calculated from different footstrikes at the speeds of 3, 5, and 7 m s^{-1}.

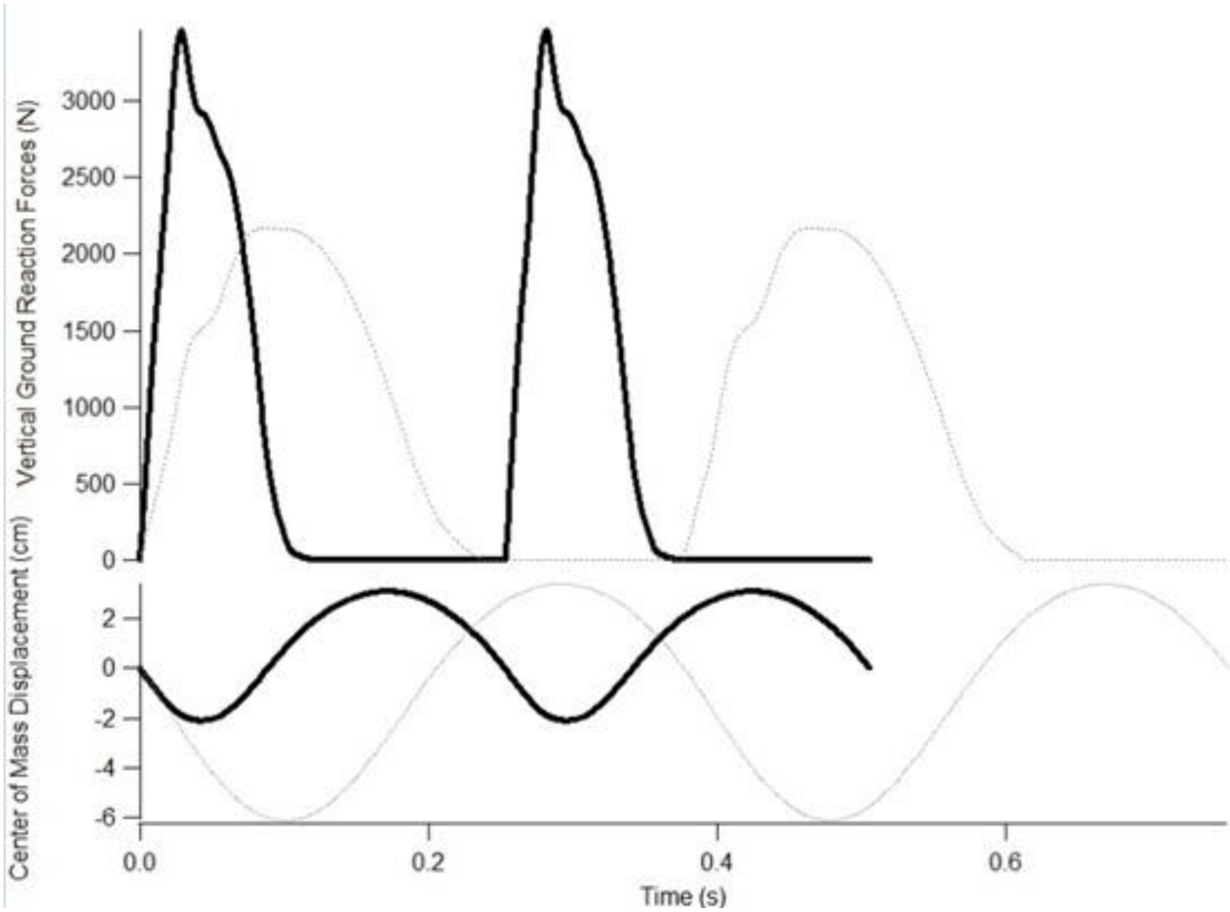
*Significantly different from same speed different foot contact $p<0.05$. †Significantly different from same speed normal foot contact $p<0.05$

Figure 1:



A
I
II
III
B
Vertical GRF (xW$_b$)
ΔCoM (cm)
Time (s)
0.00
0.05
0.10
0.15
C
Vertical GRF (xW$_b$)
CoM Displacement (cm)
0.0
0.5
1.0
1.5
2.0
2.5

Figure 2:



Vertical Ground Reaction Forces (N)
3000
2500
2000
1500
1000
500
0
Center of Mass Displacement (cm)
2
0
-2
-4
-6
0.0
0.2
0.4
0.6
Time (s)

Figure 3:

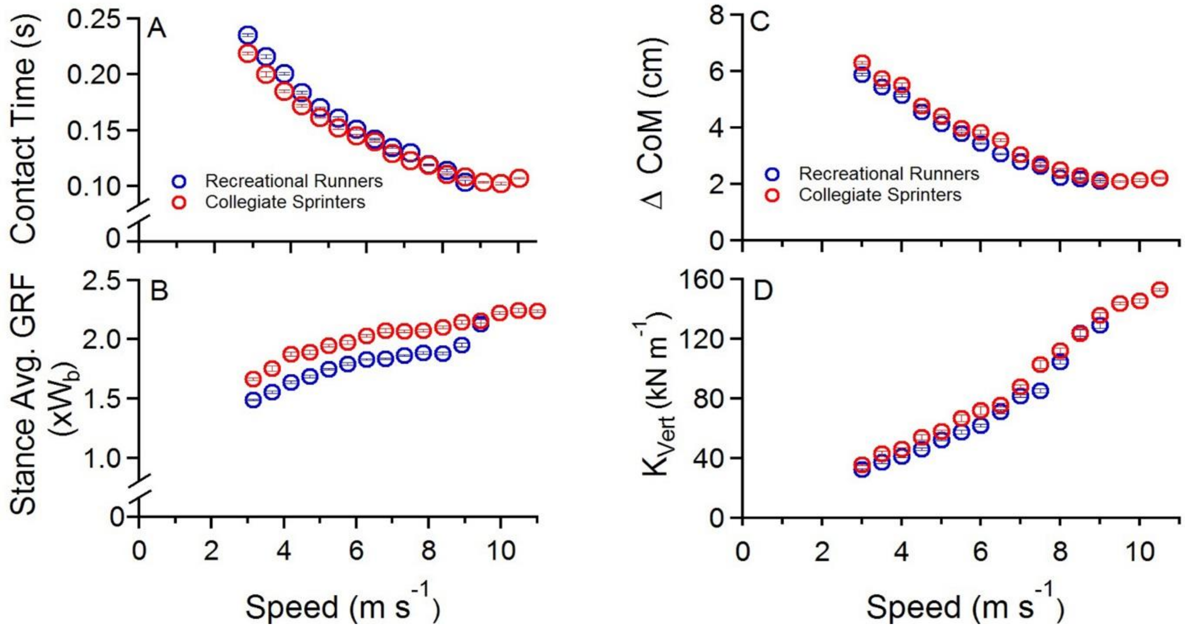
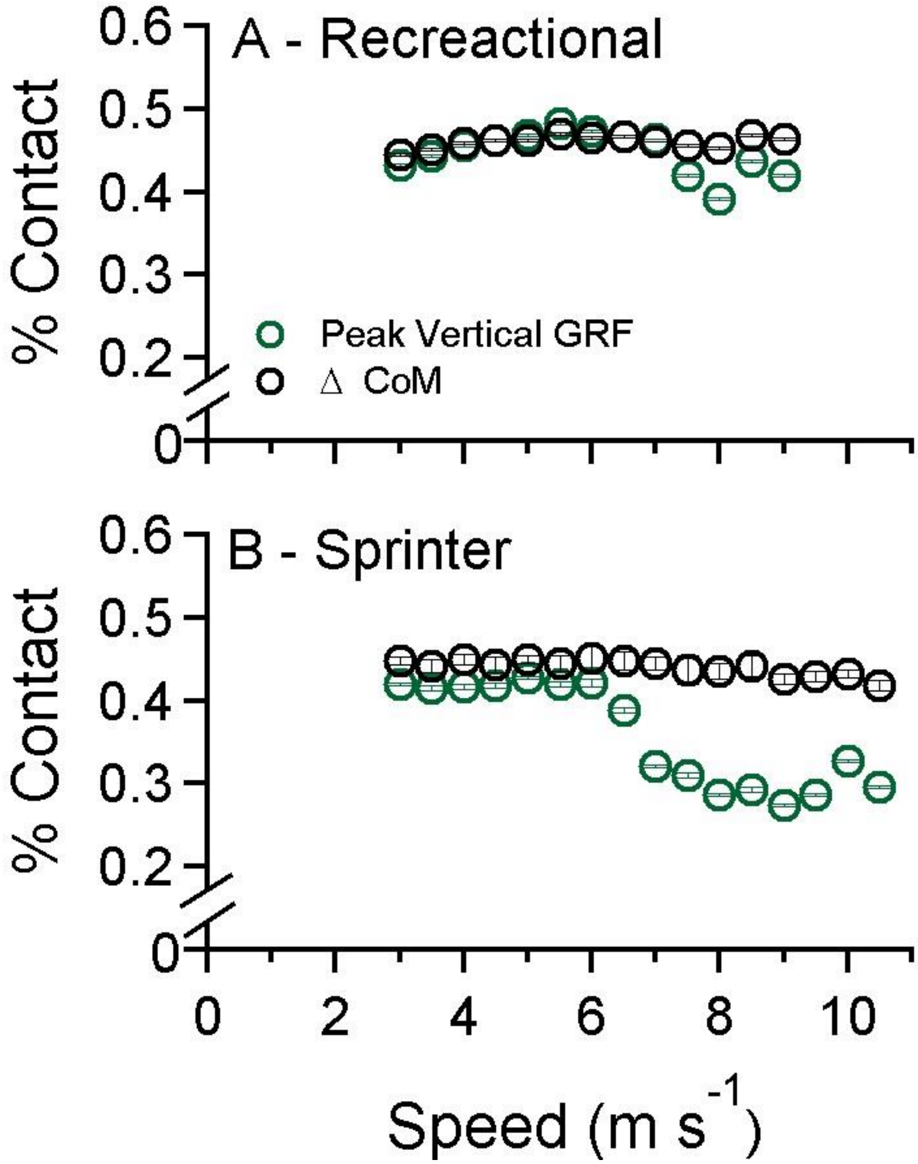
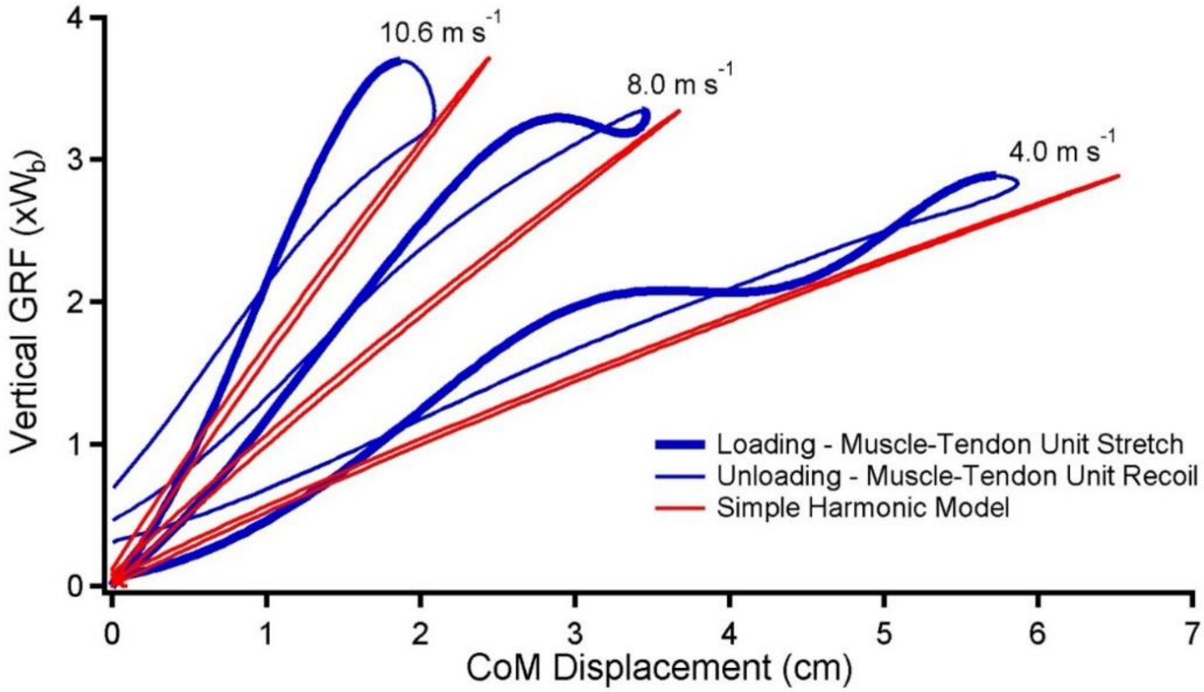

Contact Time (s)
A
Recreational Runners
Collegiate Sprinters
Stance Avg. GRF (xW_b)
B
Speed (m s^{-1})
Δ CoM (cm)
C
Recreational Runners
Collegiate Sprinters
K_{Vert} (kN m^{-1})
D
Speed (m s^{-1})

Figure 4:



A - Recreactional
% Contact
0.6
0.5
0.4
0.3
0.2
0
Peak Vertical GRF
Δ CoM
B - Sprinter
% Contact
0.6
0.5
0.4
0.3
0.2
0
0
2
4
6
8
10
Speed (m s-1)

Figure 5:



10.6 m s-1
8.0 m s-1
4.0 m s-1
Vertical GRF (xWb)
CoM Displacement (cm)
Loading - Muscle-Tendon Unit Stretch
Unloading - Muscle-Tendon Unit Recoil
Simple Harmonic Model
0
1
2
3
4
5
6
7

Figure 6:

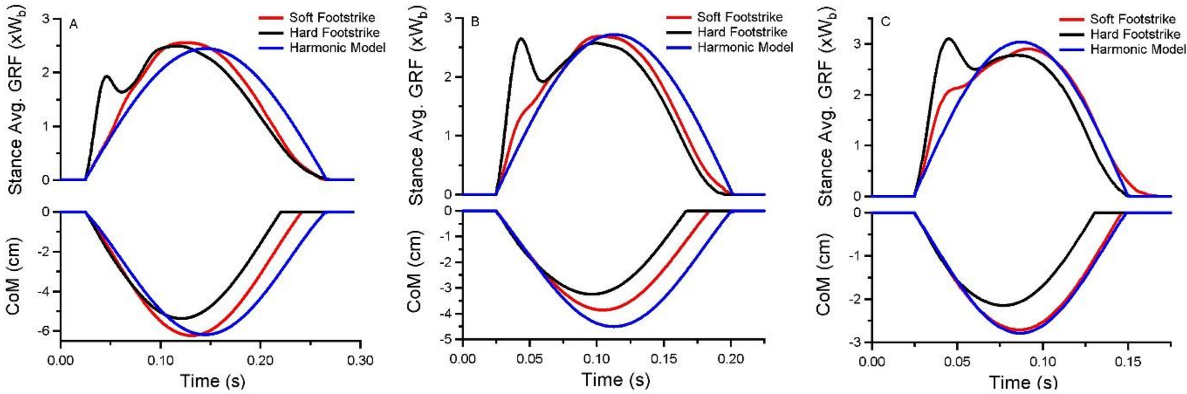

A
B
C
Soft Footstrike
Hard Footstrike
Harmonic Model
Stance Avg. GRF (xW$_b$)
CoM (cm)
Time (s)

Figure 7:

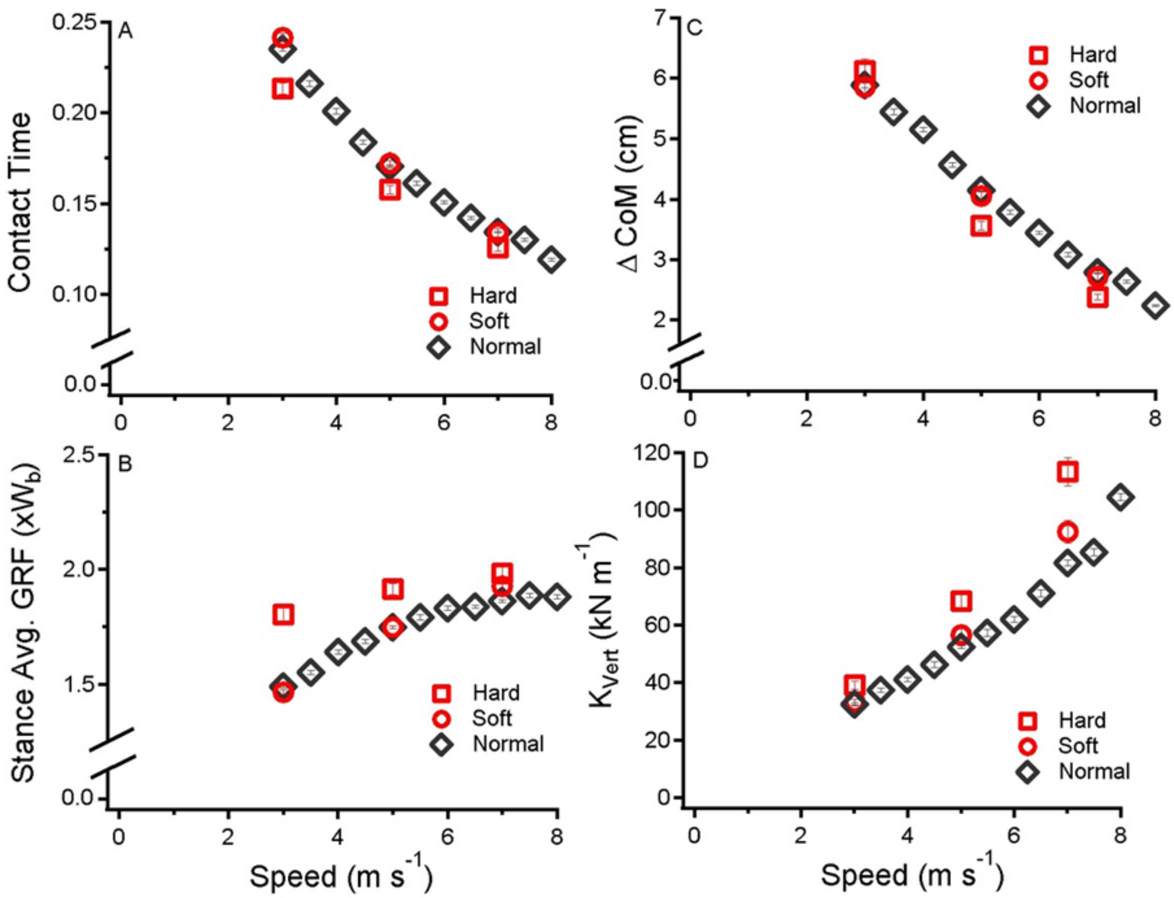
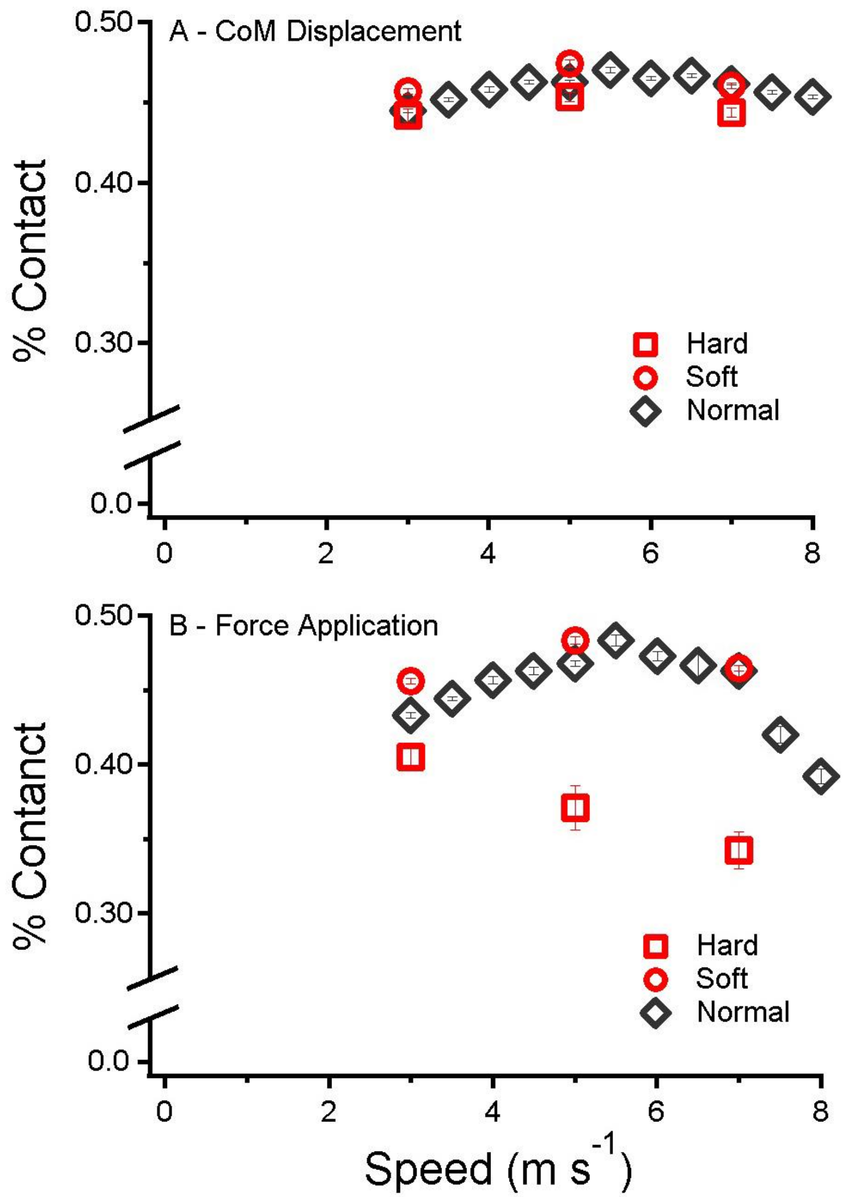
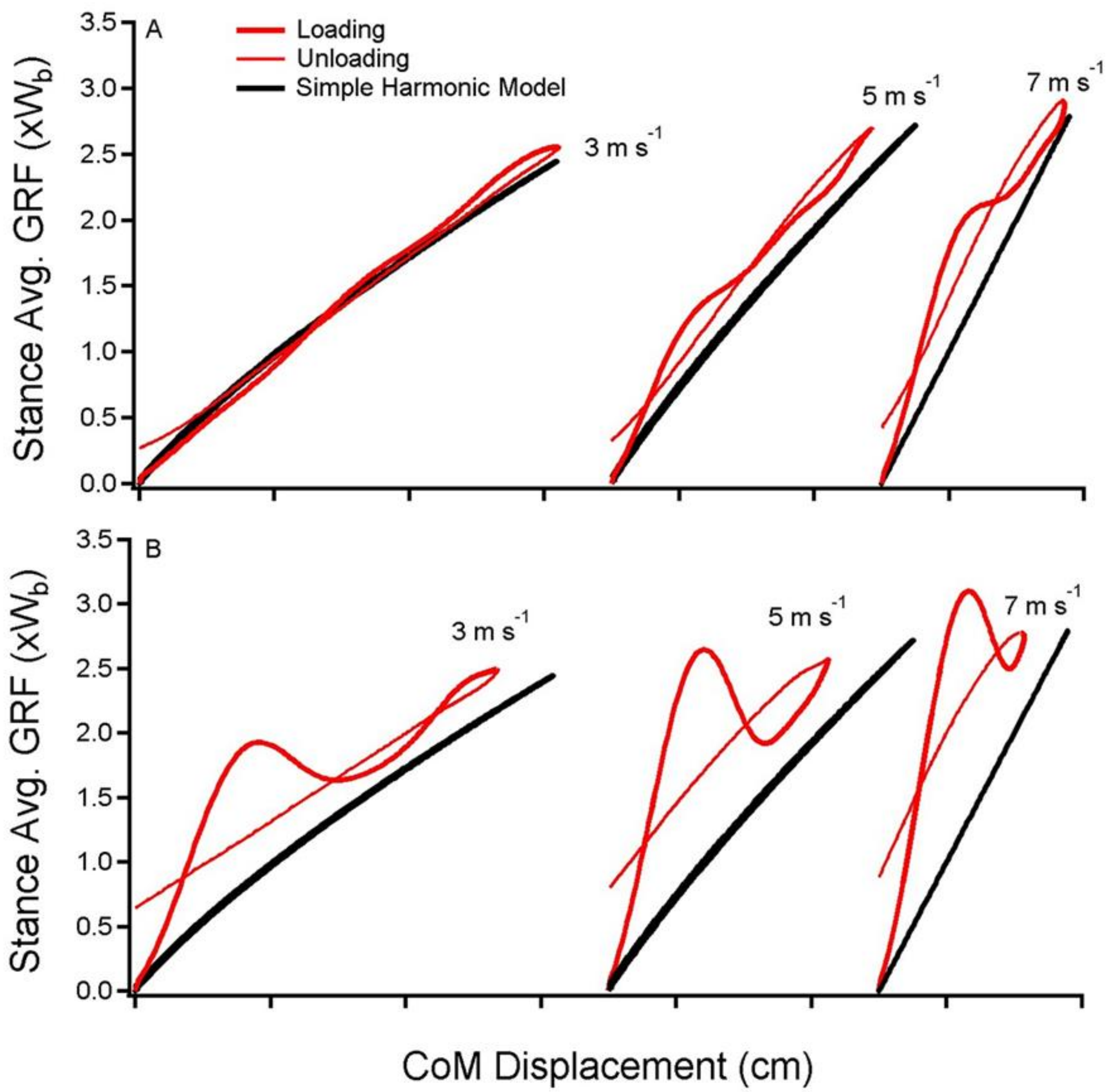

A
Contact Time
0.25
0.20
0.15
0.10
0.0
0
2
4
6
8
Hard
Soft
Normal
C
Δ CoM (cm)
7
6
5
4
3
2
0.0
Hard
Soft
Normal
B
Stance Avg. GRF (xW_b)
2.5
2.0
1.5
0.0
Hard
Soft
Normal
Speed (m s^-1)
D
K_Vert (kN m^-1)
120
100
80
60
40
20
0
Hard
Soft
Normal
Speed (m s^-1)

Figure 8:



0.50
A - CoM Displacement
0.40
% Contact
0.30
Hard
Soft
Normal
0.0
0
2
4
6
8
0.50
B - Force Application
0.40
% Contanct
0.30
Hard
Soft
Normal
0.0
0
2
4
6
8
Speed (m s^{-1})

Figure 9:


A
Loading
Unloading
Simple Harmonic Model
3 m s-1
5 m s-1
7 m s-1
Stance Avg. GRF (xWb)
3.5
3.0
2.5
2.0
1.5
1.0
0.5
0.0
B
3 m s-1
5 m s-1
7 m s-1
Stance Avg. GRF (xWb)
CoM Displacement (cm)