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Testing of high speed ground reaction force treadmill for kinetic analysis for running in static and dynamic loading conditions

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TESTING OF HIGH SPEED GROUND REACTION FORCE TREADMILL FOR KINETIC ANALYSIS FOR RUNNING IN STATIC AND DYNAMIC LOADING CONDITIONS

By

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Abstract

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This thesis proposal describes the design, fabrication, and testing of a high-speed force treadmill capable of measuring the vertical and horizontal ground reaction forces imparted to the bed during the stance phase of human running. Natural frequency, crosstalk, linearity, use of aluminum line shaft with flexible couplings, center of pressures, and comparison to over-ground running were investigated. The natural frequency of the treadmill bed was 113 Hz achieved in part through the use of the line shaft to remove the mass of the motor (36.8 kg) with negligible dissipation of imparted force to the treadmill. The linearity test results showed a high linearity of the force treadmill up to expected forces of human sprinting as well as minimal crosstalk. The center of pressure accuracy was extremely high using measured vertical forces to determine the point of force application. Ground reaction force waveforms in both the horizontal and vertical directions for speeds up to 8.1 ms^{-1} on the high speed force treadmill were nearly identical to the established methods of in ground force plates in both magnitude of force and duration of contact for same speeds by same subject. The high speed force treadmill demonstrated acceptable accuracy for ground reaction forces such that it can be used for a variety of research gait analysis applications up to the limits of human running speed.

Chapter one: Introduction

This research proposal describes the design, fabrication, and testing of a high-speed force treadmill capable of measuring the vertical and horizontal ground reaction forces imparted to the bed during the stance phase of human running. Measurements of ground reaction force permit clinical and research based gait analysis. The development of force treadmills for this purpose has reduced the data-collection time of locomotion experiments, allows for feedback to subjects and/or patients, and has made possible experiments that could not have been done with more traditional equipment (22). The basic premise of kinetic gait analysis is both the horizontal and vertical components of the ground reaction forces are needed to determine the braking, propulsive, and support phases of the stride cycle (7). One of the specific problems in doing this via instrumented treadmill is having a natural frequency high enough to distinguish mechanical vibrations from biological signal (30). Moreover, a design feature of our device was to reach speeds in excess of the record human speed of 12.34m/s, set by Usain Bolt in the 2009 12th International Association of Athletics Federations World Championships in Athletics (16) making it possible to test the true limits of human capability. In order to understand the requirements of the high-speed force treadmill at these speeds the timing and forces of the phases of the gait need to be examined.

The legs of human runners act as springs due storage and release of elastic energy from the musculo-tendinous structures during the gait cycle (5, 10, 13, 21, 26, 27). During foot-contact periods peak forces of ≈3500 N at top speed have been observed (33). At top speed, faster runners apply greater mass-specific ground forces and do so in shorter periods of foot-ground contact (34). Human runners typically require aerial times of 0.12 s or more to attain the minimum swing time of ≈0.350 s generally observed at top speeds (32, 33). Foot-ground contacts in true elites can be as low as 0.085 s (24) and 0.0811 s (33). Thus the world’s fastest runners strike the ground with contacts that are 5 times the body's weight and do this in less than a tenth of a second (33). These mechanics and physiology provide challenges to the design of high speed treadmills, specifically these factors require a treadmill of high stiffness, high natural frequency, and a motor setup able to respond fast enough with torque capable of maintaining a constant speed throughout the gait cycle.

Force treadmill designs must overcome measurement challenges to be used for sprint running and be able to capture forces in all three planes. Existing approaches have included inserting a force platform within a treadmill (15, 18, 23), mounting a treadmill on top of an installed force platform (9, 12, 22), or mounting a treadmill on top of multiple force sensors (2, 11, 14, 19, 25, 29). Force treadmills with an internal force platform or mounted on a force platform measured the vertical component of the ground reaction force but are unable to measure horizontal forces (22). Designs using multiple loads cells are able to measure all three components of ground reaction forces but the horizontal components have generally been of unsatisfactory fidelity (22). Each of these designs has been hampered by bed stiffness, large treadmill mass, and mechanical vibrations induced by the motor or rollers (22). Our design approach consisted of mounting our treadmill bed on top of multiple loads cells and specifically addressing the previous mentioned obstacles as well as the additional necessity of delivering massive instantaneous torques to maintain constant speed throughout the gait cycle.

Inadequate structural stiffness of the bed will cause the natural frequency, intrinsic frequency at which any elastic body tends to oscillate in absence of a driving or dampening force (31), to be too low. When this occurs vibrations within the bed will show up in the ground reaction force waveforms at frequencies similar to the biological signals. When two waves occupy the same space at the same time their amplitudes can sum together meaning that the resultant waveform may have a larger amplitude, constructive interferences, or smaller amplitude, destructive interference, than the biological signal alone(20). Signal filtering is a process that takes input data, sorts the signals, and allows a desired range of signals to pass through while suppressing the rest. With a high natural frequency that is distinct from the biological signal, a signal filtering can be applied to remove unwanted signal artifact, in this range, without any degradation to the biological signal. We set our natural frequency goal to be an order of magnitude higher than the expected biological signals. During top speed sprinting maximum step frequencies of 5 Hz have been recorded with a braking phase and propulsive phase during each foot-contact (33). This creates a signal frequency during each contact making a biological frequency of 10 Hz. Our goal was therefore 100 Hz for the high-speed force treadmill to be able to filter the vibrations of the system from the collected force waveforms.

The mass of the load supported by the transducers is important because with increments in mass the natural frequency decreases, as can be seen from equation (1). This relation also shows that frequency can be increased with a higher stiffness.   Each foot-contact is a collision of two bodies, during this short time period both the foot and the bed are exerting forces on one another (20). The force applied during a foot-ground contact will cause the bed to deflect downwards and during the following aerial phase the bed will sway about its neutral position before either returning to rest or deflect once again from the consecutive foot-contact. The larger the mass that is set into motion, the longer it takes for the oscillations to attenuate and return to equilibrium because of the greater momentum imparted. By reducing the mass supported above the load cells the extent of mechanical interference in the force waveforms will be reduced. To reduce the mass we selected lightweight materials wherever possible such as (1) the aluminum honeycomb pane, (2) hollow aluminum rollers, and (3) removal of the motor from the load cells.

Even with a high natural frequency vibrations induced by moving parts can appear in collected signals making it difficult to acquire force waveforms. Vibrations from moving parts can be present in the force waveforms because (1) sampling periods are short and (2) the load cells are sensitive enough to pick up this movement. Vibrations can occur from loose components and rotational movements that are mounted off axis. . To reduce vibration from movement of the belt, motor, and roller we took care in aligning the roller ends, using machine balanced symmetrical rollers, using bearings built for the rotational speeds the high-speed force treadmill is capable of producing, and tightly adhering parts directly epoxied to the bed (Loctite Hysol 9460 Epoxy, Henkel Corporation, Westlake, Ohio). With stiffness, mass, and vibrations addressed the next concern was selecting a motor that could maintain the high speeds throughout the foot-ground contacts.

The large vertical forces and braking forces of the gait cycle coupled with the short ground-contact times create large impulses during contact phase that significantly increase frictional forces between the belt and the bed compared to the aerial phase. This sudden increase in frictional force has to be matched by the motor to maintain a constant speed requiring the motor to produce massive instantaneous torques. A motor was chosen accordingly and we reduced the frictional load with a silicone lubricant that reduced friction between the belt and the bed. Furthermore, the rapid changes in applied force during the gait cycle mean the motor control has to respond to rapidly to maintain constant speed and is achieved by a dedicated computer that monitors and delivers power to the motor.

Our goal was to build a high-speed force treadmill capable of measuring all 3 components of the ground reaction force during sprinting in order to measure the breaking phase, propulsive phase, center of pressure, and vertical support forces during the gait cycle. To do this the high-speed force treadmill would have to possess an adequate natural frequency that we estimate to be 100Hz. We also wanted to be able to test maximum human speed so designed the treadmill to run in excess of 14.3 ms^{-1}. With this goal achieved the experimental benefits provided by the current device are consecutive footfalls, unlimited data collection intervals, constant speed, ease of motion capture, reduced subject effort due to bypassing acceleration phase of the trial, reduced research effort, reduced analysis time, and ability to perform experiments that are not possible with conventional equipment.

Chapter Two: Methodology

Instrument Design

We wanted to create a force treadmill with the material properties sufficiently high to ensure that the highest frequency components of the ground reaction force are faithfully recorded, generally an order of magnitude higher than the biological signal frequency for this to occur (3). During top speed sprinting maximum step frequencies of 5 Hz have been recorded with a braking phase and propulsive phase during each foot-contact (33). This creates a positive and negative oscillation during each contact making a biological frequency of 10 Hz. Our goal was therefore 100 Hz for the high-speed force treadmill. To do this we approximated the natural frequency using the following (2):

$$(1) \quad Natural\ Frequency\ (Hz) = \frac{1}{2\pi}\sqrt{\frac{P}{\delta m}}$$

Where P is the maximum force we expect to see during running, δ is the vertical deflection resulting from P, and m is the mass of all parts supported by the load cells. We expected peak forces to be under 5 kN, based on that at top speeds sprinters can produce forces 4-5 times their own body weight (37).

We used two simple beam theory equations designed for sandwich structures to estimate the vertical deflection under load from simple beam theory. For a simply supported beam under a central load:

$$(2) \quad \delta = \frac{2k_b P l^3}{E_f t_f h^2 b} + \frac{k_s P l}{bhG_c}$$

where k_b is the beam-bending deflection coefficient with a value of $\frac{1}{48}$, k_s is the beam-shear deflection coefficient with a value of $\frac{1}{4}$, l is the beam length, E_f is the modulus of elasticity of facing skin,

t_f is the thickness of the facing skin, h is the distance between facing skin centers, b is the beam width, and $G_c$is the core shear modulus in the direction of the applied load (1). The length of the bed was chosen to permit excursion of the foot during sprinting while still keeping in mind an increase in length decreases the natural frequency due to increased mass and span. Estimated foot excursion at top speeds from Munro et al (1987) is:

$$(2) \quad D_c = 0.530 + 0.095V$$

Where D_c is the contact distance and V is the running speed (28). A length of 2.0 m allows speeds of 12.34 ms^{-1} to be attained with 0.30 m of extra length allowing the runner to deviate slightly from the exact center of the bed. A width of 0.87m was chosen to accommodate human and non-human runners as well as unconventional modes of exercise such as cross country skiing. The following panel dimensions were chosen to reduce mass while maintaining an adequate stiffness: cell size of 6.35 mm, commercial grade aluminum foil gauge, and facings 3.175 mm thick. These dimensions coupled with equations (1) and (2) provided an estimation of 85 Hz for the panel in the vertical direction. The second estimation for deflection modifies equation (2) with the use of Poisson's ratio and provides an estimation of 91 Hz(4).

We used finite element analysis, a computer application that applies simple physics equations to discrete sections of a system to determine whole system response to an applied force (4,34), to construct a virtual test panel which matched the dimensions of a human foot (63.4mm by 77.1mm) to model for compression failure and dimpling. This gave us an approximation of how many multiples of the intended load the structure can support before failure. We determined the safety factor for compression for the majority of a single foot fall was 3 with a minimum safety factor of 1.58. For dimpling of a single cell the majority had a safety factory of 15 with the minimum safety factor of 7.56 in discrete locations. Under the maximum expected force, failure should not occur.

A large mass reduction was accomplished through the use a 43 cm aluminum line shaft with flexible couplings on either end (R+W Coupling Technology, ZA 150, Bensenville, IL) to remove the weight of the motor from the force treadmill.   The belt is driven by a brushless servomotor (Baldor Electric Company, BSM100C-6150-AA, Fort Smith, AR), capable of speeds in excess of 14.3 ms^{-1}, has a continuous stall torque of 30 Nm , a peak torque of 90 Nm, and a mass of 36.8 kg. The purpose of the coupling was to allow the motor to remain stationary and allow the bed to move freely so that the motor will not dissipate the measured forces. This couple allows for 5.8 mm of parallel misalignment with no resistive forces. The stiffness of the load cells according to manufacturer testing were 2.0 x 10^7 N/m meaning if a load of 5000 N were to be centered on one load cell the load cell would only deflect 0.017mm, well within the tolerance of the line shaft.

During a foot-contact a force was applied to the treadmill which is measured by the strain gauges of the load cells under each corner of the bed.   These readings were relayed to the digital amplifiers (MiniAmp MSA-6, Advanced Mechanical Technology Incorporated, Watertown, MA). From here the data passed through the digitizer (Digidata 1322A, Axon Instruments Inc., Union City, CA) before reaching the dedicated computer. We created digital filters that were applied post hoc (Igor Pro 6.34A, WaveMetrics Inc., Lake Oswego, OR). The filtered data was then converted to Newtons from millitvolts and center of pressure was calculated

Around the treadmill was a safety platform that was in no contact with the force treadmill. It provided decking to the side of the ITM to allow the subject to straddle the bed, handrails so the subject could lower himself onto the belt, and a suspended harness in case of a fall.

Instrument Testing

Flexible drive shaft

The motor is supported by its own bracket that is permanently epoxied to the floor. To test if force was transmitted to the motor support via the flexible line shaft a force plate was placed under the motor support before affixing the bracket to the ground and performed a 9 point static calibration test with forces ranging from 198 N to 1778 N by placing known masses on the center of the bed. Force data was sampled at 1000 Hz. The amount of force transferred to the force plate is reported as a mean percentage and standard deviation of the total applied load.

Natural frequency

The natural frequency of the force treadmill was estimated by observing the impulsive response of the vertical component of the GRF created by dropping a wooden ball on the bed from a height of one meter (3.8 cm, 18.86 g). A fast Fourier transform was used to calculate the first harmonic of the natural frequency of the force treadmill.

Crosstalk

Forces were measured with a 16 point static calibration of forces ranging from 245 N to 2164 N centered on the bed. Measured horizontal forces were compared to measured vertical forces. The mechanical crosstalk was determined along the x and y axis as a percentage by $(F_x \times F_v^{-1}) \times 100$ and $(F_y \times F_v^{-1}) \times 100$, respectively, where F_x is horizontal component of the ground reaction force in the x-direction, F_y is the horizontal component of the ground reaction force in the y-direction, and F_v is the vertical component of the ground reaction force.

Center of pressure

Center of pressure tests were performed with the belt stationary by stacking dead weight (20.25 kg*)* on top of a stylus (0.4 cm diameter) at 65 known locations on the high-speed force treadmill (11). A grid of 13 rows by 5 columns spaced ≈ 15 cm was centered on the bed. Locations of the grid were determined by digitizing (Matlab R2010bSP2, MathWorks, Natick, MA) four views of the treadmill bed through methods described in by Hedrick (17).

The location of the center of pressure in the x and y direction was calculated as (11)

$$(5) \quad CoP_x = \frac{B(F_{v1} + F_{v4})}{\sum_1^4 F_v}$$

and

$$(6) \quad CoP_y = \frac{L(F_{v1} + F_{v2})}{\sum_1^4 F_v}$$

where B was the distance between the left and right transducers (0.7684 m), L was the distance between the rear and front transducers (1.8984 m), F_v is the vertical force and the number denotes the location of the load cell (shown in figure 1), and $\sum_1^4 F_v$ is the sum of F_v from all transducers. The mean absolute error of the calculated center of pressure from the known location was determined.

Comparison to over-ground running

A young healthy male sprinter provided written informed consent in accordance with the guidelines of the University of Montana. Top speed was determined via a progressive, discontinuous test to failure as follows. After a brief warm-up, the test began at 3.0 ms^{-1}. The speed for successive trials increased 0.5-1.0 ms^{-1} at slower speeds and 0.2ms^{-1} at faster speeds. For each trial the treadmill belt was set at desired trial speed while the subject straddled the moving belt. Once the belt reached the desired speed, the subject transferred his weight onto the moving belt using the handrails adjacent to the treadmill. The subject was allowed to take as many handrail-assisted steps as he chose prior to release. Trials were considered successful if a minimum of eight steps were completed without backward drift after full release of the handrails. The subject was encouraged to take as much rest as necessary to fully recover between bouts of sprints (32-34).

The maximum over-ground speed was determined over the last 3 meters of a 42-m straightaway to allow for full acceleration to the maximum sprinting speed before the timing zone. The time of voltage pulses corresponding to the interruption of infrared photocell beams positioned at 39- and 42-m marks at a height of 1.15 m was recorded at 2 000 Hz by a data-acquisition system (6). Within this 3-m distance two force plates (4060, Bertec Corporation, Columbus, OH) were placed in ground on which the subject made a full contact on one plate. Run speed was determined by dividing the time elapsed between beam interruptions by the 3-m distance separating the photocells. The subject was instructed to take the time necessary for full recovery between attempts. Sprint trials will be repeated until subject believed he attained his fastest possible burst speed. The fastest speed recorded was considered the maximum over-ground speed.

Ground reaction force waveforms from the treadmill were compared to the established method of an in-ground force plate for the same speed in the vertical and horizontal planes.

Chapter Three: Results

Flexible drive shaft. The mean vertical force transferred from the bed to the motor mount was 0.15% ± 0.079% of the applied static load (Figure 2B). The greatest transfer of force was 2.10 N when a static load of 1569 N was applied to the bed.

Natural frequency. The natural frequency of the force treadmill in the vertical direction is 113 Hz.

Crosstalk. The crosstalk in the x-direction is 0.20% ± 0.10% and in the y-direction is 0.16% ± 0.034%. The maximum force detected in the x-direction was 2.97 N and in the y-direction was 3.32 N.

Center of pressure. The average error for the center of pressure is 2.49 mm ± 1.71 mm in the x-direction and 2.83 mm ± 1.48 mm in the y-direction.

Comparison to over-ground running. The vertical forces and horizontal forces in the y direction during running are similar for the same speed (Figure 3).

Chapter Four: Discussion

The force treadmill presented in this study has proven to be highly accurate in measuring ground reaction forces and point of force application. Furthermore, the measurement of ground reaction forces during sprinting on the force treadmill matches the established method of in ground force plates. According to Biewener and Full there are seven important criteria to the design of a force platform: independent measurement of force in three orthogonal planes, low cross-talk between force components, high frequency response, linear response over a sufficient range and sensitivity of force measurement, uniform response over the platform's surface, resolution of the point of application of the ground reaction force, and proper dimensions of the platform's surface in relation to the animal to be studied (3). These aspects of force plates led the design of this force treadmill and testing has shown satisfactory results of all criteria.

This type of device improves research capacity for gait analysis and human performance. With the range of speeds capable of this device, true top performance of human sprinters can be measured with great accuracy. This coupled with the ability to measure an unlimited number of strides may unveil valuable information. Many obstacles were overcome to be able to measure top speeds, but this has opened the door for a wide range of studies of performance.

Our estimations for a desired natural frequency of 100 Hz may have been in excess of what is necessary for accurate measurement. The fastest events during a foot contact while sprinting are around 20 ms during the impact peak (8). This is equivalent to 12.5 Hz sine wave, a sine wave whose peak is at 20 ms, where a fast Fourier with terms to 75 Hz reproduces the data to a coefficient of determination of 0.9942. Meaning a treadmill with a natural frequency of 75 Hz would be adequate for measuring ground reaction forces of human sprinters while the treadmill presented has a natural frequency of 120 Hz. This higher natural frequency is likely due to using estimations for a simply supported beam rather than a fixed beam. Length may have been increased or height decreased to bring the natural frequency closer the minimum requirements, but with limited literature on honeycomb panels for this type of use we felt comfortable overestimating.

If this project were to be performed again greater care would be taken in running data cables and power cables due to the amount of interference that can occur. These obstacles were overcome in this design, but can be avoided by proper shielding and separation of cables. This device is subject to damage over time due to moving parts and mechanical properties will need to be monitored.

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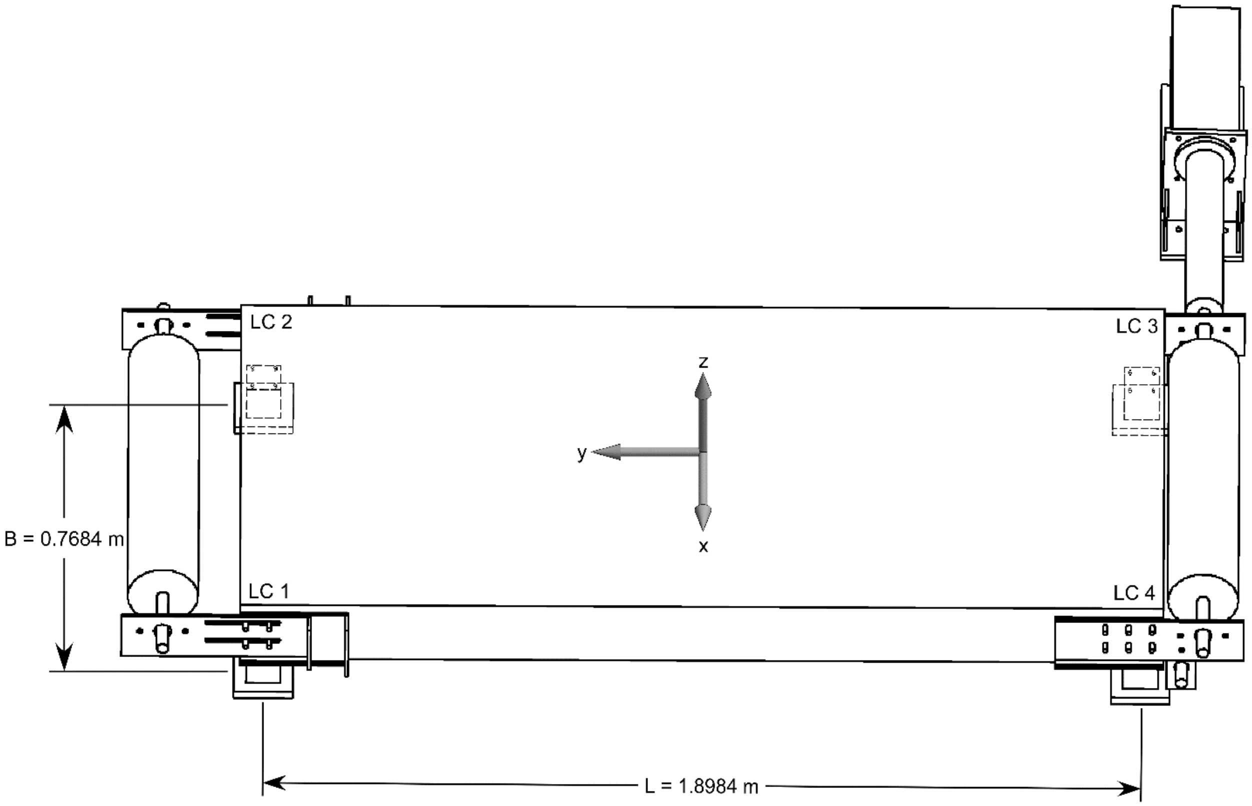

Figure 1. Force treadmill with safety decking. Axis at center shows orientation of force waveform post signal analysis. Load cell (LC) are labeled for identification and calculations. Safety frame and suspended harness not shown.

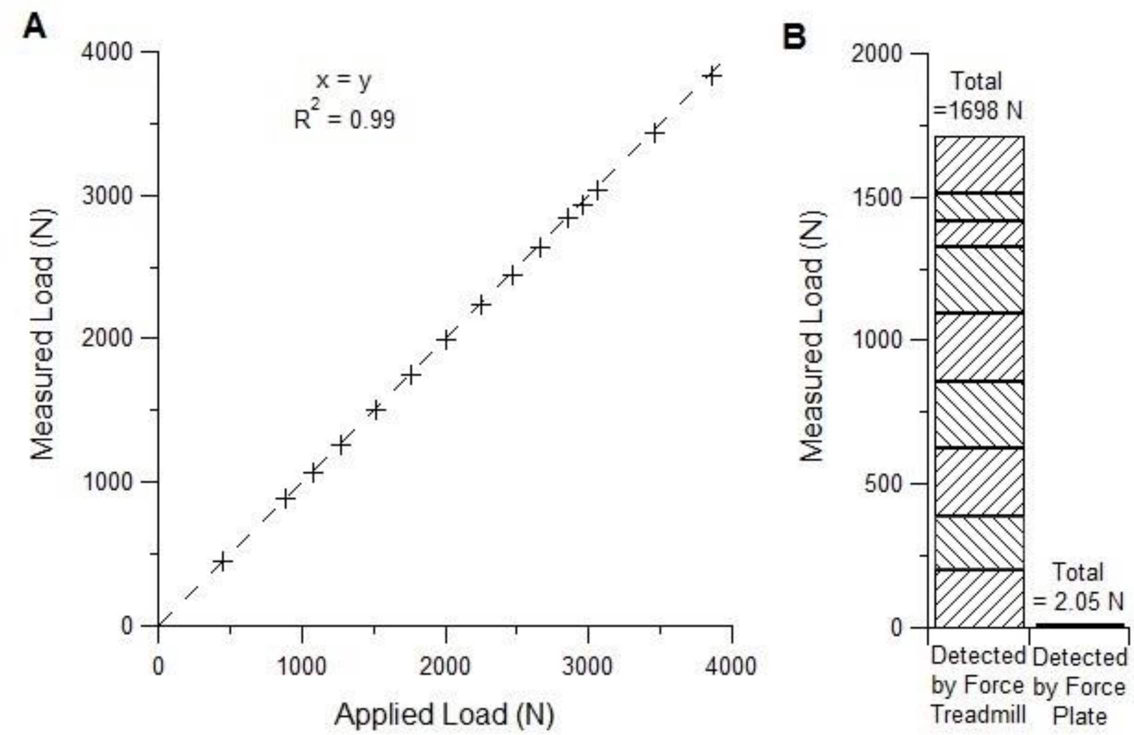

Figure 2. A 15 point calibration test of loads ranging from 451 N- 3857 N were placed on the center of the bed (A). The coefficient of determination was determined to compare the line of identity to measured values. A nine point statically applied load of forces ranging from 198 N to 1 778 N was applied to the treadmill with a force plate under motor mount to measure force transferred from the treadmill to the motor mount via the flexible line shaft (B).

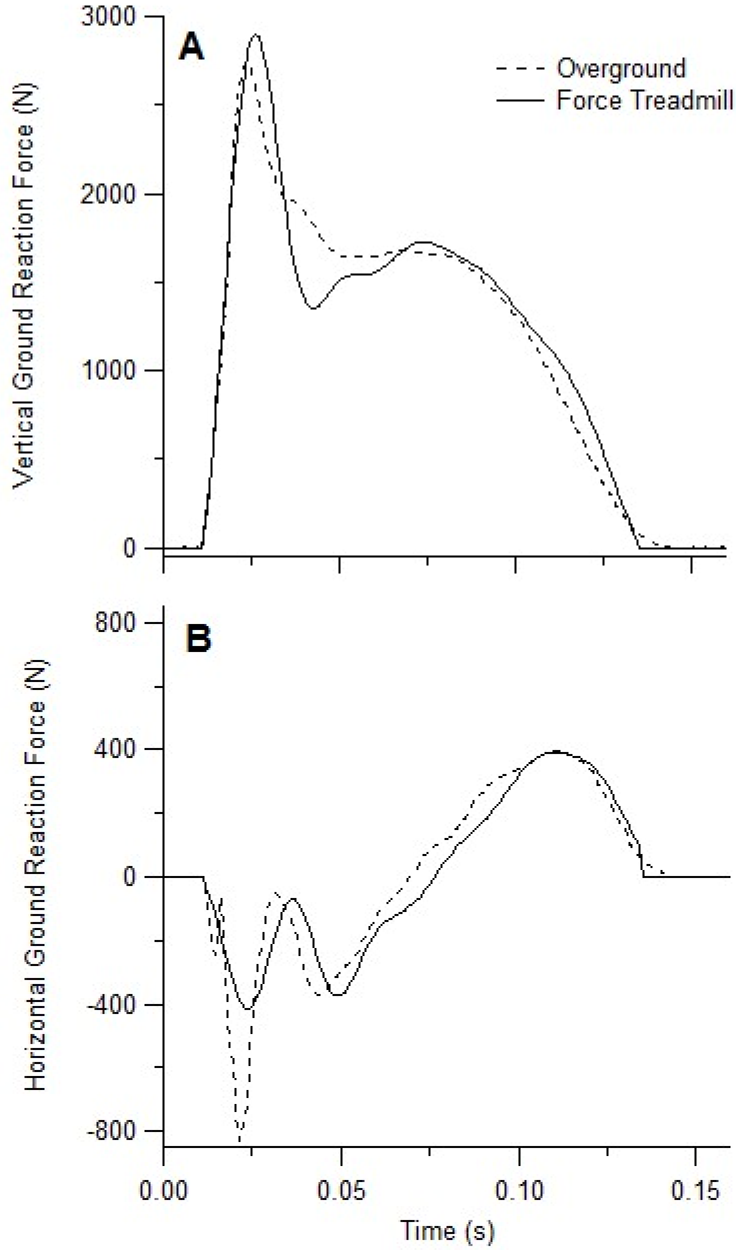

Figure 3. Comparison of vertical (A) and horizontal (B) ground reaction forces obtained for a subject (74.9kg) running at 8.1 ms^{-1} over ground and on force treadmill. Force treadmill signals were filtered with 6 pole Butterworth filter with an effective filter frequency of 40 Hz.