A NOVEL APPROACH TO MONITOR THE GROUND REACTION FORCE

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Custom developed spikes have been used to monitor the vertical GRF temporally resolved for body weight, walking, and jumping on a synthetic surface. All the data is derived from the time dependent voltage variations of piezo spikes, which are observed with the aid of a computer controlled transient recorder. The detection is obtained on two piezoelectric spikes of identical size and shape placed under heel and forefoot. Monitoring is performed on ten healthy athletes with age 19 ± 4 years, mass of (57 ± 6) kg, and a BMI of 21 ± 3. The force resolution is ± 0.25 N and the temporal resolution is 0.01 ms. The contact time, take off time, impact force, active force phases of heel and forefoot have been quantified. The system is of compact size and battery driven, and allows for monitoring of on-field GRF sporting activities.

KEY WORDS: Ground reaction force, piezo-spike for GRF monitoring, vertical component of GRF, piezo-electric sensors, point specific ground reaction force.

INTRODUCTION: Monitoring of the vertical component of the ground reaction force (GRF) is essential in sports and general physical activities. In order to accelerate, decelerate, and rest the body, movements by all living beings require a ground reaction force (GRF). When we step on the ground we produce a force vector that is primarily downwards directed. The ground reacts with the opposing force vector that is primarily upwards directed, addressed as the ground reaction force (GRF). Monitoring of GRF is of importance in the competitive field of sports, general physical activity, and rehabilitation. GRF determine the jumping height, running speed, kicking, and throwing speed of ball, which are essential to most sports. In jogging and running, there is an aerial phase, a time when the limbs are not touching the ground. Aside from wind resistance and gravity, there are no external forces acting on the body during the aerial phase. Therefore, it is the stance phase (the time when a limb is in contact with the ground) of running, jogging, walking, and jumping that must be modified in order to change speed. In order to move faster, stance time must be shorter. Shortening stance time, however, gives less time to produce an impulse, so the peak forces must be higher. We can measure the forces involved in the stance phase of walking, jogging, running, and jumping. Forces of these physical activities are usually measured with a force platform, in-shoe pressure sensors, accelerometer sensors (Adrian & Cooper, 1995, Alexander, 1992, Rose & Gamble, 1996), micro-electro-mechanical system (MEMS), and capacitive pressure sensors (Pritchard & Mahfouz 2011). Our approach is to monitor the reaction force on the contact point of the foot rather than a contact surface or indirect calculation of GRF from the measured acceleration.
Piezo-electric sensors, cut to the proper size and mounted in the spikes were produced for applications, such as beepers. They sell at a price of about $ 0.5 each and were designed to operate without saturation in the high-pressure environment (www.conrad.de). They are manufactured from a ‘‘hard’’ lead zirconate titanate ($PbZr_xTi_{1-x}O_3$, or PZT) ceramic as round discs each with an impedance of 200 Ω at the resonance frequency of 2900 (± 500) Hz, a diameter of 35 mm and a thickness of the PZT layer of 0.58 mm, mounted on a 0.3 mm metal disc (Type: EPZ-35MS29). The output voltage of a piezo element is given by $V = Q/C$, where C is the capacitance of the element, $Q = d_{33}F$ is the charge induced by a force F applied in a direction perpendicular to the electrode surfaces, and d33 is the relevant piezo coefficient (340×10^{-12} C/N) for the actual material. Each piezo-element had a capacitance C = 30 nF and generated an output voltage of 174 mV per Newton. The output signal from the sensors on the assembled spikes was monitored by a computer controlled transient recorder. The piezo-sensor generates a charge proportional to the force on the piezo spike. This leads to a current and voltage, both defined by the resistive shunt from the 1 MOhm input impedance of the preamplifier.

METHOD: Ten healthy athletes with age of 19 ± 4 years, mass of (57 ± 6) kg and a BMI of 21 ± 3 volunteered for this study. Athletes wore pairs of shoes (GM cricket shoe®, Size: UK-11, about € 46, 10 spikes) and had free access to all kinds of movements. Athletes warmed up by stretching and exercising prior to testing. Test movements were practiced by the athletes several times. Data was recorded five times for each subject for body weight, walking, and jumping respectively. All monitoring was performed on the same synthetic surface. Identical shoes were used by all subjects. Two of ten spikes were replaced by the custom developed piezo-spikes. One was placed underneath the posterior lateral side of the heel and the other under the top toe, on the left shoe. Monitoring was performed with hardware and sensors as described in figure 1.

RESULTS: When the monitored athlete stood on one foot with all weight placed on the heel, the sensor at the heel registered a force of about (561.7 ± 0.2) N, equal to the body weight as shown in figure 3. The GRF then reached to the maximum value of (754.5 ± 0.2) N, while pushing off, finally drops to zero, when stepping-off the foot from ground. No force is registered on front spike because it gets little or no contact on the ground during standing. The height of the center of mass (CM) varies due to the variation of the knee and ankle angle during standing on single leg. So, the force doesn't remain equal to Mg. Instead, the force decreases with lowering CG and increases with rising of the CG, as observed in the active phase of single leg stanch (figure 3, right).

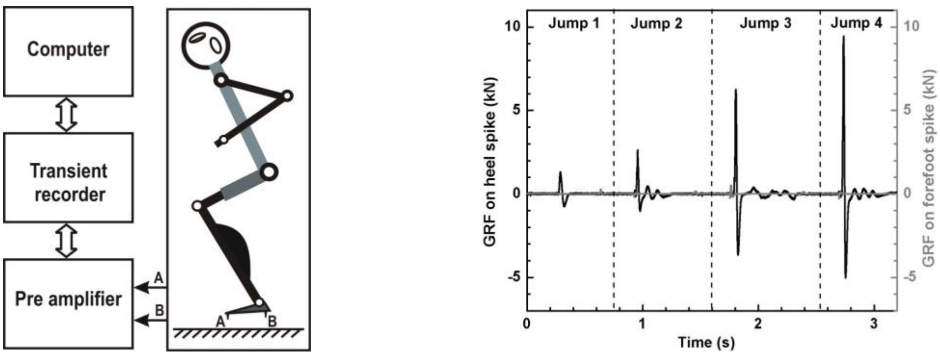

Figure 1: Schematic diagram of the data acquisition procedure (left) for on-field monitoring of the ground reaction force. A: forefoot spike, B: heel spike sensor. Right: Graph of the obtained data for four consecutive vertical jumps.

Alternatively, if one stands on the spike starting with the crouching position and then stands up straight, a different result is obtained as demonstrated in figure 2. The CM starts with zero speed, accelerates to finite speed, and then decelerates to a new resting position. During this alteration, the spike registers a force F given by F – Mg = Ma, where "a" is the acceleration of the CM vertically upward. Figure 1 (right) shows the wave form observed for four consecutive vertical jumps, the jump height is consecutively increased by the athlete. In the 4th jump, displayed magnified in figure 2, when jumping off the ground and landing on the ground surface from a height of a few cm, the force rises rapidly to a value of (9.31 ± 0.25) kN, significantly larger than Mg.

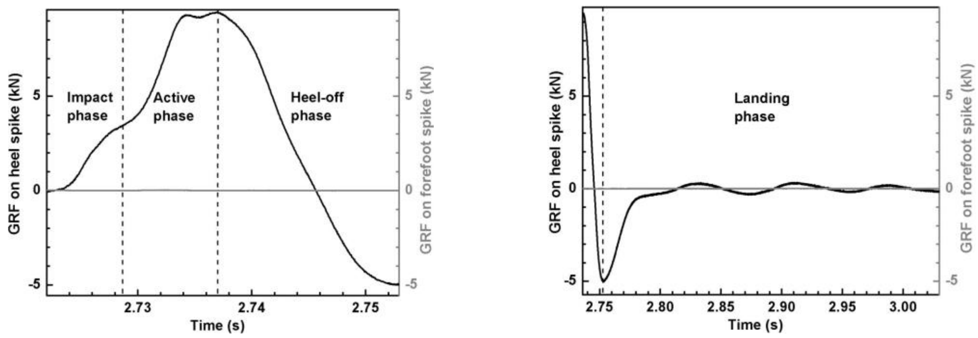

Figure 2: The left graph represents a magnified view of the fourth jump with the illustration of the first three phases; the right graph represent a magnified view of the landing phase of the same jump.

The magnitude of the impact force can be reduced by allowing the knees to bend more on contact or increased by keeping the legs straight. The CM has a negative velocity at contact, decreasing rapidly to zero with a slight positive velocity and an overshoot due to flexure of the knees. The initial acceleration is therefore large and positive in a direction vertically upward with F > Mg initially. This is a good example for a case where a deceleration in one direction can usefully be interpreted as acceleration in the opposite direction. The magnitude of the force can be calculated, using estimates of the initial velocity and the time taken to come to rest. The total time required for jumping from heel-strike to heel-off, the duration of the impact phase, the active phase, and the heel-off phase are presented in table 1. The temporal characteristics of body weight and walking are presented in this table too. The GRF observed during heel-strike to heel-off, impact phase, rate of change of force during active phase, and change of force during heel-off phase are presented in table 2 for body weight, walking, & jumping respectively.

Table 1
The temporal quantification of different phases observed for the monitoring of own body weight, walking, and jumping.

Events	Heel-strike to -off (ms)	Impact phase (ms)	Active phase (ms)	Heel-off phase (ms)
Body weight	173.22 ± 0.01	31.38 ± 0.01	82.70 ± 0.01	59.12 ± 0.01
Walking	129.38 ± 0.01	34.81 ± 0.01	37.88 ± 0.01	50.88 ± 0.01
Jumping	29.92 ± 0.01	6.08 ± 0.01	7.72 ± 0.01	15.92 ± 0.01

Table 2
The spatial quantification of different phases observed for the monitoring of own body weight, walking, and jumping.

Events	Heel-strike to -off (N)	Impact phase (N)	Active phase (kN/s)	Heel-off phase (N)
Body weight	(561.8. to 745.5) ± 0.3	561.8 ± 0.3	0.361 ± 0.006	740.8 to 86.1) ± 0.3
Walking	(805.4 to 2903.4) ± 0.3	805.4 ± 0.3	1.283 ± 0.007	(2899.4 to -1286.7) ± 0.3
Jumping	(1020.4 to 2737.0) ± 0.3	1020.4 ± 0.3	5.372 ± 0.005	(9465.1 to -5005.8) ± 0.3

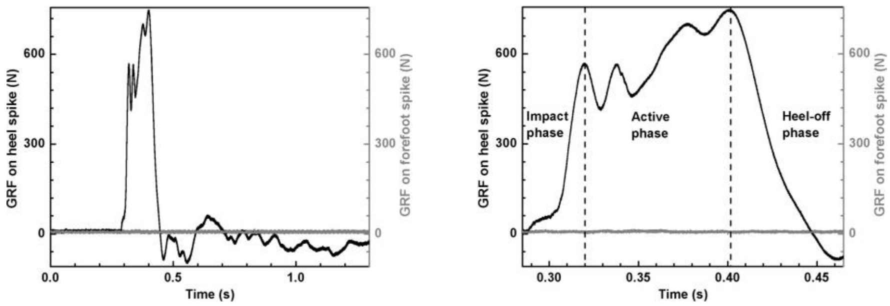

Figure 3: The left graph represents the GRF data that are observed due to the own bodyweight on the sensor spikes, data recorded at standing position on single leg. The right graph is the magnified view of the initial phases of the same observation.

The response corresponding to after landing induced vibrations is observed for jumping (figure 2, right). No vibration phase is observed during walking (figure 4, left). The dip in the middle of the force wave form is due to the centripetal force associated with motion of the center of mass along a curved path. The second peak is found to be higher than the first when walking at a fast pace, but a satisfactory explanation of this effect could so far not be identified in the literature.

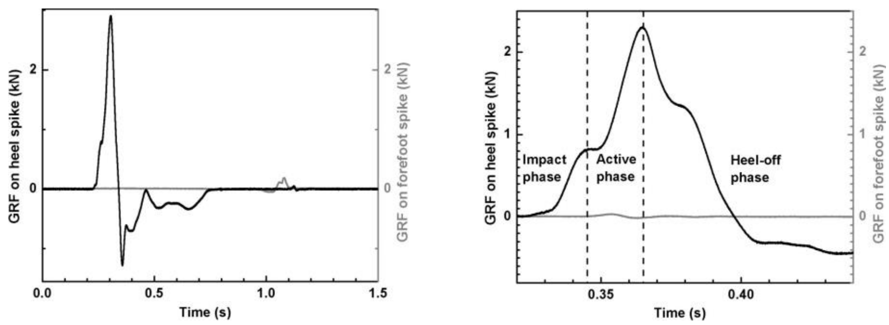

Figure 4: Displayed (left) is the data observed for a full cycle of walking and the right plot depicts the data from heel-strike to heel-off of this walking cycle.

The vertical components of the ground reaction forces on the heel spike of the left foot while walking and jumping have shown that the force rises from zero to a maximum value of significantly higher than Mg, then raises further in the active phase, and finally drops to zero. The raise in the active phase of the force wave form is due to the force associated with the transformation of body weight to the left leg.

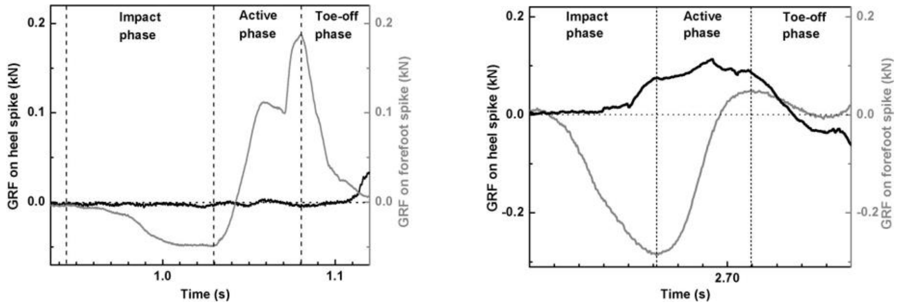

Figure 5: Magnified view of the data observed on forefoot spike for left: walking and right: jumping

In figure 5 the observed waveform for the vertical component of GRF on the forefoot foot spike shows an initial drop of force below Mg because of the lowering of CG at the toe contact to the ground as the knee straightened force raise and finally due to body weight transfer, the active foot force raises further and finally drops to zero. The forefoot spike, placed on to the very front of the shoe, reaches ground contact at the toe contact phase of walking and during the landing phase of jumping. That is why the value of GRF observed on the forefoot spike is insignificant. The observed temporal quantifications and GRF on the forefoot spike for walking and jumping are listed in table 3 and 4 respectively.

Table 3
The temporal quantification of different phases observed for the monitoring of walking and jumping.

Events	Toe-strike to -off (ms)	Impact phase (ms)	Active phase (ms)	Toe-off phase (ms)
Walking	850.16 ± 0.01	34.81 ± 0.01	37.88 ± 0.01	38.82 ± 0.01
Jumping	12.16 ± 0.01	4.92 ± 0.01	3.81 ± 0.01	3.52 ± 0.01

Table 4
The spatial quantification of different phases observed for the monitoring of walking and jumping.

Events	Toe-strike to -off (N)	Impact phase (N)	Active phase (N/s)	Toe-off phase (N)
Walking	(-.49.0 to 187.4) ± 0.3	-49.0 ± 0.3	96.0 ± 0.3	(187.4 to 0.5) ± 0.3
Jumping	(-286.2 to 46.9) ± 0.3	-286.2 ± 0.3	450.2 ± 0.3	(46.9 to 0.4) ± 0.3

CONCLUSION: The developed detection scheme is suitable for on-line monitoring of the ground reaction force on different points underneath the shoe during all kinds of physical activity, which cannot be observed by available monitoring systems. A quantitative analysis of the different stages of physical activity can be achieved. By the developed monitoring scheme also so far unobserved parameters are accessible. These include contact time, take off time, impact force, active force, and the areal phases of heel and forefoot. A so far not reported temporal resolution has been achieved. The developed detection scheme presented in this study can be used for wireless monitoring of ground reaction forces for human movement without impeding natural motion or restricting natural activity. Currently a lightweight assembly with a small battery operated PC is already available for carry-on monitoring of athletes.

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