

Study on Estimation of Peak Ground Reaction Forces using Tibial Accelerations in Running

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Abstract— Ground Reaction Forces (GRF) are exerted by a surface as a reaction to a person standing, walking or running on the ground. In elite and recreational sports, GRFs are measured and studied to facilitate performance improvement and enhance injury management. Although, GRFs can be measured accurately using force platforms, such a hardware can only operate in a constrained laboratory environment and hence may limit and potentially alter a subject's natural walking or running pattern. Alternatively, a system that can measure GRFs in a more natural environment with less constraints can provide valuable insights of how humans move naturally given different gait patterns, terrain conditions and shoe types. In this regard, inertial Micro-Electrical-Mechanical-Sensors (MEMS), such as accelerometers and gyroscopes, are a promising alternative to laboratory constrained data collection systems. Kinematics of various body parts, such as their accelerations and angular velocities, can be quantified by attaching these sensors at points of interest on human body.

In this paper, we investigate the relationship between the vertical GRF peaks measured by an OR6 series AMTI force plate, and accelerations along the tibial axis measured by a MEMS sensor. Our measuring system consists of two low-power wireless inertial units (ViPerform), containing one tri-axis accelerometer placed on the medial tibia of each leg. We investigate the accuracy of the measured and estimated GRF peak in 3 subjects, by means of the Root Mean Square Error (RMSE). The RMSE achieved across the speeds of 6, 9, 12, 15, 18, 21km/h and sprinting were 157 and 151N, 106 and 153N, and 130 and 162N for the left and right legs respectively for Subjects 1, 2, and 3. We achieved normalized errors of 6.1%, 5.9% and 5.4% for all the subjects.

I. INTRODUCTION

Interpretation of human motion and gait patterns is paramount to the understanding of risk of injury in elite or recreational sports. Specifically in running, past studies established the critical role of limb kinematics in risk of injury or rehabilitation of injured players [1]. Gait patterns and running strategies in different terrains across different running speeds are correlated with tibia injury [2]. Gait pattern in running affects the ground reaction force (GRF) acting on the body through the feet and the resulting tibial shock, i.e., the impact force transmitted to the tibia [3].

Traditionally, GRFs in running are measured using force platforms or plates mounted in the ground. Force plates may also provide additional information such as balance, center of

pressure and similar biomechanics parameters for a subject [4]. While force plates accurately measure various features of human locomotion including GRFs, their use is constrained by a fixed laboratory setting, where subjects may not be able to replicate their natural running patterns. In contrast, an ambulatory system that provides accurate measurement of GRF outside of the laboratory setting can produce better insights into the natural running pattern of an individual. In this paper, we introduce ViPerform [5], a fully ambulatory system that uses inertial sensors placed on the tibia and accurately measures the GRFs using tibial accelerations.

A number of previous studies investigated the correlation between accelerations and GRFs when walking or running. Early work by Lafortune et al [6] quantified tibial shock during walking and running using the relationship between the tibia axial acceleration (TAA) and GRFs. Authors reported the first reliable TAA data and suggested a linear relationship between peaks of differentiated GRF and TAA. A later study by Lafortune et al [7], analyzed the GRF and TAA relationship by means of a Fast-Fourier-Transform (FFT). Under the assumption that the body behaved as a linear system, authors re-estimated the TAA acceleration by using only part of the main harmonics of a combined transfer function between GRF peak and TAA. However, reported errors and large phase shifts prevented this process from being reliably implemented [7].

Recent advancements in inertial micro-electrical-mechanical (MEMS) sensors, such as accelerometers and gyroscopes, have led to significant interest in their use for human biomechanics tracking [8], [9]. Due to their small form factor and low-cost, these devices can be comfortably worn on body, thus allowing ambulatory monitoring of daily living activities outside the laboratorial setting. In a recent work, Hunter et al [10] studied the relationship between GRF impulse and sprint velocity in athletes. This work revealed a non-linear relationship between sprint velocity and also vertical and brake GRF impulse. Daoud [11] studied TAA using MEMS sensors from runners with Reverse-Strike (RS) or Heel-Strike (HS) running pattern, while running bare-foot or with shoes. It was reported that overall, lower limb compliance was a significant predictor of higher GRF only in HS runners. A more recent experiment



Fig. 1. Subject wearing two ViPerform sensors on the tibia.

performed by Rowlands et al [12], showed that waist acceleration showed positive correlation with GRF in activities such as walking, jogging, running, jumps and box drops. However, no estimation of GRF was reported by the authors.

Whilst previous studies compare and correlate GRFs with accelerations of different parts of the body, few attempts have been reported that directly estimated these forces based on accelerations. In this paper, we report on the estimation of GRFs using TAA measured by a low-power wireless inertial system (ViPerform). We also investigate the effects of the body mass in this estimation and correlate this in an experimental protocol combining seven different speeds and three subjects. We validate the accuracy of our system in comparison with the gold standard AMTI *OR6 – 6 Series* force plate.

The rest of this paper is arranged as follows: Section II provides an overview of the Viperform system and the hardware used; Section III describes the sensor and GRF data processing flows, the body mass compensation and describes the running protocol adopted by all subjects; Section IV and V provide the experimental results and discussion respectively, while the conclusions are summarized in Section VI.

II. SYSTEM OVERVIEW

A. Hardware

ViPerform consists of two measurement units and a base station. Each unit comprises a **3D** Accelerometer, **3D** Gyroscope and **3D** Magnetometer. In this study, only one low-power **3D** accelerometer (ST Microelectronics *LSM303DLHC*[13]), with a I^2C serial interface digital output and full scale acceleration input of $\pm 24g$ is used. For this application, each unit samples accelerations at 100, 20, 20Hz on the x-, y- and z-axis respectively, and transmits the recordings through one *nRF24AP2* Nordic Semiconductor [14] ANT wireless chip to a base station. Data can be offloaded from the base station for further off-line analysis on a PC.

B. Sensor Placement and Orientation

The two measurement units are placed on the legs along the tibial axis in the mid point between the lower edge of the medial malleolus and the medial joint line of the knee as depicted in Figure 1. This method of placement has previously been reported as a reliable landmark for measurements of tibial accelerations [15], [16]. The base station is placed on the left upper arm. In this paper, the sensor frame of reference ($\{x, y, z\}$ in Figure 2) refers to the 3D frame that moves through space with the sensors. Accelerations along the tibial axis are measured by the x-axis of the accelerometer, whereas accelerations on the y- and z-axis represent measurements of the anterior-posterior and medio-lateral planes respectively.

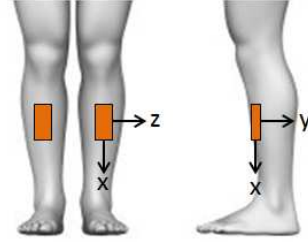


Fig. 2. Placement of ViPerform on the tibia.

III. EXPERIMENTAL METHODOLOGY

A. TAA Feature extraction

A typical TAA is represented in Figure 3 [15]. Four events namely, Heel-Strike (HS), Initial Peak Acceleration (IPA), Maximum Peak (MP) and Peak-to-Peak (Pk2Pk) were correlated with the vertical GRF active peak. Figure 4 shows typical vertical GRF traces from a force plate at three different speeds of $6km/h$, $15km/h$ and sprinting. As the speed increases, the impact peak is reduced, whereas the active peak increases [3].

B. Linear and Logarithmic Approximations and Body mass compensation

Data from peak vertical GRF and TAA at different running speeds are plotted in a scatter plot for analysis. Linear and logarithmic approximations are employed to assess correlation. In addition, the relationship between the body mass and the peak GRF is investigated.

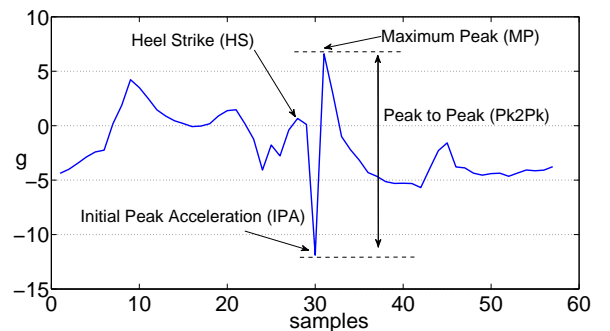


Fig. 3. Typical TAA graph at the speed of $15km/h$.

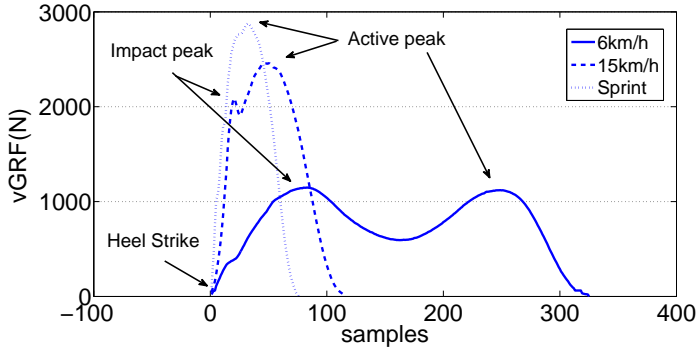


Fig. 4. Typical vertical GRF graph at the speed of 15km/h.

C. Data Collection

All experiments were performed in the Biomechanics Laboratory of Victoria University, Melbourne, Australia. Sensor data was collected from three subjects with no recorded lower limb disorders or running impairments. All subjects gave verbal consent. Experiments consisted of each subject performing a protocol of 42 running trials, divided in blocks of approximate speeds of 6 km/h, 9 km/h, 12 km/h, 15 km/h, 18 km/h, 21 km/h and sprinting. For each trial, each subject was instructed to run through two speed towers at a nominated speed, with one practice run to judge the correct pace. In addition, subjects were instructed to hit the force plate with their left and right legs three times respectively to assess asymmetry variability [15] and wireless link quality between the sensors and the base station.

Video recording was employed to match the relevant acceleration stride with the corresponding force plate data. Each subject was instructed to stand still for ten seconds before each data collection.

D. Force Plate

An AMTI OR6 Series force plate [17] was employed to validate the sensor signal. At the start of the protocol, each subject was instructed to stand still for five seconds to calibrate the recording trigger threshold of the force plate. Data was sampled at 300Hz and recorded without any filtering.

E. Accuracy Assessment

The quality of the GRF estimation was assessed by calculating the Root Mean Square Error (RMSE) of the GRFs recorded by the force plate and those estimated by the ViPerform units. It is defined as:

$$RMSE = \sqrt{\frac{\sum_{i=1}^N (GRF_S(i) - GRF_{FP}(i))^2}{N}} \quad (1)$$

where GRF_S and GRF_{FP} represent the sensor and force plate GRFs and N the number of compared strides.

IV. EXPERIMENTAL RESULTS

A. Data Analysis

Table I shows that the measured walking and running speeds for all subjects were close to the given target. Subjects 1, 2 and

TABLE I
MEASURED RUNNING SPEEDS (KM/H)

Target Speed	Subject#1	Subject#2	Subject#3	Average
6	6.0±0.11	6.1±0.03	6.3±0.11	6.1±0.13
9	9.3±0.04	9.1±0.08	9.0±0.13	9.±0.13
12	13.0±0.07	11.9±0.09	13.0±0.10	12.6±0.62
15	14.9±0.06	15.2±0.04	14.4±0.05	14.8±0.40
18	18.7±0.03	17.3±0.01	18.1±0.04	18.0±0.71
21	22.0±0.01	20.6±0.02	21.0±0.01	21.2±0.74
Sprint	25±0.01	27.1±0.02	25.3±0.04	25.8±1.11

3 walked at 6.0, 6.1 and 6.3km/h with low standard deviation of 0.11, 0.03 and 0.11km/h. Between the running speeds of 9 and 21km/h, the largest error with respect the target speed was 1.0km/h at 12km/h for Subjects 1 and 3. The largest standard deviation was 0.13km/h for Subject 3 at 9km/h.

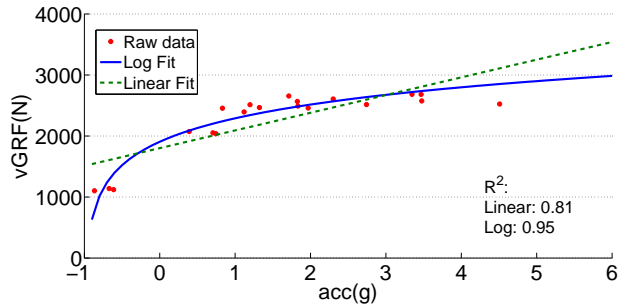


Fig. 5. Linear and Logarithmic approximations of accelerations at Heel Strike (HS) and peak vertical GRF for Subject 1.

Figure 5 shows the logarithmic and linear approximations of Heel-Strike **HS** acceleration points mapped to vertical GRF active peaks. Accelerations between the speeds of 9 and 21km/h ranged from 0.8g to 4.5g, whilst for the walking speed of 6km/h, the averages are mapped from $-0.5g$ and $-0.8g$. The Logarithmic approach (depicted as the solid line in Figure 5) shows a higher correlation (0.95) as compared to the linear approach (0.81). We observed a similar correlation pattern in mapping of **IPA**, **Pk2Pk** and **MP** events to vertical GRF. Therefore, we adopted the logarithmic approach and the linear results are not reported in the rest of this paper. The logarithmic function is defined as:

$$u = \log_2(acc + b) \quad (2)$$

where acc is the acceleration at a specified event. The coefficient b is set to 1 to avoid negative logarithmic numbers.

TABLE II
CORRELATION OF LOGARITHMIC FUNCTION BASED ON TAA FEATURES

Feature	Subject#1	Subject#2	Subject#3
HS	0.88	0.76	0.84
IPA	0.83	0.87	0.88
MP	0.96	0.95	0.96
Pk2Pk	0.91	0.93	0.96

Table II lists the correlation between the log function of Equation 2 and vertical GRFs at **HS**, **IPA**, **Pk2Pk** and **MP** events. For all subjects, highest values of correlation between

GRF and the logarithmic function were found for the **MP** event with correlation values of 0.96, 0.95 and 0.96 for subjects 1, 2, and 3, respectively. The lowest correlations were found at the event **HS** with correlations values of 0.86, 0.76 and 0.84 for the three subjects.

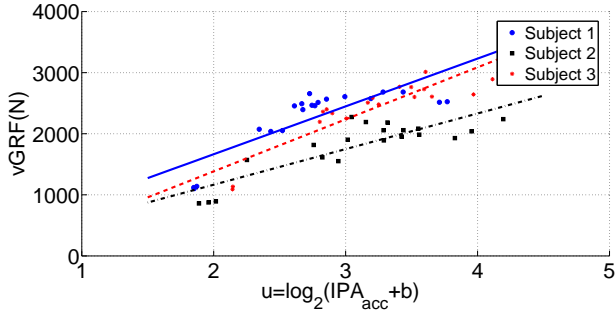


Fig. 6. Logarithmic approximation of the Initial Peak Acceleration (IPA) and peak vertical GRF for Subject 1, 2 and 3.

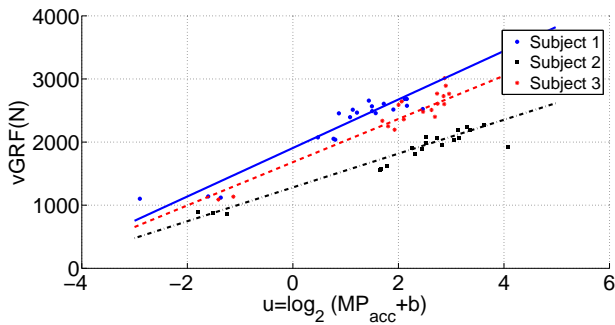


Fig. 7. Logarithmic approximation of the maximum acceleration (MP) peaks and peak vertical GRF for Subject 1, 2 and 3.

Figures 6 and 7 show scatter plots of the logarithmic function for accelerations at events (**IPA**) and (**MP**) respectively, and the corresponding vertical GRF peak for each speed. It can be observed from Figure 6 that the data points are non-uniformly distributed over the logarithmic estimation. Subject 3 (dotted line) showed a higher slope in his results in comparison with Subjects 1 and 2. As the logarithmic function employing the **MP** events shows the best overall correlation with the vertical GRF, we restricted our investigation to this method for the rest of this study.

We investigated the effect of body mass on the logarithmic approximation based on maximum TAA peaks (**MP**). As shown in Figure 7, slopes of the logarithmic approximation are $384.3N/\log(g)$, $267.9N/\log(g)$ and $342.4N/\log(g)$ while offsets are $1906N$, $1281.5N$ and, $1681N$ respectively, for the three subjects. The slope and intercept values for left and right leg GRFs as a function of body mass of each subject are shown in Figure 8. In this figure, slope is represented on the y-axis as coefficient a (left-hand pane) and offset as coefficient c (right-hand pane). The empirical equations 3 and 4 that correlate the body mass with the slope and intercept are defined below as:

$$a(m) = 4.66 * m - 76.6 \quad (3)$$

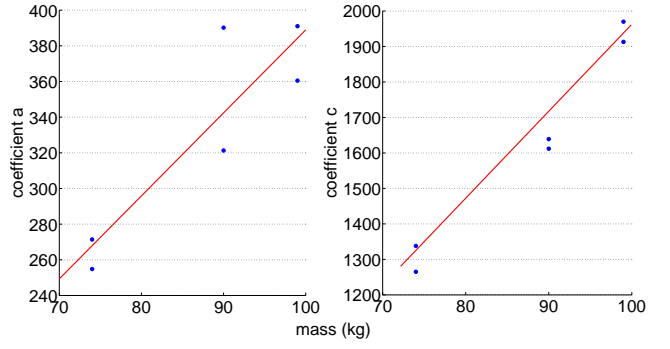


Fig. 8. Variation of the coefficient a and c as a function of the 3 subjects.

$$c(m) = 24.98 * m - 566.83 \quad (4)$$

where m is the body mass.

Using the above equations, we extend our initial logarithmic approximation function (Equation 2) with body mass compensation as follows:

$$GRF(m) = a(m) * \log_2(acc + b) + c(m) \quad (5)$$

where m is the body mass and b is set empirically to 1 and acc is maximum TAA peak at event **MP**.

B. Approximation Results

Bar graphs shown in Figure 9 shows the RMS errors across 3 different strides for each leg, when employing a linear

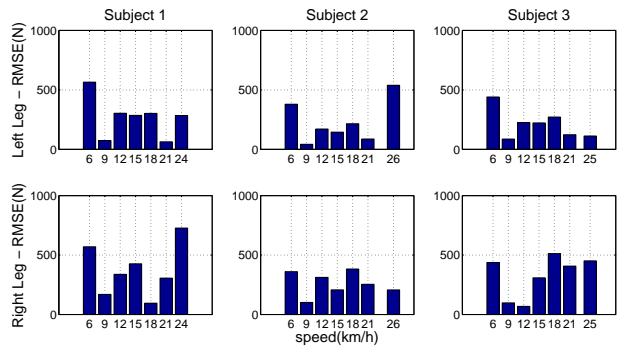


Fig. 9. Plot of RMSE errors using linear approximation for the 3 subjects.

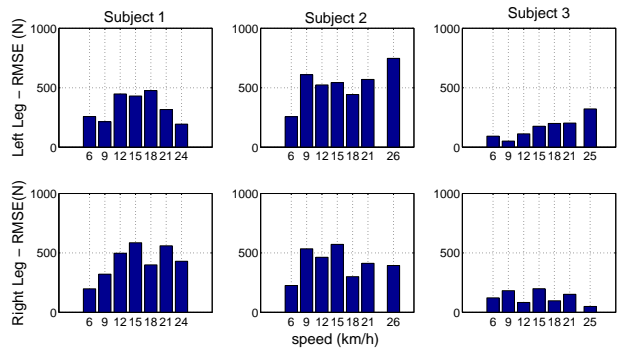


Fig. 10. Plot of RMSE errors of the 3 subjects using logarithmic approximation without body mass compensation.

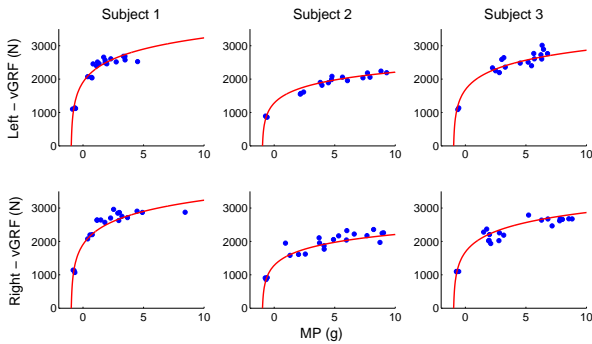


Fig. 11. Prediction of GRF using tibial accelerations for the 3 subjects.

approximation. It can be observed that at 6km/h , for the 3 subjects on both legs, the errors reached 500N on average. At the speeds $9, 12, 15, 18, 21\text{km/h}$, the errors increased from 50N reaching 480N in average at the fastest speed. Both legs showed similar results with a higher error in the walking speed (6km/h) and at sprinting.

Figure 10 shows the RMSE errors based on the logarithmic approach without body mass compensation. It can be observed that for Subject 3 (body mass 90kg), the errors were found at approximately 100N when walking (6km/h) and between 100 to 400N as the speed increased. Larger errors were found for subjects 1 and 2: approximately 250N when walking for both subjects on both legs, and reaching 750N on the left leg for Subject 2. For subject 1, the average approximation error was 400N , while errors tended to increase with faster speeds.

Finally, the estimation of peak vertical GRF for each subject using the logarithmic approach with body mass compensation (Equation 5) for left and right legs is shown in Figure 11. The plots for subject 2 show that for $9 - 21\text{km/h}$ speeds, the peak accelerations mostly ranged from -1 to 10g on both legs, while vertical GRF ranged from 1500 to 2000N . For subject 3, the accelerations and GRF were distributed from 2 to 7g and 1800N and 3000N . Finally, the bar graphs in Figure 12 show that the errors of the estimation for subjects 1, 2 and 3 on the left and right legs were on average 157N and 151N , 106N and 153N , and 130 and 162N respectively. Average error in sprinting for all three subjects was approximately 250N .

V. DISCUSSION

Logarithmic approximation of GRF with body mass compensation achieved an RMSE average of 151N , 106N and 130N respectively, for the three subjects. The respective normalized errors with respect to the peak GRF measured by the force plate were 6.1% , 5.9% and 5.4% on average across all speeds, as observed in Figure 12. Results showed that a logarithmic relationship between the peak vertical GRF and the TAA performed better than a linear approximation. This suggests a non-linear nature of the tibial accelerations amplitudes at different running velocities. The same result can be observed in Figure 9, where the errors at 6km/h for all subjects were considerably larger for all the speeds in the linear case compared to the logarithmic approach. In addition, when the subject ran at faster speeds, the linear approximation showed increasing errors ranging from 100N to 500N correlated with the speeds.

The Maximum Peak (MP) event presented the highest overall correlation values with vertical GRFs as shown in Table II. It is hypothesized that the MP occurs close to the mid-stance phase, where in running and walking speeds, it is characterized by the maximum loading of body on the foot. Hence, the maximum vertical GRF presents higher correlation to the push-off onset acceleration of the leg. However, future work using ViPerform sensors data synchronized with the force plate data will investigate this aspect further.

Results in Figure 12 showed a large improvement of up to 75% by using a linear relationship between the body mass and the slope (a), and the offset (c) of the logarithmic relationship. For all subjects, average error reduction was 30% . This result is in agreement with the earlier studies that reported the body mass as a relevant factor for measurements of GRFs [18].

In addition, the results showed in the scatter plots in Figure 7 revealed that the Subjects under similar running speeds showed different peak accelerations. Subject 2's accelerations had larger magnitudes although his GRF were lower when compared to Subject 1, where the accelerations were lower with higher GRF. This suggests that although the subjects showed different gait patterns and shoe types, the RMS errors in the logarithmic estimation were reduced for the 3 subjects reaching an average of 150N across all the running speeds.

No transformation to the global frame was performed [19] suggesting that the gravitational component added in the accelerometer frame had reduced effect on the amplitude of the 4 events described in Figure 3. It is hypothesized that it was added uniformly between Heel-Strike (HS) and Toe-Off (TO), reaching its maximum projection at the mid-stance phase, where the leg was geometrically perpendicular to the ground. However, further research must ensure this.

Finally, negligible differences in the errors of the left and right legs revealed that the base station placement on the left upper arm was suitable to ensure adequate wireless link quality. Further research will investigate the affect of sampling frequency on estimation accuracy and ascertain if the sampling frequency of 100Hz is sufficient to collect the acceleration signal signature at faster running speeds. Moreover, a larger trial will be undertaken in future, involving more subjects and running speed variations to further strengthen the estimation accuracy of our approach.

VI. CONCLUSION

In this paper, we discussed the use of a low-power wireless system (ViPerform) employing one 3D accelerometer as a valid tool for GRF measurement. We assessed the effectiveness of a non-linear approximation of GRFs using tibial axial acceleration at different speeds. Our results also showed that under kinetic and anatomical assumptions of the body, such as the mass, our system showed good agreement with GRF measured by a commercial force platform. This system can thus be used in analysis of running patterns outside laboratory settings and in runners' natural environment.

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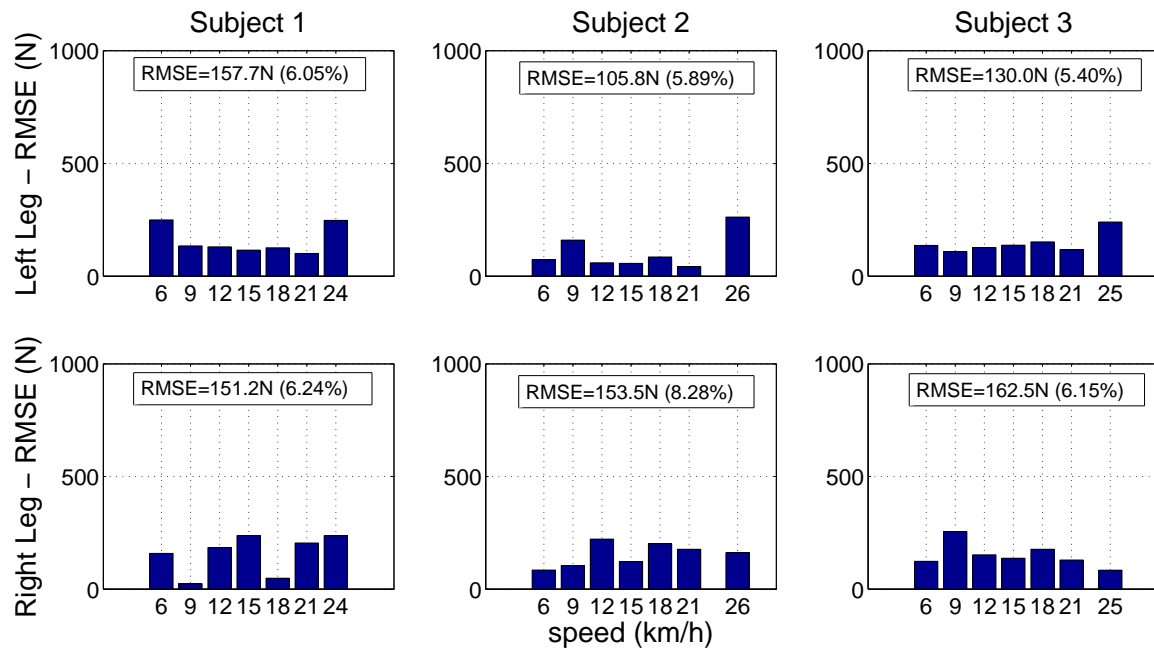


Fig. 12. Plot of RMSE errors of the 3 subjects using logarithmic approximation with body mass compensation.

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