Self Calibrating Wearable Active Running Asymmetry Measurement and Correction

M. Fiorentino, A. E. Uva, M. M. Foglia *

**Dip. di Ingegneria Meccanica e Gestionale, Politecnico di Bari, Bari, Italy*
(e-mail: m.fiorentino@ poliba.it , mm.foglia@ poliba.it, a.uva @ poliba.it)

Abstract: We present a novel self-calibrating wearable device to improve running training by active vibro feedback. The system consists of a pair of spandex shorts embedding: a processor unit, 2 three-axial accelerometers, 2 vibro motors, a SD card reader\writer module and a real time clock. Two supplementary wireless accelerometers are located on the shoes. We present an algorithm to compute gait asymmetry from the four sensors data. The main novelty is the auto calibration algorithm which uses principal component analysis on each sensor based on kinematic assumptions. The system provides two important advantages: data logging and real time active correction. The active correction is performed sending signals to the user in real time via vibrations cells (rumble feedback). The vibration signals are sent to the specific leg and its intensity is proportional to the entity of required correction. This training system for running can be very useful to athletes and to sport medicine in order to improve speed, posture, fatigue and reduce joints osteoarthrotic problems.

Keywords: running training, active posture correction, wearable wireless device, vibro feedback.

1. INTRODUCTION

Running athletes usually exercise strength, aerobic conditioning, cross-training, stretching, etc. Posture and coordination are usually over-look by athletes, professional trainers and even from sport medicine. However posture has been proven fundamental in sport efficiency and performance (see Williams et a. 1991). When the proper application of mechanical forces in human body is disrupted, inappropriate assistance (synergistic) and opposing (antagonistic) muscle contractions are required. When any of these conditions are not fully evaluated and corrected, abnormal patterns of biomechanical alignment are produced and impaired movement can occur. A long term lack of balance in posture and coordination can lead to early fatigue, cramps and physiological problems. Research on this topic is wide and interests specific sport activities domains and in particularly, running, (Radin et al. 1985). Running requires the coordination of the multitude of degrees of freedom of lower limbs. Experiments in this sport discipline demonstrated that rapidly applied loads on articulations (impulsive loads) cause osteoarthrotic problems. Repeated movement and impulsive loading have to be applied together to produce joint damage (Olney et al. 1991).

Most of the commercial training assistant devices focus on performances measurements (average speed, total time, heart rate and zone, etc.) or on goal motivation (GPS location, estimated time of arrival, speed). Posture and kinematic measurements are restricted to a very limited use in professional athletes and in rehabilitation. Nowadays sensors miniaturization, reliable wireless connections, and wearable computer processing can leverage the training support and let it being affordable to the occasional sportsman. Therefore our main interest is to develop a simple and easy to use device to support running coordination (0).

Fig. 1. The weareable running assistant with two rumble devices and four accellerometers mounted on spandex shorts and shoes.

We want to acquire in real time the motion of both legs in the 3D space, evaluate the asymmetries and provide the user with a helpful feedback. We acquire the inertial data from four accelerometers located on each leg (knee and foot position as depicted in 0), process the correction, and provide the user with rumble feedback to improve his/her sport performance. The design specifications of the system are: (i) low cost, (ii) wearable, (iii) easy to use. The main application is for professional running, but it could be easily applied to other similar sports like jogging and recreational training. In the next section we present how the lack of symmetry and coordination due to fatigue can cause performance loss and long term damage on articulation joints.

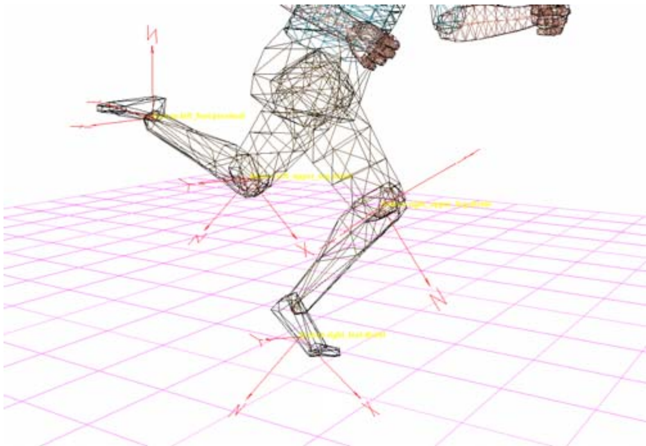

Fig. 2. 3D kinematic of running and the accellerometers positions.

2. RELATED WORK

Many works in literature study the changes of leg kinematics occurring in different stages of endurance activities, presumably as a result of fatigue (e.g. Hamill et al. (1992), Williams et al. (1991) and Siler et al. (1991)). Others have found differences in rear foot motion in over ground or treadmill running before and after exhaustion treatments.

Few relevant kinetic data are available in literature: e.g. ground reaction forces or resultant joint forces and moments. Brüggemann et al. (1994) reported a decreased impact during endurance running at constant speed sometimes accompanied by an increase in rearfoot motion during the early stance phase. Nicol et al. (1991) studied the influence of exhausting running on neuromuscular performance. They found significant differences in parameters indicating power in lower leg activities during short stretch-shortening cycles before, during and after a marathon. Results indicated the impact and peak active ground reaction forces during the stance phase of a sprint test were lower after running a marathon. This pattern was found in both vertical and horizontal components. The duration of the stance phase increased. From these results they concluded that fatigue resulted in a reduced ability to sustain stretch loads. However, as the maximal velocity of the sprint decreased from one sprint test to the next, care must be taken in the interpretation of these results. It can be speculated that muscular and neural fatigue lead to significant changes in muscle activation patterns, which in turn influence the stiffness and probably the elastic energy storage capacity of the muscle-tendon complex.

For a long time sagittal plane kinetic has been used to define normal or pathological gait, see Olney et al. (1991). However, the sagittal view provides only a portion of the movement in gait. Eng and Winter (1995) measured that 23% of power is spent in transversal plane motions. This proves that a large amount of energy is wasted in medial plane movements due to asymmetry. Kuo et al. (2010) examined the basic principles of dynamic walking, an approach that combines an inverted pendulum model of the stance leg with a pendulum model of the swing leg and its impact with the ground. The dynamic walking approach can predict the consequences of disruptions to normal biomechanics, and the associated observations can help to explain some aspects of impaired gait. Senden et al. (2011) demonstrated how acceleration-based gait analysis is sensitive for different walking conditions and is able to differentiate functional knee limitations.

The presented related works motivates the interest in this field and the development of the proposed training tool as described in the following section.

3. OVERVIEW OF THE SYSTEM

We measure the motion of each leg and foot using four 3D accelerometers located at knee level on tights and on top of the shoes (see 0 for nomenclature). We provide the user with a real time feedback of the asymmetry to correct the movements and improve sport performance. The system overview is depicted in the scheme in 0. The hardware is composed by the central unit (main unit), two wired (KL and KR) and two wireless (FL and FR) tri-axial accelerometers and two rumble feedback devices (Left-Right motor). We stitched the components on a pair of spandex shorts. Foot accelerometers are wireless connected.

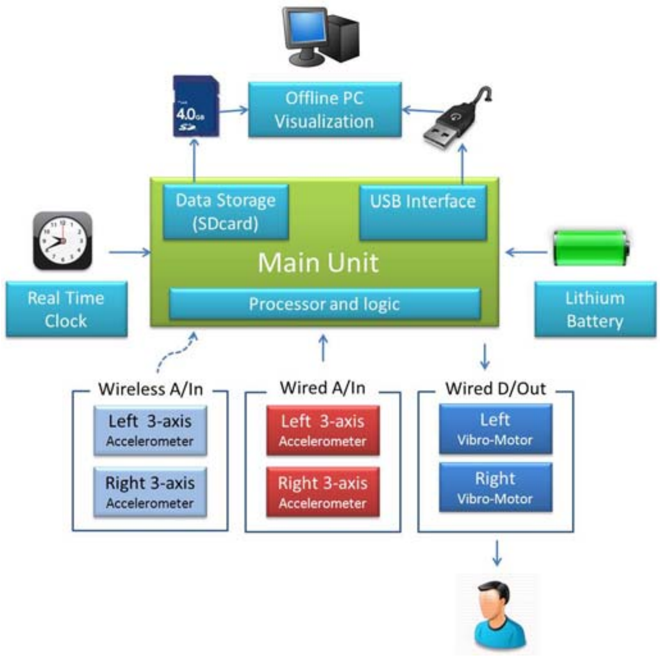

Fig. 3. Scheme of the our running assistant system.

The system provides the following two functions: (i) real time vibro-feedback for active correction and, (ii) data logging for offline post-analysis and training planning. Both scenarios may be useful to improve performance and motivation.

3.1 The central processing unit

The central unit has two main functions: data collection and processing. We use an open-source microcontroller to develop our mobile measurement system (*Arduino Mega 2560* board with an Atmel AVR processor, Arduino (2011). We equipped the main unit with a SD card reader\writer shield, Adafruit (2011), a digital clock module and a set of sensors. The input/output data flow is detailed in 0. We connect the wired accelerometers to the 12 analog inputs port, while we use 2 digital outputs to drive the micro motors. We use Xbee Shield to the wireless access to the shoe sensors. The device is powered by a 1200mAh capacity lithium ion battery for a theoretical functional time of 8.6 hours. Battery drain can vary according the amount of vibro signals sent to the user (which is basically related to his training level). We tested the system for more than 3.5 hours with a very low trained user, which proved the autonomy to be sufficient for most training sessions.

3.2 3D acellerometers

Four sensors are used to evaluate knees and feet acceleration along 3 axis. We use the ADXL330, a small, thin, low power, 3-axis accelerometer with signal conditioned voltage outputs. It measures acceleration with a range of ±3 g. It can measure the static acceleration of gravity in tilt-sensing applications, as well as dynamic acceleration resulting from motion, shock, or vibration. The user selects the bandwidth of the accelerometer using the CX, CY, and CZ capacitors at the XOUT, YOUT, and ZOUT pins. Bandwidths can be selected to suit the application, with a range of 0.5 Hz to 1600 Hz for X and Y axes, and a range of 0.5 Hz to 550 Hz for the Z axis. The ADXL330 is available in a small, low profile, 4 mm × 4 mm × 1.45 mm, plastic frame chip package.

0 depicts the location and the reference frame of the four accelerometers. Shoe mounted sensors have their own battery and an embedded Xbee transmitter module. Those sensors are firmly strapped to the runner shoes and do not denote any limitation to movements because of their light weight (less than 15g).

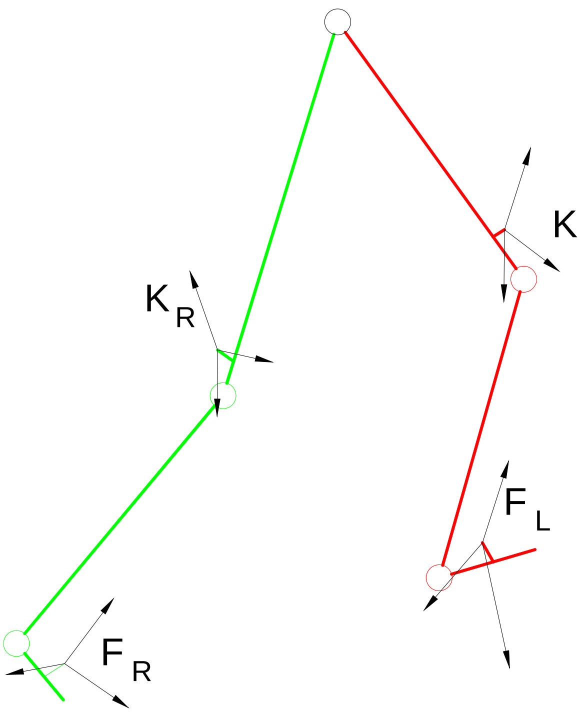

Fig. 4. Reference frames of the four accelerometers and their nomenclature: Knee Left (K_L), Knee Right (K_R), Foot Left (F_L), Foot Right (F_R).

3.3. Vibro feedback

Providing real time feedback to the runner is not a trivial task. Early tests demonstrated clearly how visual displays (coloured LED lights, wristwatch, LCD monitors, etc.) are not suitable in running. Two are the major drawbacks: they require cognitive effort therefore distract runner from street dangers; secondly any form of visual communication will take a latency between the signal and perception. Audio signals can be very fast in delivery, but they are not applicable because annoying and not feasible due to safety issues (e.g. road regulations forbid cyclists to use headsets in several countries). Our novel solution is to provide fast and intuitive signals via vibro feedback. Vibro feedback, or rumble feedback, is a tactile stimulus obtained by the vibration of fast rotating eccentric masses. This technology is common in mobile phones and it is available at low cost. Yang et al. (2002) demonstrated with user tests, how the tactile feedback can be used to direct the motion of the limbs, indicating also direction and intensity. The main advantage of vibrations compared to other stimuli for this application is the low latency.

We developed a rumble feedback output specifically suited to be effective in a running training. We stitched two vibro motors on the spandex fabric, near the accelerometers (see 0). We can command each motor independently for of feedback output with a wide spectrum of effects. We tested several functions of intensity values using a dedicated editor. Default values can be also changed by the user via software using the USB connection and a PC.

4. SELF CALIBRATING LOCAL REFERENCE PLANE

We propose a novel method to extract significant data from accelerometers without complex calibration phases. We first need to find the plane in which each accelerometer is moving. If we consider a typical reference system $\mathbf{S}_b$ for human body, we assume the vertical axis as $\mathbf{z}_b$ axis, the forward axis as $\mathbf{y}_b$, and the left-right horizontal axis as $\mathbf{x}_b$ by consequence. $\mathbf{z}_b$ and $\mathbf{y}_b$, are on the sagittal plane.

We make some assumptions, supported by literature:

a) Each accelerometer moves principally (within an error ε) in a plane, that we call local reference plane π_r;

b) Each accelerometer has a spin axis mainly normal to the local reference plane;

c) Each local reference plane π_r is almost parallel to the sagittal plane of the body.

Using these assumptions we developed the following calibration-less approach. For each tri-axial accelerometer {KL,NR,FL,FR} we collect N acceleration vectors in a period of time *Ts*. We set T_S equal of three gait cycles, by default.

Each accelerometer has its own reference system with $\mathbf{x}_a$, $\mathbf{y}_a$, $\mathbf{z}_a$ axes, that we call $\mathbf{S}_a$, generally not aligned and not left-right symmetrical. The components of the i^{th} acceleration vector $\mathbf{v}_a^i$ are evaluated (see 0) in the accelerometer reference system $\mathbf{S}_a$ as follows:

$$\mathbf{v}_a^i = \{ v_{ax}^i, v_{ay}^i, v_{az}^i \}$$

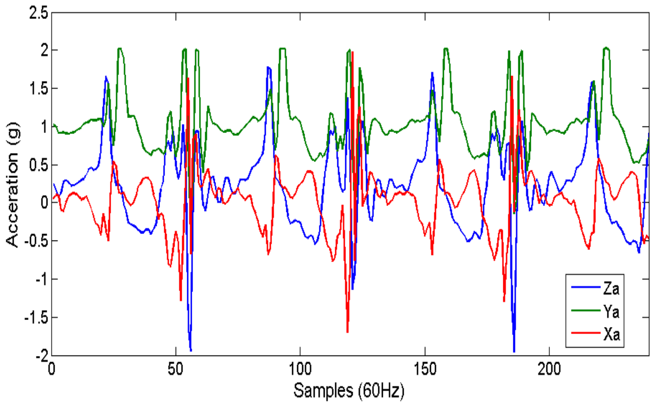

Fig. 5. Experimental components for the KR accelerometer in its reference frame.

Because of assumptions a) and b) all the vectors $\mathbf{v}_a^i$ mostly lie in a 2D plane (reference plane π_r) embedded in the 3D measurement space. In case of planar motion at constant speed, in every gait cycle, the mean of acceleration vectors should be the gravity acceleration **g** with no instant component out of the plane. In real gait, because of assumptions b) and c), the gravity vector lies on the reference plane π_r, therefore it does not affect the plane evaluation, while a component off the plane exists. This component is what makes the movement asymmetric.

We use PCA (Principal Component Analysis) to find π_r. Given a set of N vectors in R^3, the covariance matrix **C** of the set is:

$$C = \frac{1}{N-1}\left(D^T D\right) \tag{1}$$

where D is the matrix

$$D = \begin{pmatrix} v_{ax}^1 - v_{ax}^0 & v_{ay}^1 - v_{ay}^0 & v_{az}^1 - v_{az}^0 \\ \vdots & \vdots & \vdots \\ v_{ax}^N - v_{ax}^0 & v_{ay}^N - v_{ay}^0 & v_{az}^N - v_{az}^0 \end{pmatrix} \tag{2}$$

and

$$\bar{c} = \begin{pmatrix} v_{ax}^0 \\ v_{ay}^0 \\ v_{az}^0 \end{pmatrix} \tag{3}$$

is the geometrical mean, i.e. the centroid, of the N vectors. PCA is done by eigenvalue decomposition where the 3x3 matrix **C** can be factored as $\mathbf{U}^T\mathbf{L}\mathbf{U}$, where **L** is diagonal and **U** is an orthonormal matrix. PCA gives the orthogonal basis in which the covariance matrix **C** of our data is diagonal. **U** contains the eigenvectors of this basis point in successive orthogonal directions in which the data variance is maximum. **L** contains the eigenvalues which represents the variances.

In the case of data mainly residing on a 2D plane, the variance is much greater along the two first eigenvectors, which define our plane of interest in the accelerometer reference system,. The first eigenvector corresponding to the largest eigenvalue (greatest variance) of the covariance matrix is $\mathbf{e}_{max}$, the second is $\mathbf{e}_{mid}$ and the last, corresponding to the lowest variance is $\mathbf{e}_{min}$. These eigenvectors are called the principal components (see 0). We can evaluate the local reference plane π_r estimating its normal **n** using $\mathbf{e}_{min}$. This is true also for large movement because of assumption b). We confirmed, in fact, by experimental data that the accelerometer is moving mainly in the reference plane (assumption a).

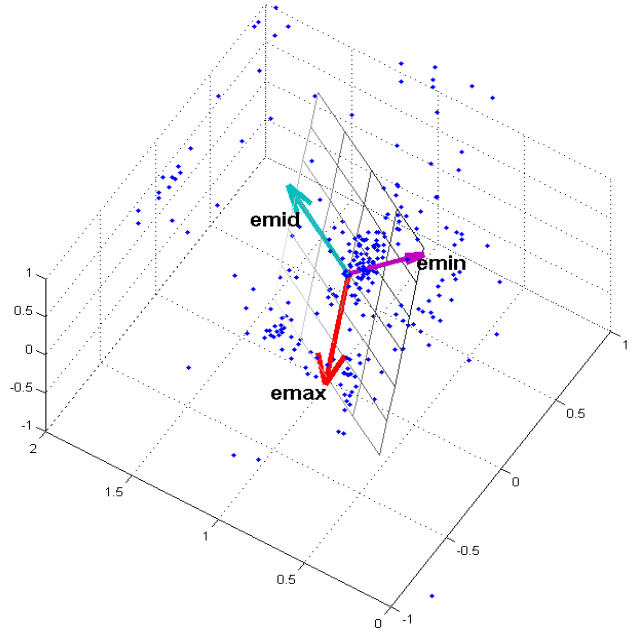

Fig. 6. Eigenvectors from PCA and reference plane for experimental data in 0.

To compute correctly the PCA, we need to consider all the acceleration data in one gait cycle. To define the period we consider the gap between two maxima in acceleration values of F_L and F_R corresponding to the ground contact. We can easily evaluate the footstep cycle (using the data timestamp) for each leg with a high level of confidence. The PCA calculation is linear in N, number of vectors. The essential cost of the operation is the calculation of the covariance matrix **C**. The calculation of the eigenvalues and eigenvectors is a fixed-cost operation, as it is performed for a 3x3 matrix. Therefore we can evaluate the reference plane in each footstep cycle without the need of calibration.

It is noticeable that we can easily map each accelerometer reference system $\mathbf{S}_a$ to the human body reference system $\mathbf{S}_b$ using more sophisticated (and more expensive) sensors to retrieve also gyro data, i.e. rotation of accelerometers. To calibrate this mapping we could use the three components of gravitational acceleration in $\mathbf{S}_a$ that must be mapped to $\mathbf{g}_b=\{0,0,-g\}^T$ in $\mathbf{S}_b$. Relative rotation matrices defining the orientation of $\mathbf{S}_a$ measured in $\mathbf{S}_b$ can be dynamically constructed using gyro data. But for the goal of our system, this complexity is not necessary.

Using only accelerations, without any calibration phase, we can define only the axis $\mathbf{x}_b$ which is sufficient to compare the movements of the four accelerometers and provide the user with feedback as described in the next section.

5. ERROR COMPUTATION AND USER FEEDBACK

During training, coordination errors due to fatigue or posture will lead to the generation of parasites accelerations which drift form the ideal kinematics of running. We use the four reference planes evaluated at each footstep cycle for the four sensors (see 0). We consider ε, the deviation of each accelerometer from the reference plane, as a measure of the symmetry of the movement in each of the four locations.

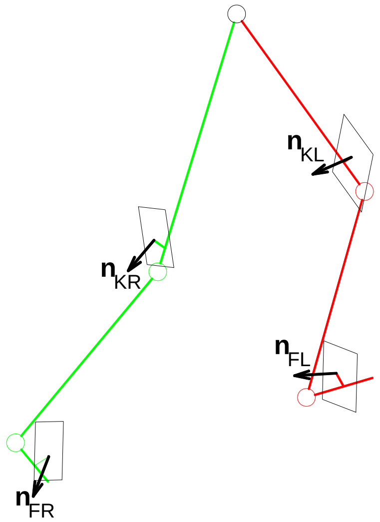

Fig. 7. The normal vector and motion plane.

We measure ε as the minimum eigenvalue for each accelerometer. Since knee joints relates foot with tights, the reference frame FL is linked to KL and FR is linked to KR. We compute also the PCA on the relative acceleration obtained by difference from foot and knee accelerations. The planar error on this set is a good guess of the planarity of the leg movement. We evaluate, for each step cycle, the following asymmetry indicators:

Table 1. Assymmetry indicators

	Left	**Right**
Knee	ε_{KL}	ε_{KR}
Foot	ε_{FL}	ε_{FR}
Δ(Foot-Knee)	ε_{FKL}	ε_{FKR}

Others irregularity indicators are the time-shift values between the maxima in acceleration for left and right knees (T_K) and feet (T_F). We developed an empirical asymmetry function considering all the aforementioned contributions:

$$F_{asym} = f(\varepsilon_{KL}, \varepsilon_{KR}, \varepsilon_{FL}, \varepsilon_{FR}, \varepsilon_{FKL}, \varepsilon_{FKR}, T_K, T_F)$$

If the asymmetry value is over a training threshold, the leg with disproportionate value will be advised by a vibrofeedback proportional to the error (frequency of activation of the vibro-motor).

Connecting the main unit to a PC, the user (coach or sport medicine doctor) can set the gain and the threshold value for the vibro activation. Data can be transferred in real-time to a connected PC for indoor (i.e. treadmill) complex analysis or saved to a SD card for post process and training.

6. IMPLEMENTATION

We developed the code in the Arduino microcontroller using the provided language and libraries. We wrote each single logical operation in a function. The system flowchart is summarized in the following pseudo-code:

```
WHILE <acceleration≥running_threshold>
    FOR EACH <N footstep cycles >
        FOR EACH <sensor>
            DO read data from 3Dsensors;
            DO store acceleration data in cache;
        DO detected footstep period;
        DO compute time-shift;
        DO compute PCA for data in N cycles;
        DO evaluate reference plane and ε;
        IF <asymmetry function≥training_threshold>;
            DO activate vibro feedback;
            DO store asymmetry data;
```

The N number of cycles can be set by the user according to his\her preference (3 is the default value). A higher number of cycles reduces the sensibility and it is suggested to novice runners, while professionals will prefer a sharp correction. While running, the first operation after the acquisition of sensor data in the first footstep cycle is to detect the four periods and their relative time-shifts. Then we can evaluate PCA to extract reference plane and ε for each sensor. Subsequently we evaluate the F_{asym} for each cycle. If the asymmetry value exceeds a threshold value, a vibro-feedback signal is triggered using a lookup table. We implemented a step signal proportional to the asymmetry value, but other feedback function can be edited.

7. CONCLUSION AND FUTURE WORK

We presented a novel wearable device to improve running training by active vibro feedback. The system consists of a pair of spandex shorts embedding: a processor unit, 2 three-axial accelerometers, 2 vibro motors, a SD card reader\writer module and a real time clock. Two supplementary wireless accelerometers are located on the shoes. We present an algorithm to compute gait asymmetry from the four sensors data. The main novelty is the auto calibration algorithm which uses principal component analysis on each sensor based on kinematic assumptions. The system provides two important advantages: data logging and real time active correction. The device stores all the training data in a SD card. Those log files can be post processed using a dedicated application on a PC. This information can be very useful for training planning and analysis. The active correction is performed in real time by signal sent to the user via rumble feedback. The vibration signal is sent to the specific leg and its intensity is proportional to the entity of asymmetry. This approach to running training is very useful to athletes and to amateur runners in order to improve speed, posture, fatigue problems and to reduce joints osteoarthrotic problems. In both cases our system can increase the muscular energy transfer by reducing ineffective movements and most important creating the correct muscular memory. Beyond this, the proposed system can be also useful in other medical appliances such as rehabilitation.

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