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Asymmetry measurement for vibroactive correction in lower limbs mobility

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Abstract. We developed a wearable device to improve lower limb sport training. The presented system consists of a pair of spandex shorts which embed a processor unit, 2 accelerometers and 2 vibro motors. The accelerometers are located in proximity to the knees and measure tri-axial accelerations. We present a novel method to compute and correct asymmetry of lower limbs during training. The user performs an initial calibration phase which sets the accelerometer reference frames. While running, the system continuously refines the calibration using principal component analysis to take in account occasional shorts assessments. The system activates a corrective vibro feedback on the specific leg according to an asymmetry metric based on: (i) foot ground impact, (ii) phase error between legs, (iii) transversal knee movements. User tests demonstrated that the device is ergonomic to wear, easy to use and the corrective vibro feedback is appreciated during the training.

Keywords: wearable training device, vibro feedback, asymmetry correction, sport training, lower limb rehabilitation

1. Introduction

To excel in sports, athletes commonly focus on strength training, aerobic conditioning, cross-training, stretching, etc. Posture and body kinematics are usually over-looked by athletes, trainers and even by sport medicine doctors. However the effects of posture in performance and fitness are critical. 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 movements can occur. The research activity on this topic is wide and focuses on specific sport activities domains as walking, running and pedaling. In particular, lower limbs articulation requires a high level of coordination because of the multitude of degrees of freedom. Simplified kinematic models presume the articulations as spherical or cylindrical hinges, as the in Jack human simulation system (Figure 1), developed at the Center for Human Modeling and Simulation at the University of Pennsylvania [1].

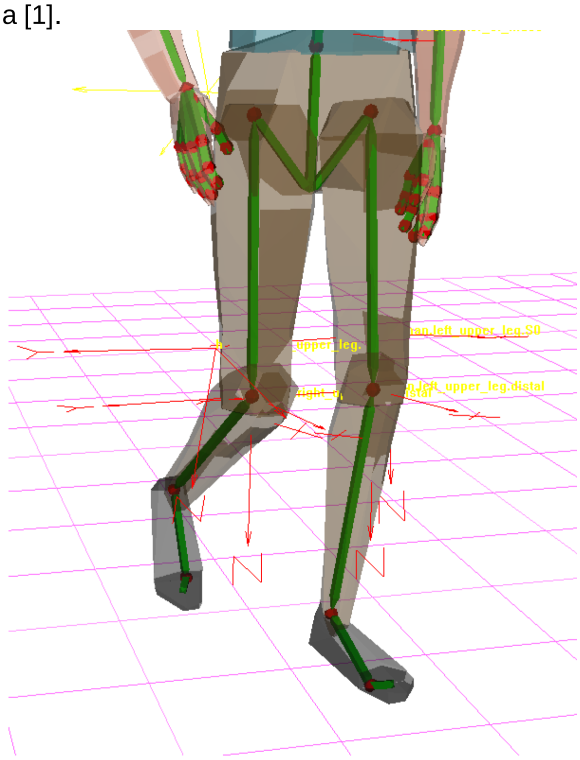

Fig. 1. Lower limb 3D kinematics scheme

Several wearable running devices with real time user motion sensors are commercially available at low cost for sport training. The need to keep a correct movement is also justified by the fact that impulsive loads on articulations during jogging and running have been demonstrated to cause osteoarthrotic problems [2][3]. Nike offers a simple shoe sensor to realize a simple, low cost and user friendly system with an IPhone to measure pace and to estimate speed using the step length [4]. Suunto proposes the Foot Pod Mini which wireless connects to Suunto training watches to estimate pace, distance and speed [5]. More precise motion capture systems are used in movie industry, in kinematic researches, in rehabilitation, and in occupational health and safety. Commercial proprioceptive inertial accelerometers, often coupled with gyros, for motion capture, e.g. Xsens [6], offers high performances (±50m/s^2, resolution 0.02m/s^2) at the expense of a difficult setup and of high computational cost. Optical sensors, e.g. Vicom [7], provide precise three dimensional position and orientation but require a complex camera setup and a fixed workspace. Ferrari et al. presented an experimental comparison of these techniques for gait analysis [8]. Percoco presents a specially designed low cost photogrammetric 3D scanners for the human body [9]. Although the precision obtained is more than sufficient for kinematic analysis, the system requires a calibrated and fixed setup.

Fig. 2. The weareable rumble device mounted on spandex shorts

The main requirements for a daily useable training device are the following: (i) effective user feedback, (ii) low cost, (iii) wearability, (iv) autonomy. In Fiorentino et al. [10] we presented a simple and easy to use wearable device to support lower limb coordination mounted on spandex running shorts (Figure 2). Our idea is to retrieve inertial data from two accelerometers located on each legs, processing a user correction function and providing the user with feedback to improve performances. One of the most important issues in the development of this device is the referencing of the accelerometers which can easily shift with user movements. Another important issue is related to the definition of a metric to measure the user asymmetry while running. This paper presents a novel mathematical definition of the asymmetry based on: (i) foot ground impact, (ii) phase error between legs, (iii) transversal knee movements. This metric is used as input for the activation of the corrective vibro feedback on the specific leg of the user.

2. Related work

Previous works have focused on the alteration of legs and feet kinematic due to fatigue [11-14]. Brüggemann et al. [15] reported a decreased impact during endurance running at constant speed. Nicol et al. [16] studied the influence of exhausting running on neuromuscular performance. Their results show that muscular and neural fatigue lead to significant changes in muscle activation patterns, which influence the stiffness and the elastic energy storage capacity of the muscle-tendon complex. The study of the kinematic on the sagittal plane has been used to define normal or pathological gait [17]. Eng and Winter [3] measured that 23% of the total walking power is spent in transversal plane motions. This proves that a large amount of energy is wasted in medial plane movements due to asymmetry. Haddad et al. [18] assessed gait symmetry using spatial, temporal, and higher-order symmetry measures. They pointed out that velocity measurements were more effective than angular ones to detect asymmetries. Kuo et al. [19] examined the basic principles of dynamic walking with an approach that combines an inverted pendulum model of the stance leg with a pendulum model of the swing leg. The dynamic walking approach can predict the consequences of disruptions in the correct biomechanics, while the associated observations can explain some aspects of the impaired gait.

3. Overview of the system

The hardware is composed by the main unit, two sensors and two rumble feedback devices. We fitted these components on a pair of spandex shorts (Figure 2), hiding the wiring in the seams. The scheme of the hardware setup is detailed in Figure 3.

3.1. Main Unit

In our prototype we use the *Arduino Uno* board with an Atmel AVR processor [20] coupled to a SD card reader\writer shield [21]. The device is powered by a 600mAh capacity lithium-ion battery for a average working time of 5 hours.

3.2. 3D accelerometers

The system uses two sensors to evaluate knees acceleration along 3 axis. We selected the ADXL330 sensor by Analog Devices, a small, thin, low power, 3-axis accelerometer with signal conditioned voltage outputs [22]. It measures accelerations within a range of ±3 g in a small, low profile, 4×4×1.45mm, plastic package.

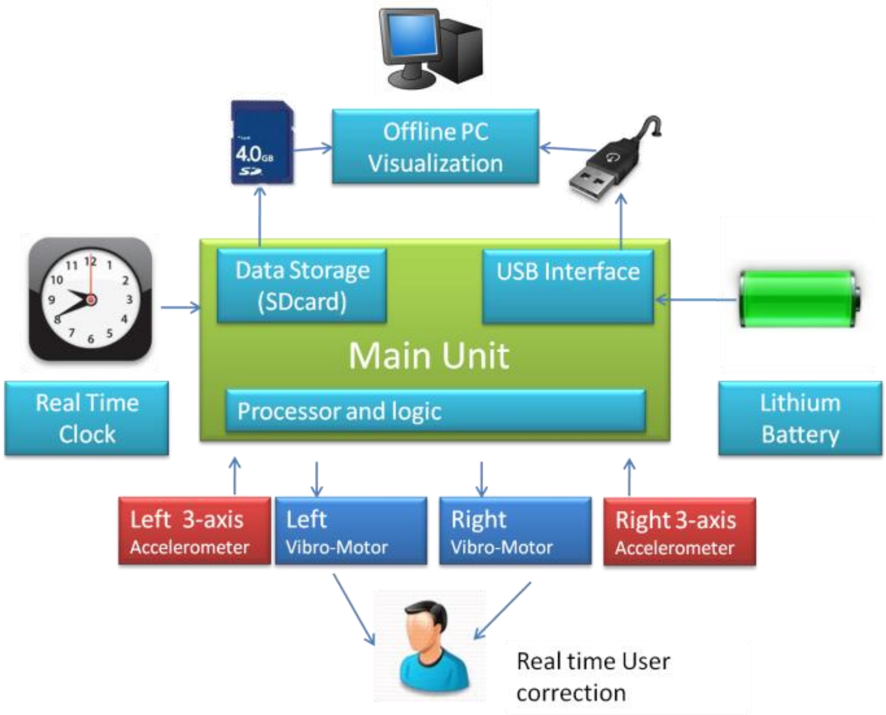

Fig. 3. Scheme of the hardware setup

3.3. Vibro feedback

A visual display may be not very convenient to convey information on the motion to the athlete because it requires cognitive effort and it may be distracting. On the other hand, audio signals will be annoying and non-feasible because of road safety laws. Our idea is to provide a fast and intuitive signal via vibro feedback which is a tactile stimulus obtained by the rotation of small eccentric masses. These off the shelf devices are common and cheap because of their massive usage in mobile phones. Yang et al. [23] demonstrated, with user tests, how the tactile feedback can be used to direct the motion of the limbs, indicating also directions 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 real training. We stitched two vibro motors on the spandex fabric, near the accelerometers (Figure 4) and in contact with the user skin. The vibrofeedback stimulus to the user is proportional to the evaluated user asymmetry.

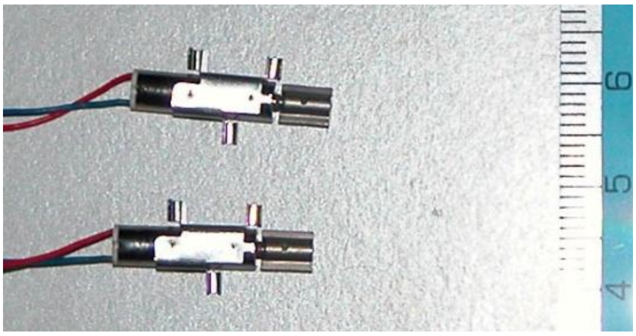

Fig. 4. Vibro feedback motors, with a centimeter scale

To estimate this asymmetry we need to know the position and the orientation of the two accelerometers respect to the user body [24][25]. We developed a specific hybrid calibration procedure to reference the accelerometers as described in the following section.

4. Calibration

A first "factory" calibration must be performed after the accelerometers are installed and wired on the shorts. This setup realigns the sensors built-in error equalizing the two accelerometers to improve the precision [26]. After this one time setup, there are two calibrations which must be executed during usage: (i) a "static" user calibration and (ii) an "on-line" automatic calibration refinement.

4.1. Static User Calibration

The user, after wearing the spandex shorts, must activate the "static calibration" procedure. This calibration is required because the 3D accelerometers position can vary according to how the user wears the shorts. In this phase we want to roughly estimate the orientation of the accelerometers in the body reference system.

Considering a reference system $\mathbf{S}_b$ for the human body (Figure 5), we assume the vertical axis as $\mathbf{z}_b$ axis, the forward axis as $\mathbf{y}_b$, and the left-right horizontal (transverse) axis as $\mathbf{x}_b$ by consequence. The axes $\mathbf{z}_b$ and $\mathbf{y}_b$, define the sagittal plane. Each accelerometer has its own reference system $\mathbf{S}_a$ with (appointed hereinafter with subscripts *L* and *R* for left and right leg) with $\mathbf{x}_a$, $\mathbf{y}_a$, $\mathbf{z}_a$ axes. Generally, $\mathbf{S}_{aL}$ and $\mathbf{S}_{aR}$ are not parallel to the body reference system. The following calibration procedure produces a different parameter set for each accelerometer. Because the algorithm is the same for both knees, we will describe the calibration math without involving the subscripts *L* and *R*. The user must keep, as still as possible, two postures (step 1 and 2 as depicted in Figure 6) for about one second in order to complete the static calibration.

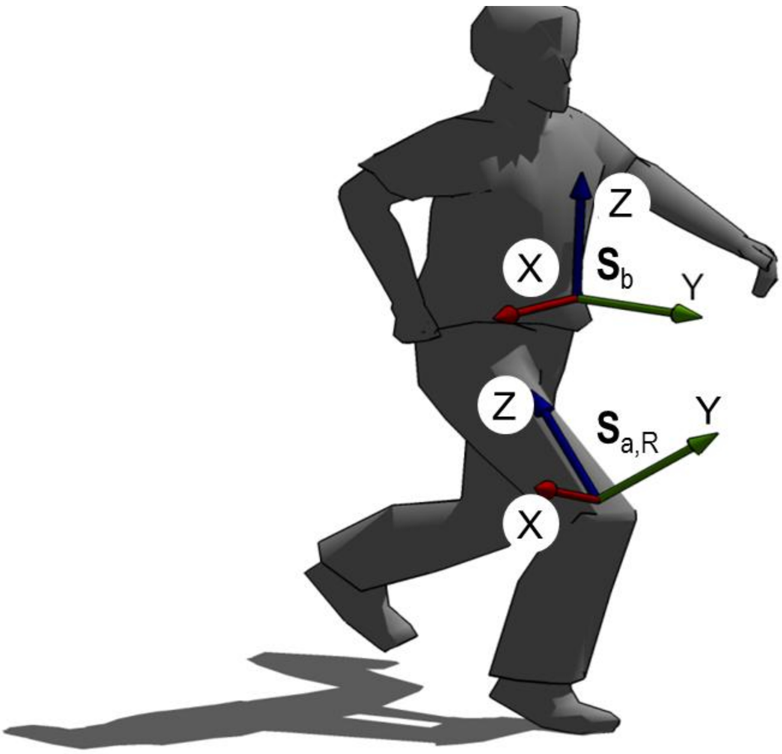

Fig. 5. User body (S_b) and acellerometers (S_a) reference frames

In step 1 (subscript 1 refers to calibration step 1), the accelerometer measures the three components of gravitational acceleration **g** vector in reference $\mathbf{S}_a$, therefore we have the vector:

$$\mathbf{v}_{a1} = \{v_{ax1}, v_{ay1}, v_{ay1}\}$$

whose module is about **g**. We can express the gravity acceleration in the body reference $\mathbf{S}_b$ as:

$$\mathbf{g}_b = \{0, 0, -g\}$$

approximating the $\mathbf{z}_b$ axis with the ground vertical axis. We define an unknown rotation matrix $\left(\mathbf{R}_a^b\right)_1$ that represents the orientation of $\mathbf{S}_a$ measured in $\mathbf{S}_b$. Therefore:

$$\mathbf{g}_b = \left(\mathbf{R}_a^b\right)_1 \mathbf{v}_{a1} \quad (1)$$

In step 2, we assume that $\mathbf{S}_a$ rotates “only” around the axis $\mathbf{x}_b$ and that the rotation angle is nearly 90°. In this new position, the gravity acceleration is expressed by:

$$\mathbf{v}_{a2} = \{ v_{ax2}, v_{ay2}, v_{az2} \}$$

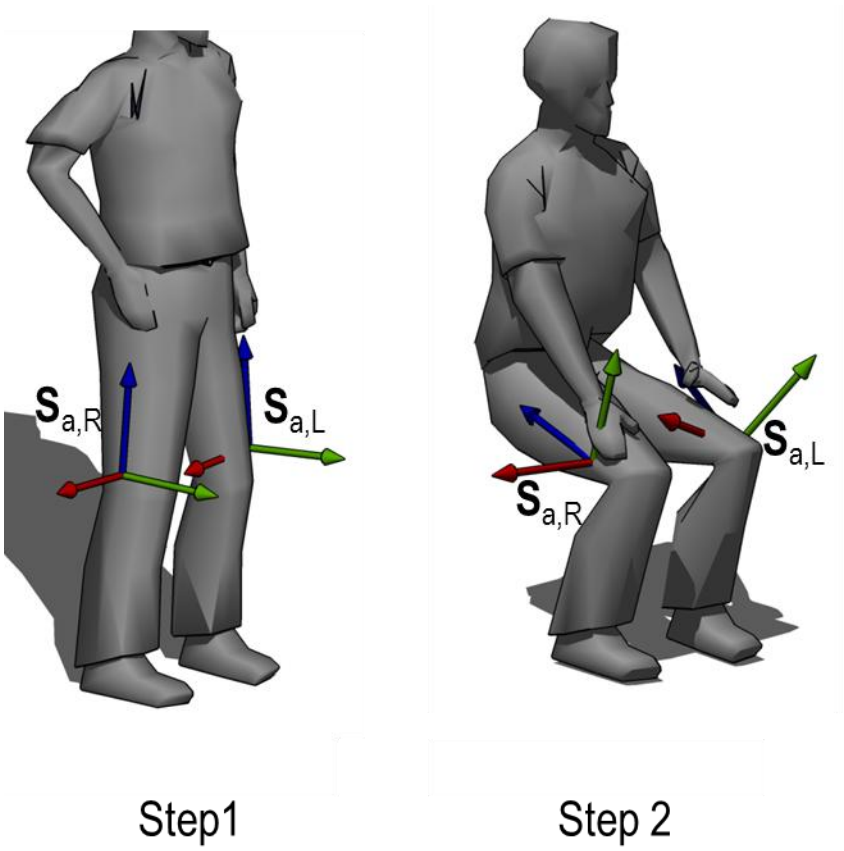

Fig. 6. User postures and orientations of the references systems during the static calibration

As in step 1, we can use a new unknown rotation matrix $\left(\mathbf{R}_a^b\right)_2$ that represents the new orientation of $\mathbf{S}_a$ in step 2 expressed in $\mathbf{S}_b$. As in eq (1) we can write:

$$\mathbf{g}_b = \left(\mathbf{R}_a^b\right)_2 \mathbf{v}_{a2} \quad (2)$$

If we assume that $\mathbf{S}_a$ rotates by 90° from step 1 to step 2, we can consider the relationship between $\left(\mathbf{R}_a^b\right)_2$ and $\left(\mathbf{R}_a^b\right)_1$ such as:

$$\left(\mathbf{R}_a^b\right)_2 = \mathbf{R}_x \left(\mathbf{R}_a^b\right)_1 \tag{3}$$

where $\mathbf{R}_x$ is a 90° rotation matrix around the $\mathbf{x}_b$ axis. In equation (3), the components of $\left(\mathbf{R}_a^b\right)_1$ and $\left(\mathbf{R}_a^b\right)_2$ are still unknown. Also using equation (1) and (2), we cannot uniquely determinate the real orientation of the accelerometer references systems. We still need a third independent posture to find the unknowns. But we can easily reference the body sagittal plane in the accelerometer reference systems evaluating the normal unit vector $\mathbf{i}_b$ of $\mathbf{x}_b$ in $\mathbf{S}_a$ by the following equation (4):

$$\mathbf{i}_b = \frac{\mathbf{v}_{a1} \wedge \mathbf{v}_{a2}}{\left|\mathbf{v}_{a1} \wedge \mathbf{v}_{a2}\right|} \tag{4}$$

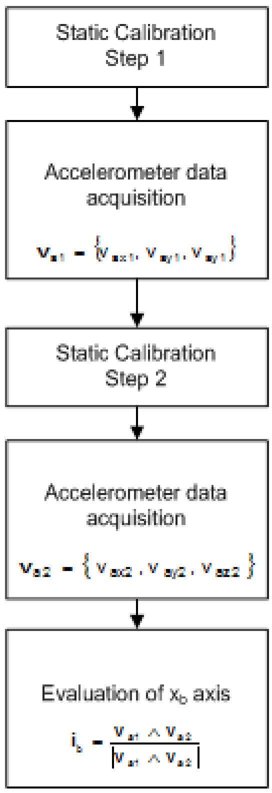

Fig. 7. Flowchart for static calibration

In a correct motion, the variation of $\mathbf{S}_a$ in $\mathbf{S}_b$ can be approximated by a simple rotation around $\mathbf{i}_b$. Figure 7 depicts the flowchart of the proposed static calibration procedure.

While running, we continuously measure $\mathbf{v}_a$ for each leg and evaluate $\mathbf{a}_n$, the acceleration normal component to the sagittal plane, and $\mathbf{a}_s$, the component on the plane:

$$\begin{aligned} \mathbf{a}_n &= (\mathbf{v}_a \cdot \mathbf{i}_b)\mathbf{i}_b \\ \mathbf{a}_s &= \mathbf{v}_a - \mathbf{a}_n \end{aligned} \tag{5}$$

where $\mathbf{i}_b$ is defined in (4). In the next section we describe the on-line refinement approach to trim the $\mathbf{i}_b$ direction continuously while running.

4.2. On line calibration refinement

We observed that, during running movements, the spandex shorts can slightly move over the thighs and thus void the precision of the "static" initial calibration. The main idea is based on the fact that we can statistically evaluate, for each gait cycle, the plane in which each accelerometer is moving. We can then compare this plane with the starting reference frame and update it accordingly.

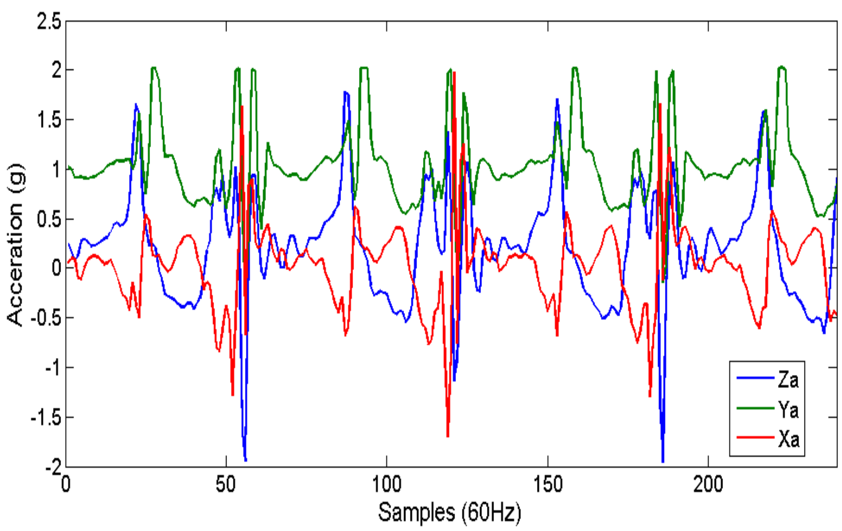

Fig. 8. Experimental components of accelerometer vector $\mathbf{v}_a$

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 (as supposed in the previous section);
c) Each (Left/Right) local reference plane π_r is almost parallel to the sagittal plane of the body.

We developed the following approach. For each tri-axial accelerometer, we collect N acceleration vectors in a period of time 3xT_{gait}. To define the gait period T_{gait}, we consider the gap between peaks of acceleration $\boldsymbol{a}_s$ due to the ground impact.

The i^{th} sampled acceleration vector $\mathbf{v}^i_a$ is measured (Figure 8) in the accelerometer reference system $\mathbf{S}_a$ as:

$$\mathbf{v}^i{}_a = \left\{ \mathbf{v}^i{}_{ax}, \mathbf{v}^i{}_{ay}, \mathbf{v}^i{}_{az} \right\}$$

Because of assumptions a) and b) all the vectors $\mathbf{v}^i_a$ mostly lie in a 2D plane (reference plane π_r) embedded in the 3D measurement space. We use PCA to find π_r. Given a set of N vectors in R^3, the covariance matrix $\mathbf{C}$ of the set is:

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

where D is the matrix:

$$D = \begin{pmatrix} \mathbf{v}^1_{ax} - \mathbf{v}^0_{ax} & \mathbf{v}^1_{ay} - \mathbf{v}^0_{ay} & \mathbf{v}^1_{ay} - \mathbf{v}^0_{ay} \\ \vdots & \vdots & \vdots \\ \mathbf{v}^N_{ax} - \mathbf{v}^0_{ax} & \mathbf{v}^N_{ay} - \mathbf{v}^0_{ay} & \mathbf{v}^N_{az} - \mathbf{v}^0_{az} \end{pmatrix}$$

The vector:

$$\bar{c} = \begin{pmatrix} \mathbf{v}^0_{ax} \\ \mathbf{v}^0_{ay} \\ \mathbf{v}^0_{ay} \end{pmatrix}$$

is the geometrical mean, i.e. the centroid, of the N vectors. PCA is done by eigenvalue decomposition where the 3x3 matrix $\mathbf{C}$ can be factored as $\mathbf{U}^T\mathbf{L}\mathbf{U}$, where $\mathbf{L}$ is diagonal and $\mathbf{U}$ is an orthonormal matrix. PCA gives the orthogonal basis in which the covariance matrix $\mathbf{C}$ of our data is diagonal. $\mathbf{U}$ contains the eigenvectors of this basis point in successive orthogonal directions in which the data variance is maximum. $\mathbf{L}$ contains the eigenvalues which represent 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 (Figure 9). We can evaluate the local reference plane π_r estimating its normal $\mathbf{n}$ using $\mathbf{e}_{min}$.

In the case of motion at constant speed, in every gait cycle, the mean of the acceleration vectors is the gravity acceleration **g**. This mean vector lies on the reference plane π_r and consequently it does not affect the evaluation of $\mathbf{e}_{min}$. To compute correctly the PCA, we need to consider all the acceleration data in one gait cycle. 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. Each T_{gait} we update the $\mathbf{x}_b$, using the $\mathbf{e}_{min}$ calculated by PCA in the last 3 cycles. We check the angle between the new $\mathbf{x}_b$ and the original $\mathbf{x}_b$ obtained by static calibration. If larger than 15° we warn the user for a new static calibration.

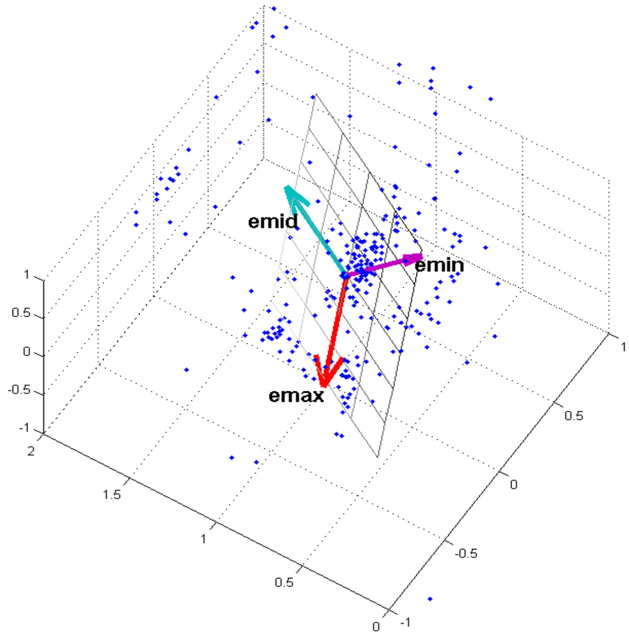

Fig. 9. PCA eigenvectors and reference plane for the experimental data in Figure 8.

5. Asymmetry evaluation

During training, coordination errors due to fatigue or posture will lead to parasites accelerations which drift form the ideal kinematics of running. We define a metric to evaluate the kinematic asymmetry considering the following hypothesis:

a) The shorts must be correctly worn (left\right symmetry and no warping along the legs).
b) Factory and static calibration must be correctly completed.

c) The path is straight (the curvature radius is high enough to ignore the centripetal acceleration)

Due to the time shift between the movement of the left and right leg, the asymmetry must be evaluated for the data buffered in the last gait cycle. The acceleration signals acquired by sensors in an ideal gait should be periodic and equal for left and right leg and shifted by half a gait period (i.e. $T_{gait}/2$). In real gait we propose to evaluate an asymmetric behavior using three parameters: feet impact error, phase error and transverse error.

The *feet impact error* (ε_p**)** measures the difference between left\right accelerations peaks of $\mathbf{a}_s$ caused by foot ground impact. This indicator is computed using the maximum values of the acceleration component in sagittal plane $\mathbf{a}_s$, as follows:

$$\varepsilon_p = \frac{\mathrm{Max}(a_{s,L}) - \mathrm{Max}(a_{s,R})}{\mathrm{Max}(a_{s,L}) + \mathrm{Max}(a_{s,R})} \in [-1, 1] \quad (\mathbf{6})$$

where $a_{s,L}$ and $a_{s,R}$ are the modules of the vectors $\mathbf{a}_s$ for left and right leg evaluated by eq. 5. Values of ε_p near zero indicate a symmetric feet pressure. A negative value of ε_p means that the left foot hits the ground with lower pressure compared to the right one.

The *phase error* (ε_φ) measures the time shift between the left and the right leg. This indicator is computed using the left foot phase $\varphi_{L/R}$ of $\mathbf{a}_s$ peak, measured in the right foot cycle, as follows:

$$\varepsilon_\varphi = \frac{\varphi_{L/R} - 90°}{90°} \in [-1, 1] \quad (\mathbf{7})$$

where values near zero are indicator of a symmetric cadence. A negative value of ε_φ means that the left foot hits the ground earlier than expected.

The *transverse errors* ε_t measure, for each leg, the deviation from the sagittal plane (i.e. along the transverse axis). This indicator is computed, for each accelerometer, as the minimum eigenvalue evaluated for the on-line calibration normalized with maximum eigenvalue to consider the different gait speeds, as follows:

$$\varepsilon_{t,L} = \frac{\lambda_{min,L}}{\lambda_{max,L}};\ \varepsilon_{t,R} = \frac{\lambda_{min,R}}{\lambda_{max,R}} \in [0, 1] \quad (\mathbf{8})$$

where λ_{min} and λ_{max} are the eigenvalues associated to the $\mathbf{e}_{min}$ and $\mathbf{e}_{max}$ eigenvectors, for left and right knees. The higher is the error value, the higher is the off plane displacement for each knee.

6. User feedback

We use an online vibro feedback to notify the user in real time about her\his asymmetric behavior.

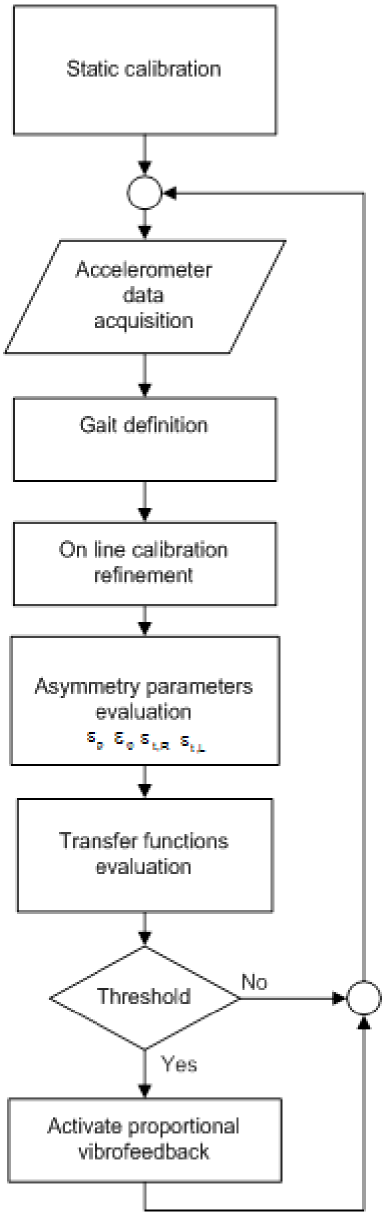

Fig. 10. Flowchart of the online algorithm

We developed a transfer function *F* which converts the error parameters into a leading signal for the vibro motors. The *transverse error* ε_t is already separated for left and right leg. We need to identify the side that needs correction accordingly to the first two parameters (ε_p, ε_φ) which are differential. We use the following filters:

$$\begin{aligned} E_{p,L} &= \frac{-\varepsilon}{2}\left(1-\mathrm{sgn}\left(\varepsilon_{p,L}\right)\right) & \qquad E_{\varphi,L} &= \frac{-\varepsilon}{2}\left(1-\mathrm{sgn}\left(\varepsilon_{\varphi,L}\right)\right) \\ E_{p,R} &= \frac{\varepsilon}{2}\left(1+\mathrm{sgn}\left(\varepsilon_{p,R}\right)\right) & \qquad E_{\varphi,R} &= \frac{\varepsilon}{2}\left(1+\mathrm{sgn}\left(\varepsilon_{\varphi,R}\right)\right) \end{aligned} \tag{9}$$

where all the errors E range in [0, 1]

We define a transfer function F for each leg using a linear combination of the errors as follows:

$$\begin{aligned} F_L &= K_p\, E_{p,L} + K_\varphi\, E_{\varphi,L} + K_t\, \varepsilon_{t,L} \\ F_R &= K_p\, E_{p,R} + K_\varphi\, E_{\varphi,R} + K_t\, \varepsilon_{t,R} \end{aligned} \tag{10}$$

with $K_p + K_\varphi + K_t = 1$ to keep the F_L and F_R values in the range [0, 1].

The three user defined gains K_p, K_φ, K_t, control the correction policy. This allows the user to profile his transfer function (assisted by a professional trainer or sport medicine doctor) to focus on one specific asymmetry cause.

If the *F* value overcomes a user defined training threshold *thr*, the leg will be advised by a vibro-feedback. We control the motor activation time proportionally to the *F* value. Training threshold *thr* can be lowered when better performance are achieved. Figure 10 shows the flowchart of the proposed asymmetry evaluation and correction procedure.

7. User test

We tested the proposed system with two professional and four amateur runners (five males and one female ranging in age from 24 to 37.). Only two of them had previously experienced pace meter devices. They read a short handbook about how to perform the static calibration and then they were asked to calibrate the system without assistance and to run in our test track (i.e., a strait path 500 m long) as they do while training. In a second run, they were asked to induce some asymmetries in lower limbs. All the acceleration data were stored for future elaborations. After the trials we interviewed the participants for their opinion. The post experiment questionnaire featured five-point Likert scale questions (1=most negative; 5=most positive) to evaluate: ease of use, feedback recognition, correction intuitiveness, and system acceptance in training.

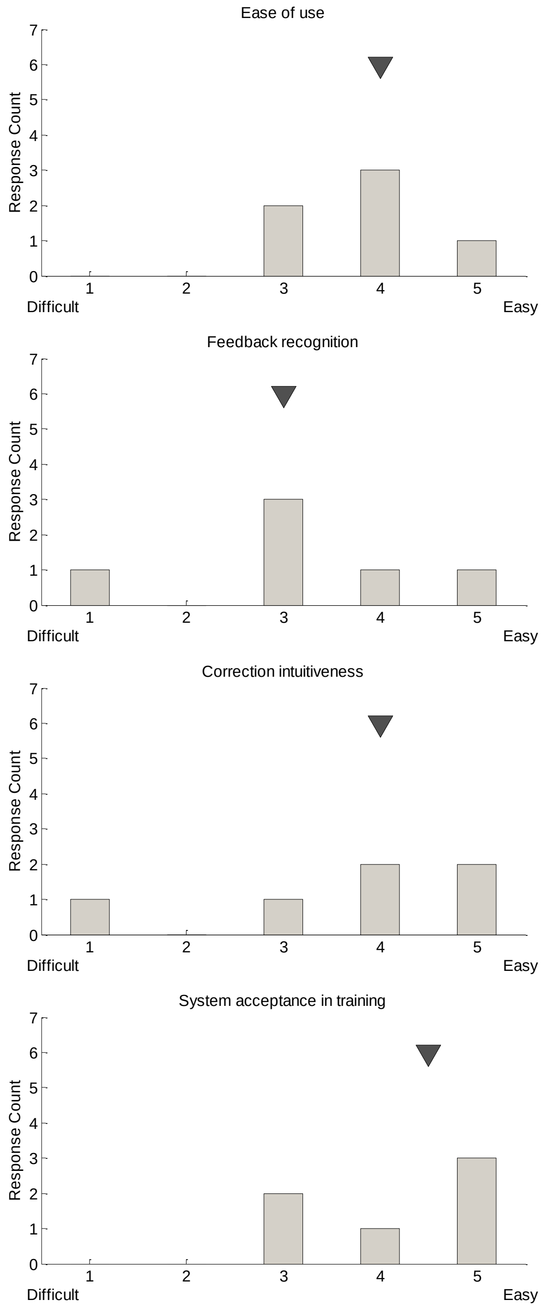

Fig. 11. Survey response histograms for post experiment questionnaire. Median values for each condition are shown as triangles

Figure 11 shows the responses of the users for the questionnaires. As to ease of use, the testers acknowledged the wearability and the handiness of the system although it is still in a prototype state (median value 4). As to feedback recognition, all the users replied positively but one, probably because he did not wear correctly the shorts and the vibromotors were not properly in contact with the skin (median value 3). The users deemed the vibrofeedback quite effective to signal the asymmetry (median value 4), with the exception of the user who did not recognize the feedback. Finally all the users were inclined to accept the proposed system in training (median value 4.5) and made themselves available for further extensive tests. All the users showed the interest in having a software tool to review, after the training, the collected data and the evaluated asymmetry parameters.

8. Conclusion and Future Work

We designed a wearable low cost device to improve lower limb sport training using active vibrofeedback. The system consists of a pair of spandex shorts which embed a processor unit, 2 accelerometers and 2 vibro motors. The accelerometers are located in proximity to the knees and are used to measure in real time the tri-axial accelerations for each leg. We present a novel algorithm to compute asymmetry from the 3 axial sensor data continuously refining the calibration during running to take in account shorts assessments. We define a novel asymmetry metric based on foot impact, cadence phase error between legs, and transversal knee moments. We use this metric to activate vibro feedback to the specific leg for correction. Due to its cost and simplicity, the system can support professional sports people to improve their performance while reducing joint problems, while occasional athletes can avoid bad kinematic and dangerous movements. In both cases our system can improve the muscular energy transfer by reducing ineffective movements and also creating the correct muscular memory. The proposed system can be also useful in other medical appliances such as rehabilitation. User test demonstrated that: i) the device is easy to wear and to use, (ii) the corrective vibro feedback is easily detected, (iii) the user is positively influenced by the vibro feedback stimulus, (iv) the users is generally inclined to accept the proposed system in training. As future work we plan to further study the vibro feedback signal and detail with statistic metrics its influence on user performance.

References

1. Badler, N., Lee, P., Phillips, C., Otani, E.: The Jack interactive human model. Concurrent engineering of mechanical systems, Vol. 1, 179-98. (1989)

2. Radin, E.L., Martin, R.B., Burr, D.B., Caterson, B., Boyd, R.D., Goodwin, C.: Mechanical factors influencing cartilage damage. Osteoarthritis: Current Clinical and Fundamental Problems, J.G. Peyron, Pans, France, 90-99. (1985)
3. Eng, J.J., Winter, D.A.: Kinetic analysis of the lower limbs during walking: what information can be gained from a three-dimensional model. Journal of Biomechanics, Vol. 28, No. 6, 753-758. (1995)
4. Nike+. [Online]. Available: http://www.nikeplus.com/ (current April 2011)
5. Suunto. [Online]. Available: http://www.suunto.com/en-us/products/pods-belts/suunto-foot-pod-mini/ (current July 2011)
6. Xsens. 3D Motion Tracking by Xsens. [Online]. Available: http://www.xsens.com/en/general/mtx/ (current April 2011)
7. Vicon MX. [Online]. Available: http://www.vicon.com/products/viconmx.html/ (current April 2011)
8. Ferrari, A., Cutti, A., Garofalo, P., Raggi, M., Heijboer, M., Cappello, A., Davalli, A.: First in vivo assessment of 'Outwalk': a novel protocol for clinical gait analysis based on inertial and magnetic sensors. Medical and Biological Engineering and Computing, Vol.48, No.1, 1-15. (2010)
9. Percoco, G.: Digital close range photogrammetry for 3D body scanning for custom-made garments. Photogrammetric Record, Vol. 26, No.133,73-90. (2011)
10. Fiorentino, M., Uva, A.E., Foglia, M.M.: Self calibrating wearable active running asymmetry measurement and correction, Control Engineering and Applied Informatics, Vol. 13, No. 2, 550–554. (2011)
11. Hamill, J., Bates, B.T., Holt, K.G.: Timing of lower extremity joint actions during treadmill running. Med. Sci. Sport and Exercise, Vol. 24, No. 7, 807-813. (1992)
12. Williams, K.R., Snow, R., Arguss, C.: Changes in distance running kinematics with fatigue. Int J Sport Biomech, Vol 7, 138–162. (1991)
13. Siler, W.L., Martin, P.E.: Changes in running pattern during a treadmill run to volitional exhaustion: fast versus slower runners. Int. J. Sport. Biom, Vol. 7, No. 12-28. (1991)
14. Hanley, B., Smith, L.C., Bissas, A.: Kinematic variations due to changes in pace during men's and women's 5 km road running. International Journal of Sports Science and Coaching, Vol.6, No. 2, 243-252. (2011)
15. Brüggemann, G.P., Amdt, A.N.: Fatigue and lower extremity function. In Proceedings of 8th Congress CSB, Calgary. (1994)
16. Nicol, C., Komi, P.V., Marconnet, P.: Fatigue effects of marathon running on neuromuscular performance. Changes in force, integrated electromyographic activity and endurance capacity, Scand J Med Sei Sports, Vol.1, 18-24. (1991)
17. Olney, S.J., Griffin, M.P., Monga, T.N., Mcbride, I.D.: Work and power in gait of stroke patients. Arch Phys Med Rehabil, Vol. 72, 309–314. (1991)
18. Haddad, J.M., Van Emmerik, R.E.A., Wheat, J.S., Hamill, J., Snapp-Childs, W.: Relative phase coordination analysis in the assessment of dynamic gait symmetry. Journal of Applied Biomechanics, Vol. 26, No. 1, 109-113. (2010)
19. Kuo, A.D., Donelan, J.M.: Dynamic principles of gait and their clinical implications. Phys Ther, Vol. 90, 157–174. (2010)
20. Arduino. [Online]. Available: http://www.arduino.cc/ (current February 2011)
21. Adafruit. Data logging shield for Arduino. [Online]. Available: http://www.adafruit.com/. (current February 2011)
22. Analog Devices Inc ADXL330 handbook. [Online]. Available: http://www.analog.com/. (current February 2011)
23. Yang, U., Jang, Y., Kim, G.J.: Designing a vibro-tactile wear for 'close range' interaction for VR-based motion training. Proceedings of international conference on artificial reality and telexistence, 4–9. (2002)

24. Amasayc, T., Zodrowa, K., Kinclb, L., Hessb, J., Kardunaa, A.: Validation of tri-axial accelerometer for the calculation of elevation angles. International Journal of Industrial Ergonomics, Vol. 39, No. 5, 783-789. (2009)
25. Yanling, H., Zhiping, L.: Analysis and calibration on installation errors of accelerometer in GFSINS. In Proceedings of the 21st annual international conference on Chinese control and decision conference (CCDC'09), 2097-2101. (2009)
26. Camps, F., Harasse, S., Monin, A.: Numerical calibration for 3-axis accelerometers and magnetometers. Conference on Electro/Information Technology (EIT '09), IEEE International, 217–221. (2009)

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