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Paper:

A Mechatronic Pneumatic Device to Improve Diastolic Function by Intermittent Action on Lower Limbs

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Individuals with walking disability, as a result of pathological conditions or traumas, show a reduction in left ventricle end diastolic volume (EDV). In fact EDV is closely related to the blood pressure gradient between the postcaval vein and the right atrium which, during locomotion, is partially due to the calf veins squeezing caused by the rhythmic contraction of the triceps surae and the crushing of the sole of the foot's veins. In this study, a mechatronic device was applied to nineteen healthy voluntary participants' lower limbs to test cardiodynamic response to a mechanical intermittent stimulation. The device consisted of inflatable bladders embedded in two shells and acting on the skin of the calf and foot of both legs. The pressure trend on the legs was regulated by a portable programmable logic controller. During the compression protocol to the legs, which involved some sequences of activation-deactivation following a peristaltic compression having a caudal-rostral trend, EDV, assessed by the impedance cardiography technique, increased of about 10% up the pre-test value. The legs compression protocol imposed by means of our pneumatic device might be useful to avoid the negative consequences for cardiovascular performance caused by deconditioning status linked to walking disabilities.

Keywords: pneumatic flexible actuators, left ventricle end diastolic volume, mechatronic devices for rehabilitation, thoracic electrical bioimpedance, walking disability

1. Introduction

Long lasting physical performances are mainly attributable to the ability of the cardiovascular apparatus to deliver oxygen for exercising muscles [1]. Physiologic enlargement of cardiac chambers is the main cardiocirculatory adaptation which may occur during a regularly repeated, aerobic exercise schedule [2]. In their review article Mihl et al. [3] highlighted that, as the consequence of the aerobic training, the heart is submitted to an elevated venous return from exercising muscles which results in a volume overload of the left ventricle. The aerobic-trained left ventricle adapts itself to this volume overload by increasing both internal diameter and wall thickness. Therefore, the aerobic-trained heart develops eccentric hypertrophy which reaches an increase in the end diastolic volume of left ventricle (EDV).

Whereas regular aerobic training is known to increase cardiomyocyte function and dimension [4], the response to detraining over a period of only 3 weeks was found to result in a reduction of training-induced ventricle hypertrophy up to 20% [5]. In a recent paper Bringard et al. [6] found that, after 35 day bed rest, upright incremental exercise showed lowered both stroke volume (-44.3%) and cardiac output (-45.1%), and VO_2max which was 38.6% lower than before bed rest. From the above considerations it may be argued that sustained physical performance, which requires adequate oxygen delivery for contracting muscles, is seriously compromised when a period of bed rest or of reduced ability to walk occurs, and the "primum movens" of this functional impairment ought to be looked for in the EDV reduction.

EDV is closely related to blood pressure gradient between postcaval vein and right atrium. Therefore, during walking an increase of the vein-to-atrium blood pressure gradient may occur due to calf veins squeezing by the rhythmic contraction of triceps surae muscles, and this increases blood flow from leg veins towards postcaval vein. So, EDV rises due to ventricles filling increase. As a logical supposition, in subjects with impaired capacity of walking a good diastolic function can be maintained, despite not moving legs, by the application to lower limb of a mechanical stimulation operated by actuators. These devices, in replacing striate muscle pump on limbs veins, may recover venous return to the heart in these patients, thus restoring end diastolic filling pressure of ventricles [7]. In this way, a cardiac output compensation would occur and aerobic capacity would be

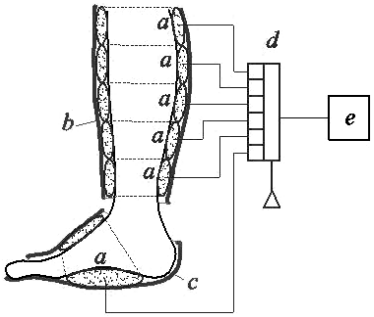

Fig. 1. The scheme of the device. a: six inflatable bladders, b: calf shell, c: foot shell, d: group of on-off 3-way electro-pneumatic valves, e: programmable logic controller.

Fig. 2. The bladders of the device: on the left a single bladder, on the right a group of assembled five bladders for the calf.

restored/maintained.

The application of a mechatronic device is proposed here. This device is made-up of two main parts: the first is represented by a mechanical system with a certain number of customised flexible pneumatic actuators assembled in an anatomic structure, the second is a device to assess thoracic electrical bio impedance (TEB) from which beat-to-beat changes in EDV can be indirectly acquired in a non invasive way. The mechanical actuating system produces several compression trends on legs, generating a pressure profile, variable in amplitude, frequency and sequence. The system allows for an operator to adjust easily the pressure law on bed-resting subject, on the basis of periodically assessed EDV values by TEB signals.

2. Methods

2.1. Subjects

A group of 19 healthy male participants was engaged in these experiments. Mean±SD values of age, weight and height were respectively: 22.6±4.5 years, 63.7±12.2 kg and 173.2±7.8 cm. All participants received information about the aims and the procedures of the study which was performed according to the Declaration of Helsinki and conformed to Good Publishing Practice in Physiology. The study was approved by the hectic committee at the university of Cagliari. All subjects read and signed informed consent forms and their privacy was protected.

2.2. The Mechatronic Device

A basic scheme of the device is depicted in **Fig. 1**. It performs a definite massaging action on specific zones of the lower limb, according to particular time laws. Moreover, it meets anatomical requirements by generating distributed forces along the entire contact surface.

Positive effects on the cardio-circulatory condition are expected on the basis of a preliminary experiment where suitable pressure distributions was imposed on defined lower limbs zones [7]. The device is provided with six inflatable bladders, included into two shells, and acting on the skin of the calf and foot. The shells are a sort of sleeves, flexible but inextensible, so self-adapting to the shape and size of the limb but also avoiding stretching and therefore directing all the pneumatic energy towards the limb.

The control system can be arranged with several levels of complexity; basically it includes a group of on-off 3-way electro-pneumatic valves and a programmable logic controller, i.e. a PLC.

The bladders are the actuators that transform the pneumatic energy into a mechanical action on the limbs, and are a primary element of the system. **Fig. 2** shows one single bladder and a group of five that were used for our application.

In order to optimize the performance, it is convenient that the air energy is directed as much as possible towards the biological tissues of the limb, rather than used to deform the materials of the device structure. To this aim, the bladders should be able to expand with limited or null stretching of the wall and this was achieved both by proper shaping of the cell and adopting a material which is compliant but, of course, air-tight.

Another key element of the device is the shell to which the bladders are attached. Its shape must be adaptable as much as possible to the anatomy of the limb, in order to minimize dead volumes that would affect the dynamic performance. On the other side, the expansion of the bladders must occur only on the limb side, with no stretching of the supporting shell. In order to meet these demands, the shells were realized with proper material which is mostly flexible and inextensible.

Finally, the pneumatic circuit is important, since it must provide the supplying of the bladders with given dynamic performance; this required an accurate definition of the general layout and a proper choice of the components.

Figure 3 shows the prototype realized and used in these tests. It is made up of two separate parts, one for the calf and one for the foot. Each part is realized as a flexible sleeve, to which the bladders are attached. Particular care has been dedicated to the choice of the materials. The bladders have been realized with the Medical Windtex® (Windtex Vagotex S.p.A., Italy), which is made up of three layers (polyester, polyurethane, polyester); this tissue is elastic and quite yielding, air tight, biocompatible and able to maintain its characteristics for long time. The external sleeves have been realized with a three-laminated

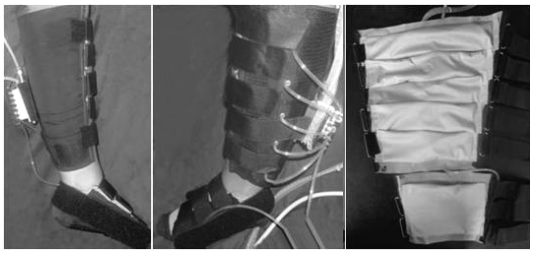

Fig. 3. The lateral sides with the air supply and the internal view of the prototype showing the pneumatic bladders.

tissue (Aqualight 160D®, D&D SNC, Italy), which is air tight, inextensible but very flexible and possesses very high resistance. The sleeves are provided with some Velcro fasteners (three for the foot and five for the calf) that allow them to be well adapted to the shape and size of the limb, in such a way as to reduce to the minimum the gap and hence the bladder dead volumes.

A group of six miniaturized three-way electropneumatic valves (FESTO, model MHP1-M1H-3/2-M3, Italy) are attached to the calf shell by means of adhesive Velcro stripes. Six polyurethane tubes connect the valves to the bladders passing through eyelets inserted into the shell wall. This allows reducing to the minimum the pneumatic layout, so achieving the highest dynamic performance of the pneumatic part.

A portable PLC controls the device imposing the pressure trend in the flexible actuators interacting with the participants. The PLC basic unit was a SIEMENS LOGO 24 RC (Siemens, Italy), provided with two digital I/O modules of the DM8 24 model and DM16 24 model.

The appropriate levels of air pressure were defined by preliminary tests, in order to preserve the comfort of the tested subject. It is to be noted that the operating pressure sensibly drops inside the flexible actuators, with respect to supply regulated pressure, reaching values in the order of one magnitude lower, due to the system dynamics and the plant characteristics. Taking this into account, it was possible to impose, for each lower limb, compression protocols in terms of pressure and spatial-time distribution which involved sequences of activation-deactivation following a peristaltic compression, having a caudal-rostral trend. The logical scheme which was chosen in these experiments was to alternate the activation of the two sleeves to simulate the muscle pattern during walking.

2.3. Inflating Tests Design

Before monitoring cardiovascular response during the activation of the device, each participant was monitored at rest for 3 min to collect baseline values of TEB variables. Then, the mechatronic device was positioned on the legs of a seated participant, with his extensor and flexor leg muscles voluntarily relaxed (**Fig. 4**). Once the preparation was completed, all volunteers underwent a compression-relaxation protocol.

As indicated in **Fig. 5**, The actuators were numbered starting from the foot plant actuator (act1) and coming up

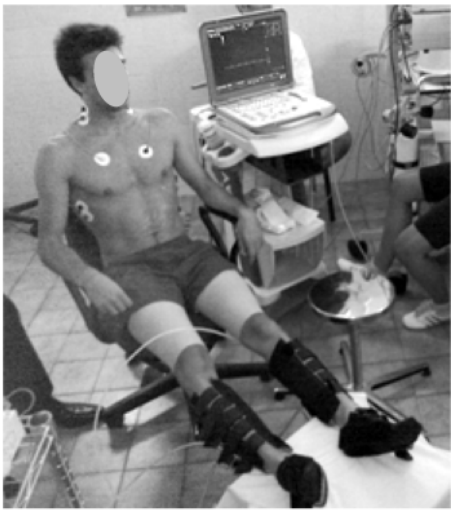

Fig. 4. The mechatronic device during tests on a patient monitored by TEB device. Elastic stocking are on thighs.

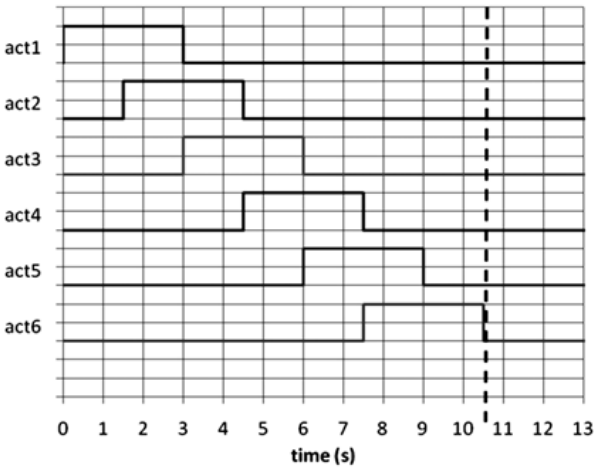

Fig. 5. The time sequence of the actuators command signals. Broken vertical line indicates the half-cycle end.

to the higher one (act6). The time interval activation for each actuator was of 3 s, with a delay between each of 1.5 s. The half-cycle for one boot lasted for 10.5 s, with a delay of 3 s between the two boots; thus the entire cycle including both boots was 24 s.

The inflating relative pressure reached the maximum value of 0.3 bar. On each thigh an elastic containment stocking was worn to reduce venous distension coming from the increase of blood flow during mechanical actuation.

2.4. Hemodynamic Assessment

Hemodynamic variables were monitored and recorded beat-to-beat by means of an impedance cardiography device (NCCOM 3, BoMed Inc., Irvine, USA) which allows for continuous non-invasive assessment of the TEB throughout all the phases of the protocol [8].

The impedance device was connected to the participant by eight ECG disposable electrodes (**Fig. 6**). Two pairs

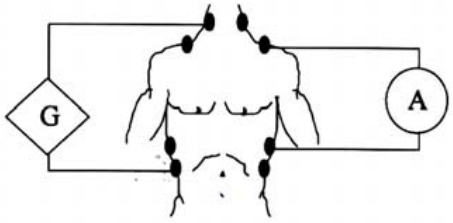

Fig. 6. Electrical circuit for toracic bioimpedance.

were thoracic and cervical electrodes to inject a constant current (2 mA, 70 kHz) from the generator (G), whereas two other pairs were sensing electrodes placed above the cervical and below the thoracic pairs and tie up the amplifier section (A) [9]. **Fig. 4** shows both cervical and thoracic pairs of TEB electrodes placed on the right side of a testing subject, while the two electrodes placed on his pectoral muscles served for the ECG assessment.

The electrical signals were recorded by a digital chart recorder (AD Instruments, PowerLab 8sp, Castle Hill, Australia) [10]. Beat-to-beat values of TEB at the end of cardiac diastole ($Z0$) were acquired and $Z0$ was utilized as an inverse index of EDV [11].

The rationale for this choice is shortly presented here so on. Considering the electrical resistors which all together constitute an electrically equivalent model of the thorax, most of they are of solid consistence (muscles, bones, cartilages, vessels, connective tissue of lungs) which do not vary their mass suddenly.

On the contrary, lung air content and thoracic liquids content are subjected to cyclical changes. So, changes in equivalent electrical impedance of the thorax, i.e. $Z0$, refer about a like sinusoidal oscillation which shows two harmonics: the main one corresponding to breathing phases and the secondary one, with a lesser amplitude, corresponding to cardiac cycle phases [12]. By utilizing a common electronic band-pass filter the main respiratory harmonic of $Z0$ can be excluded and, in this way, observed changes in $Z0$ only depends on changes in intrathoracic liquids volume [13]. These in turn depend on both blood volume into heart and thoracic vessel and on extra-vessel water volume. It is expected that, in a healthy subject, extra-vessel water volume does not change significantly, thus $Z0$ ought to depend essentially on beat-to-beat variations of thoracic blood volume which reach its relative peak when ventricles get to end diastolic volume. Since blood is a more current conducting tissue, an increase in end diastolic volume induces a corresponding reduction in $Z0$ [14].

During experimental sessions, in each subject $Z0$ values were assessed just before beginning the test ($Z0$-basal) and at the end of the test ($Z0$-test). During tests ECG traces were also assessed, so as required by safety rules during human trials. By utilizing a commercial statistic program (MedCalc Software bvba, Ostend, Belgium), assessed values of $Z0$-basal were compared with $Z0$-test values using the Students' t test for paired samples and a value of $P < 0.05$ was considered as significant.

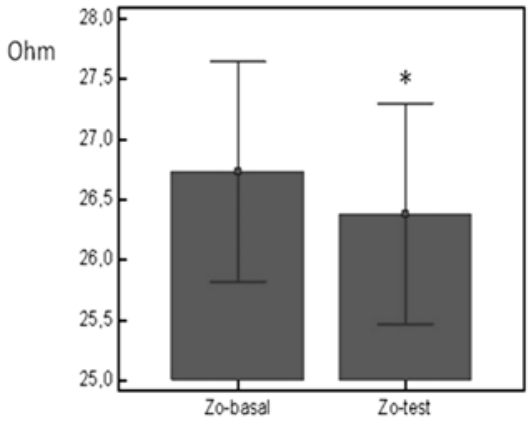

Fig. 7. Mean values of the thoracic electrical impedance.

3. Results and Discussion

The tests results are shown in **Fig. 7**. The histograms represent the mean values of the thoracic electrical impedance, in Ohm, respectively before legs compression session ($Z0$-basal) and just at the end of the compression session ($Z0$-test).

In the graphs of **Fig. 7** vertical bar in each column represent SEM and (*) indicates the P value. As it can be seen in the graph by the columns height, $Z0$-test values showed a significant reduction (-1.3%, $P < 0.0002$) with respect to $Z0$-basal values.

The basic idea of stimulating the circulation by means of some compression of legs and abdomen is not new [15] up enough to be a good therapeutic aid for patients claudicants [16, 17], and most devices include anti-G suits [18], anti-shock trousers [19], neuromuscular stimulation and pneumatic devices [20, 21]. Unfortunately the commercial devices do not always take into the account the mechanism of physiological action, and are not well focused on the main aim to accomplish a therapeutic methodology for the recovery of heart diastolic function. In order to reduce impairment of venous return, numerous attempts have been developed to replace ineffective muscle pump on lower limbs veins. At present, compressed air is used in most biomechanical applications with deformable actuators [22–24]. This solution is particularly suitable for systems interacting with human being for the following reasons: pneumatic actuators are usually lightweight, thus suitable for "wearable" devices; compressed air is clean and easily available in appropriate environments; pneumatically operated devices usually have low mechanical impedance, which is favourable for the accomplishment of efficient control techniques [24–29].

Delis et al. [30], in healthy subjects, supplied an intermittent pneumatic pressure device by 0.16 bar on the foot and 0.24 bar on the calf respectively. The blood outflow from both superficial femoral and popliteal veins was assessed. In this paper the pressure supply level was 0.3 bar with a frequency of 3 impulses/min, the same as in the Delis et al. study where mean volume flow from these superficial veins was about 7 time higher than at rest.

The mechatronic device tested here showed good possibilities to substitute physiological features in producing sufficient venous return for avoid EDV loss in bed rest or impaired of walking subjects. In fact, it implements an easy method to assess EDV: the TEB method, and a good feedback for this possibility arose from the observed statistically significant reduction in $Z0$ just after a leg compression test which lasted less than 30 s. Several previous studies support, both in animal and in humans, reciprocal and consistent relationship between $Z0$ and EDV. Luepker et al. [31], in anesthetized dogs in which the thoracic extra-vessel fluid was maintained at a constant, found that changes in central blood volume (CBV) were highly correlated with $Z0$ changes ($r = -0.98$). They also calculated a linear regression equation in which it appears that each unitary increase in CBV induced a $Z0$ decrease of about 50%. With regards humans, Okutani et al. [32], while applying head up tilt manoeuvres in healthy subjects, showed that when head up was reached the $Z0$ rose by up to +9.5% than the pre-test value. Moreover, legs venous occlusion while recovering supine position showed that $Z0$ fell but its final value was higher than that recovered without occlusion. Since in these subjects legs venous occlusion reduced EDV, the fact that $Z0$ correspondingly did not recover pre-test, lower values demonstrates that it is inversely correlated to a reached EDV. This latter conclusion is also supported by the observation that, passing from standing to supine position, a group of healthy subjects showed ultrasonographic images of left ventricle which indicates a 24% EDV increase circa while $Z0$ fell by about 7% or a reduction of about one third of $Z0$ at each unitary increase in ventricle volume (personal observation). Recently [33] it has been found that during passive recovery following repeated bouts of supramaximal exercise at a cycle ergometer, $Z0$ progressively increased ($+1.6 \pm 0.4\ \Omega$ at the 10th minute of recovery) since the absence of pedalling after a strenuous cycling exercise may result in serious blood pooling into leg veins [34]. On the contrary, when in the same experiment an active recovery was performed $Z0$ did not increase ($+0.05 \pm 0.3\ \Omega$ at corresponding time points of passive recovery period), and the difference between these two kinds of exercise recovery was statistically significant ($P < 0.05$).

Our results show that the applied compression protocol induced a $Z0$ reduction of about -1.3% with respect to the pre-test value. Considering that: i) in a previous study of Concu et al. [35] in which distance runners performed a cycle ergometer incremental exercise, their highest reduction in $Z0$ (-4.5%) was reached when workload was 50% of that maximum; ii) in another study of Schairer et al. [36] a group of endurance athletes who also performed an upright bicyle test at about 50% of maximum workload (their heart beat was 130/min) the EDV shoved its maximum increase of 33 ml which was +23% of corresponding rest value; iii) speculating now with a proportional approach it can be deduced that, if -4.5% $Z0$, from the study of Concu et al., corresponds to +23% EDV, from the study of Schairer et al., then -1.3% $Z0$ (in the present research) ought to correspond to an extrapolated EDV increase of about 10%.

An analytical function, as in Eq. (1), can be written between the variations of $Z0$ and EDV, starting from the following Sramek-Berntein's equation [37]:

$$Sv = \left[\frac{1}{Z0}\frac{dz}{dt}\Delta t V^*\right] \quad . \quad . \quad . \quad . \quad . \quad . \quad . \quad . \quad . \quad . \quad (1)$$

In this equaton Sv is the left ventricle stroke volume at each heart beat (ml), $Z0$ (Ω) is the electrical correlate of the end diastolic volume in the left ventricle (EDV), dz/dt (Ω/s) is the electrical correlate of the left ventricle ejection velocity, Δt (s) is the ventricle ejection time corresponding to the systole duration, V^* is the volume of the participating thoracic tissues to the electrical impedance (ml) which is related with the ventricle hydraulic capacity [38]. Note that the variables dz/dt, Δt and V^* are considered constant when a given subject stay at rest and mentally calm, as occurred in our participants. The variable EDV is the sum of Sv and of the blood residual volume in the ventricle after the systole (V_R), then:

$$EDV = Sv + V_R = \left[\frac{1}{Z0}\frac{dz}{dt}\Delta t V^*\right] + V_R \quad . \quad . \quad . \quad . \quad (2)$$

Whereas a changed value of EDV indicated as EDV$'$ (i.e. an increase of this volume due to the legs intermittent pneumatic compression) implies a variation $Z0'$ of $Z0$ as in following Eq. (3):

$$EDV' = Sv' + V_R = \left[\frac{1}{Z0'}\frac{dz}{dt}\Delta t V^*\right] + V_R \quad . \quad . \quad . \quad (3)$$

The variation of EDV can be written as in Eq. (4):

$$\begin{aligned} EDV' - EDV &= \Delta(EDV) \\ &= \left(Sv' + V_R\right) - \left(Sv + V_R\right) \\ &= \left\{\left[\frac{1}{Z0'}\frac{dz}{dt}\Delta t V^*\right] + V_R\right\} \\ &\quad - \left\{\left[\frac{1}{Z0}\frac{dz}{dt}\Delta t V^*\right] + V_R\right\} \\ &= \frac{dz}{dt}\Delta t V^*\left(\frac{1}{Z0'} - \frac{1}{Z0}\right) \\ &= \frac{dz}{dt}\Delta t V^*\frac{Z0 - Z0'}{Z0Z0'} \quad . \quad . \quad . \quad . \quad . \quad (4) \end{aligned}$$

The relative variation of EDV is expressed by Eq. (5):

$$\frac{\Delta(EDV)}{EDV} = -K\frac{\Delta(Z0)}{Z0} \quad . \quad . \quad . \quad . \quad . \quad . \quad . \quad . \quad . \quad (5)$$

where K is expressed as in Eq. (6):

$$K = \left\{\frac{1}{Z0'}\frac{\frac{dz}{dt}\Delta t V^*}{\left[\frac{1}{Z0}\frac{dz}{dt}\Delta t V^*\right] + V_R}\right\} \quad . \quad . \quad . \quad . \quad . \quad (6)$$

This non dimensional expression of the relative variation of EDV is related to the relative variation of $Z0$.

From these latter analytical considerations it can be rea-

sonably supposed that the leg compression protocol here imposed by our mechatronic device might result in a no negligible effect in the light of utilizing it in order to avoid, in bed rest or impaired of walking subjects, the negative consequences for cardiovascular performance. Furthermore possible use of our mechatronic device in cardiac failure patient might be considered in future since a recent paper of Bickel et al., [39] showed that intermittent sequential pneumatic compression leg sleeaves improve cardiac output in patients with congestive heart failure.

Now considering that $Z0$ values could easily be acquired from a remote surveillance medical centre by an ICT platform, and also considering that the home-based telerehabilitation model has been proposed as a promising new option to improve cardiac rehabilitation in heart failure patients [40], our mechatronic device could assume an increasing relevance in the scenery of automation technologies devoted to cardiac patients.

In conclusion, periodic $Z0$ monitoring in impaired to walk subjects who are trained by utilizing our mechatronic leg pressor device, would consent us, in a easy way, to adapt the leg compression training in such a way as to avoid undesirable deconditioning of EDV.

4. Study Limitations

It must be specified that in these experiments no bed rest or impaired to walk subjects were tested by our mechatronic device. However, at the moment there are no indications, arising from scientific literature, about possible differences in EDV behaviour between healthy subjects and in patients unable to walk, in response to the pneumatic leg compression. Thus, our experimental results could be reasonably switched towards bed-ridden subjects. Nevertheless, further evidence from bed-ridden subjects and, in any case, in people with difficulty walking, would be needed to clinically validate for good our mechatronic device in preserving their training-induced increases in EDV.

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