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# **Sensing Technologies in the Evaluation of the Mechanical Properties of the Plantar Flexors by Using the Free Vibration Technique**

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Additional information is available at the end of the chapter

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## **Abstract**

The use of free vibration techniques to evaluate the responsiveness of a one-degree-of-freedom dynamic system linked to *musculo-articular stiffness* (MAS) of the plantar flexors has become an accepted procedure in scientific literature. However, the applicability and widespread use of this parameter have been limited thus far for different issues. On the one hand, the measurement of this parameter requires a noninvasive, nonstress laboratory test, which implies the use of appropriate in-lab infrastructure, ad hoc mechanical devices, and specific signal sensing units. On the other hand, it requires an efficient treatment and processing of the signal data obtained from laboratory tests through different signal filtering techniques and mathematical transformations. In the present manuscript, the fundamentals of the free vibration technique, the measurement device, the adjustment procedure, and finally the software based on java platform are presented in order to show the applicability of the MAS as parameter to evaluate the functional response of one subject based on his mechanical response of the lower part of the body once the whole procedure is addressed and controlled to ensure the reproducibility and the reliability of the test.

**Keywords:** muscle-tendon unit (MTU), musculo-articular stiffness (MAS), ankle joint, software

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## **1. Introduction**

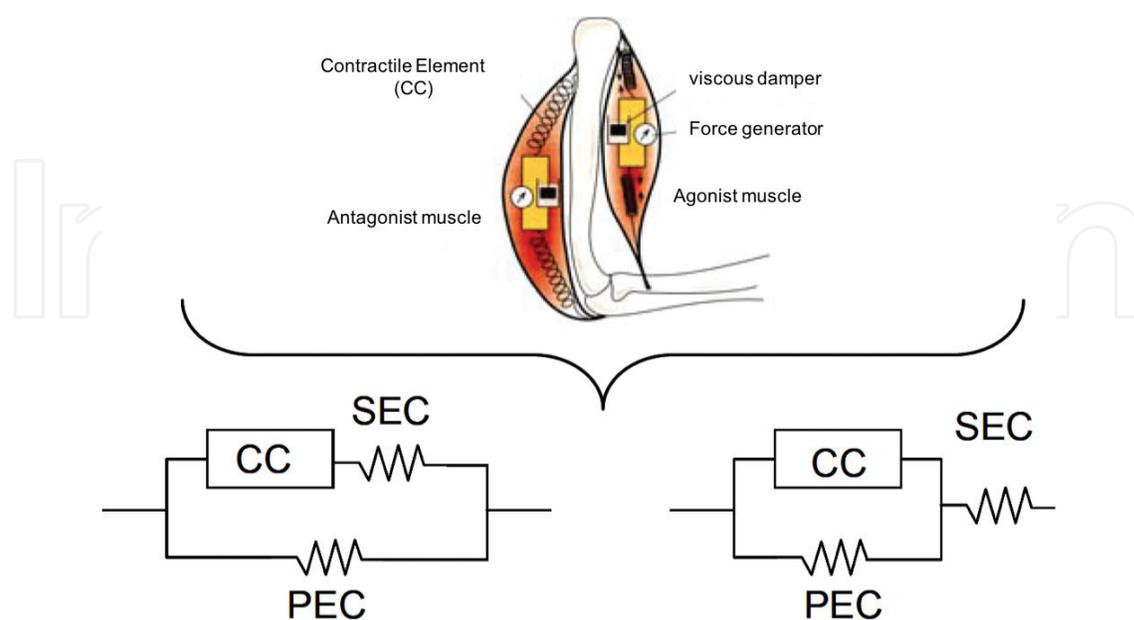
Currently, there are many procedures to evaluate the functional response of one subject based on the theoretical performance of the lower part of the body [1, 2]. In this sense,

biomechanics use different procedures to measure the response of the subject and more specifically in relation to the mechanical response of the muscle. For instance, variations of jump (SJ (squat jump), CMJ (countermovement jump), and DJ (drop jump)) [3] measures, with contact platform or with dynamometric platforms, give information that goes from the elasticity index [1, 4] (general concept that encompasses many elements such as articulation and agonist and antagonist muscles) up to force-velocity curves [5].

Another way to analyze the functional response of the subject from the mechanical response of the muscle through a theoretical model is to resemble the behavior of the muscle to a mechanical model whose variables are known and “easy” to measure through equations that describe that behavior. These equations must consider all the physiological variables associated with it. These variables must go from the structure and the elements that are made of it, taking into account the viscoelastic properties of these elements and the innervation, to the theory or model that justifies the capacity to produce tension (muscular contraction). The latter is in intimate relation with all the other variables.

The elasticity of muscle fibers and tendon fibers plays a very important role in improving the effectiveness and efficiency of human performance. Muscles and tendons have a mechanical behavior very similar to a simple spring, and therefore, their response can be defined by a simple elastic model based on muscle-tendon unit (MTU) [6]. A simple way to represent this model is shown in **Figure 1**.

This figure shows (in a simplified way) the different components that can join in the mechanical response where the serial elastic component (SEC) represents the elastic component in series, the contractile component represents the actin and myosin cross ridge structures (CC), and PEC represents the elastic component in parallel.



**Figure 1.** Schemes of MTU based on springs that represent the behavior of the muscles and tendons. CC, contractile component; SEC, serial elastic component; and PEC, parallel elastic component.

It is important to notice that, to have a direct correlation between the functional response of the subject (necessary in the movements of the daily life and in the sports field) and the viscoelastic parameters of the MTU, this MTU has to play an important role in the development of movements or displacements. For this proposal, the MTU chosen to be analyzed in the laboratory has been the plantar flexor muscles (in particular triceps surae MTU).

This MTU is the main one involved in plantar flexion movement. The plantar flexion is a phase of gait biomechanics and is therefore present in all movements in which displacement has to be carried out.

Therefore, the stiffness parameter is a very important factor related to muscle function [7, 8], to general sport performance [9, 10], and in particular to the performance during fast and slow stretch-shortening cycles (SSC) [8, 10–12].

The measurement of the mechanical properties (mainly stiffness) of MTU is widely reported in literature. In the present work, the procedure based on the free vibration technique will be followed and used.

The applications of the above methods (free vibration techniques) to obtain the stiffness of the MTU gave rise to the consistent use of the term musculo-articular stiffness (MAS) (henceforth  $k$ ) [12]. MAS is a global measure of stiffness that incorporates not only the muscle-tendon structure but also the skin, ligaments, and articular surfaces [13]. Various assessments have demonstrated that MAS is a relevant parameter; higher MAS values are associated with superior muscular performance (e.g., [8, 14, 15]) and higher levels of functional capacity [16, 17].

The stiffness value ( $k$ ) obtained from the experiments will depend on the procedure used to make the measurements, the comparison of these dependences being difficult and not being within the scope of the present work. Some of the proposed procedures measuring  $k$ , associated with the triceps surae, are based on the free vibration technique [14, 16, 18–22]. Among these proposals, in Babic and Lenarcic [19], the oscillation is rotational (rotation of the foot around the ankle articulation), while in the other proposals, the oscillation is associated with the vertical displacement of the lower part of the leg.

So far there are no tools that integrate the different variables that are required to be applicable and therefore can make the widespread use of MAS. In this sense, the following requirements have to be met: On the one hand, a noninvasive and nonstress laboratory test is required. This implies the use of adequate infrastructure in the laboratory, ad hoc mechanical devices, and specific units or devices for signal detection [23]. On the other hand, efficient treatment and processing of signal data obtained from laboratory studies are required through different signal filtering techniques, mathematical transformations, and the use of several software systems that must be integrated into the same tool.

In this chapter, a new software “FLEXOR” is presented as an integrated software solution that implements different aspects as signal processing, data management methods, algorithms, and flexible functionalities to support different adjustments. Researchers and trainers can evaluate the functional response of one subject from the mechanical response based on stiffness (i.e., MAS) of plantar flexor muscle-tendon units (MTU) while enabling the trial and error experimentation with different adjustment parameters and without using different software applications.

In this sense, FLEXOR improves the efficiency and effectiveness of the MAS analysis procedure, integrating all phases of the acquisition, processing, and postprocessing of the signal received from sensing devices in a timely manner.

## 2. Fundamentals

To measure the mechanical properties of the muscle-tendon unit, associated with the ankle joint, an *in vivo* test will be carried out. A device, specifically conceived for this purpose, has been manufactured to induce an oscillation in the lower limb of the leg. By means of an impact of controlled energy, a damped free oscillation is generated, and this oscillation is controlled by the viscoelastic mechanical properties of the triceps surae (Achilles tendon + soleus + gastrocnemius).

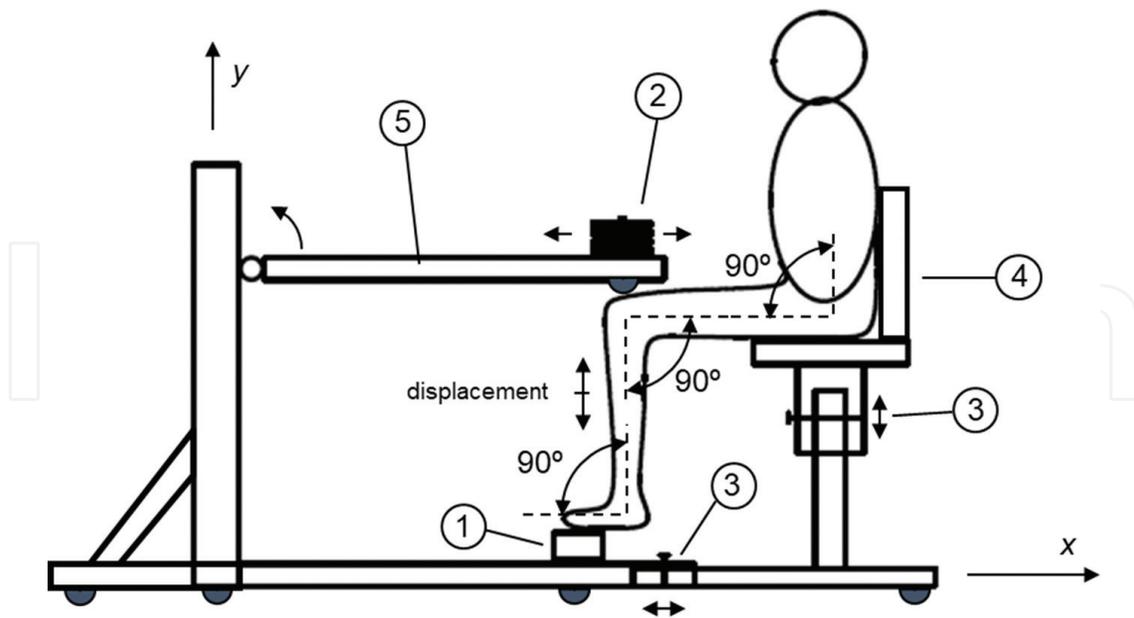
The initial record obtained by the device is the force measured during the oscillation by means of a load cell placed below the foot. This experimental record will be compared with the analytical equation of a one-degree-of-freedom system which is being assumed for the behavior of the oscillating part of the body. The parameters defining this oscillating system will be obtained by a least squares adjustment between the experimental record and the analytical function. The quality of this adjustment will be measured by using the regression coefficient parameter ( $R^2$ ).

The corresponding theoretical models and the least squares fitting procedure are described in detail as follows.

The measurement device has been developed from a previous proposal [18] that uses the free vibration techniques, and this has been detailed in [20]. In the present manuscript, the one-degree damped system is obtained from the vertical displacement of the shank, which will be used as the reference equation of motion. This vertical displacement will be accompanied by rotations of the foot and of the upper leg. Since the frontal part of the foot is simply supported by a plate, the vertical displacement considered in the measurement involves the rotation of the foot around the ankle joint.

The measurement device developed and patented (see **Figure 2**) includes the following main parts to ensure the reproducibility and reliability of the test (These two aspects have been addressed in [24]):

1. Load cell (OMEGA LCM501-10 K; 1000 Hz) registered the reaction force during the oscillation. The load cell data acquisition frequency of 1 KHz granted a satisfactory sampling of experimental data. The load capacity of 1500 N is adequate for the typical force values to be measured in the experiments.
2. The impact necessary to obtain the expected response is generated by the free fall of a known mass and always constant from the same height. In this sense, the same energy is always applied, generating the same impact force. Therefore, the mass  $M_w$  causes the vertical movement of the lower leg as a one-degree-of-freedom system. It should



**Figure 2.** Schematic diagram of the measurement device developed for determining MAS linked to the ankle joint.

be noted that the total oscillating mass  $M$  is actually equal to the sum of  $M_w$  and the one-degree-of-freedom equivalent mass of the lower limb.

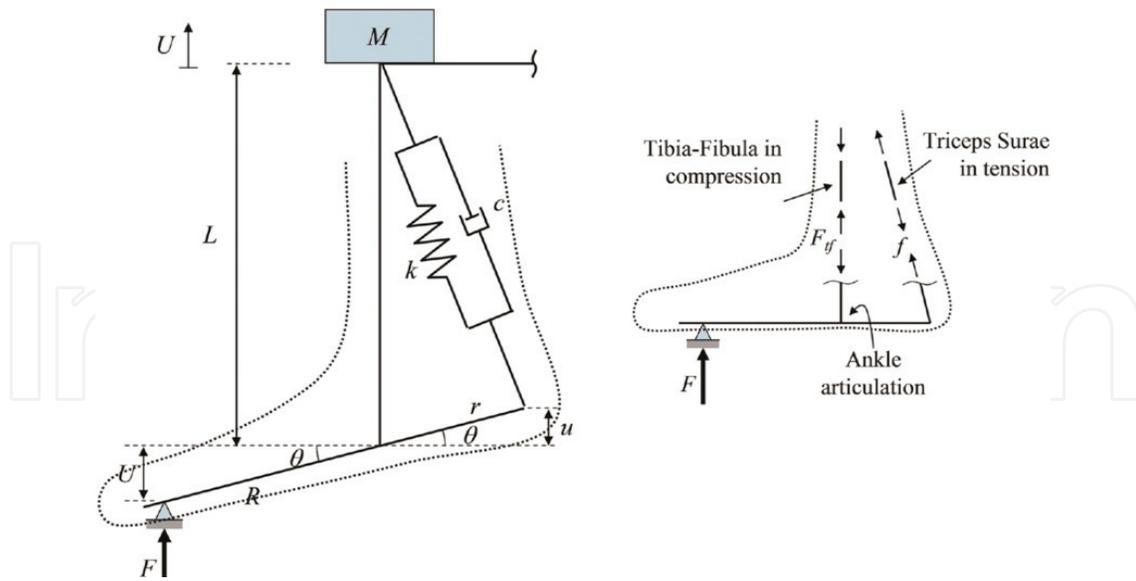
3. The regulation of these two elements (depth and height of the seat) allows the right position of the tested subject. Both degrees of freedom will allow to ensure the neutral position of the knee and  $90^\circ$  of hip and ankle joints.
4. In order to obtain the best response of the system, ensuring the vertical displacement of the shank and avoiding any horizontal displacement at the sagittal plane, a vertical backseat has been designed. Any movement of the head or shoulder in the sagittal plane can modify the response of the system. The subject must keep the position during the test in order not to induce pressure variations.
5. The loading system has been designed to facilitate the development of the test with different loads with different anthropometric features of the tested subject.

Once the impact is applied, the total mass  $M$  (i.e., the concentrated  $M_w$  plus the mass of the lower leg and the one-degree-of-freedom equivalent mass of the lower limb) starts to oscillate vertically (see **Figure 2**). During the perturbation, an arbitrary intermediate position for this movement is displayed in the sagittal plane (see **Figure 3**).

$L$  represents the length of the lower leg,  $R$  is the moment arm of the second metatarsal head, and  $r$  is the Achilles tendon moment arm (both in the sagittal plane).

From the classical Hill's model, the MTU was schematized by means of a spring of stiffness  $k$  and a damper of constant  $c$  [25].

The mass ( $M$ ) in **Figure 3** represents the total one-degree-of-freedom equivalent mass involved in the movement.



**Figure 3.** Diagram of the forces generated during the perturbation. (a) Simplified scheme of an arbitrary position taken in the vibration process. (b) Equilibrium of forces generated at the ankle joint.

The displacements  $U$  and  $u$  indicated in **Figure 3** are related by means of

$$U = \frac{R}{r}u \tag{1}$$

The equilibrium of forces stated around the ankle joint is displayed in **Figure 3**: the force  $F_{tf}$  keeps the tibia-fibula set in compression, while the force  $f$  keeps the plantar flexor muscles in tension. From the condition of dynamic equilibrium of the forces schematized in **Figure 3** with respect to the ankle articulation, it follows

$$(I + m_f r_G^2) \frac{d^2\theta}{dt^2} = FR \cos \theta - fr \cos \theta \tag{2}$$

$\theta$  is the angle (**Figure 3**) defining the instantaneous position of the foot,  $I$  is the momentum of inertia of the foot with respect to its center of gravity,  $m_f$  is the mass of the foot involved in the movement, and  $r_G$  is the distance between the center of gravity of the foot and the center of rotation. The force corresponding to the weight of the leg and concentrated mass does not produce moment with respect to the axis of rotation.

The values of parameters in the left-hand side of Eq. (2) were estimated from [26]; since these terms are much smaller than their counterpart for the right-hand side, Eq. (2) reduces to

$$FR = fr \tag{3}$$

The above equation allows the value of the force  $f$  passing through the line of action of the MTU to be obtained once the value of the reaction force  $F$  is measured by the load cell and the distances  $R$  and  $r$  are precisely determined for the tested foot.

The system shown in **Figure 3** includes two coupled degrees of freedom: the rotation around the ankle joint and the vertical displacement of the vibrating mass. Since the tibia-fibula set was assumed to be infinitely rigid, the vertical displacement of the mass  $M$  coincided with the vertical displacement of the ankle rotation point.

By taking as reference the configuration of the system including the external weight, the dynamic equilibrium condition for the total mass  $M$ , after the impact, can be expressed as

$$M\ddot{U} = -KU - C\dot{U} \quad (4)$$

By disaggregating the system, it can be written as

$$F = -KU - C\dot{U} (= M\ddot{U}) \quad (5)$$

The general solution of the differential Eq. (4) written for an underdamped system is

$$U(t) = e^{-\xi\omega t} (A \sin \omega_D t + B \cos \omega_D t) \quad (6)$$

where the constants  $A$  and  $B$  are determined from the boundary conditions of the problem.

The damping factor  $\xi$  is defined as

$$\xi = \frac{C}{C_c} = \frac{C}{2M\omega} \quad (7)$$

where  $C$  is the damping of the system,  $C_c$  is the critical damping of the system,  $M$  is the total mass involved in the movement, and  $\omega$  is the natural frequency of the vibrating system.

The frequency of vibration  $\omega_D$  of the damped system is

$$\omega_D = \omega \sqrt{1 - \xi^2} \quad (8)$$

By defining the parameter  $\gamma$  as

$$\gamma = \xi\omega = \frac{C}{2M} \quad (9)$$

$\omega_D$  can be expressed as follows:

$$\omega_D^2 = \omega^2 - \gamma^2 \quad (10)$$

The general expression of displacement (6) can be substituted in the equation of dynamic equilibrium (5). It is thus possible to express the total force  $F$  acting on the mass  $M$  as

$$F = (M\ddot{U}) = e^{-\gamma t} (A_F \sin \omega_D t + B_F \cos \omega_D t) \quad (11)$$

The total force  $F_m$  is the sum of  $F$  (load cell) plus the action of the gravity:

$$F_m = F + M_s g = e^{-\gamma t} (A_F \sin \omega_D t + B_F \cos \omega_D t) + M_s g \quad (12)$$

Once these unknown parameters are determined, the values of apparent stiffness  $K$  and damping  $C$  of the whole system can be determined as follows:

$$K = M(\omega_D^2 + \gamma^2) \quad (13)$$

$$C = 2\gamma M \quad (14)$$

Parameters  $K$  and  $C$  can be related to the apparent stiffness  $k$  and damping  $c$  of the MTU. For that purpose, the condition of dynamic equilibrium can be written also for the MTU as follows:

$$f = -ku - c\dot{u} \quad (15)$$

By substituting Eqs. (1) and (3) in Eq. (5) and rearranging the expression, it follows

$$f = -\left(\frac{R}{r}\right)^2 Ku - \left(\frac{R}{r}\right)^2 C\dot{u} \quad (16)$$

Since Eqs. (15) and (16) are equivalent, all terms contained in these expressions must coincide. The following conditions can hence be obtained:

$$k = \left(\frac{R}{r}\right)^2 K, \quad c = \left(\frac{R}{r}\right)^2 C \quad (17)$$

An important aspect from the above-described model is the precise determination of the values of  $r$  and  $R$  for the tested subject.

The separation of the muscle tissue and Achilles tendon components from the global apparent value of the stiffness can be done, following Hill's model. The spring (that corresponds to the global response) includes the spring associated with the elastic behavior of the Achilles tendon (constant value of stiffness) and set of springs in parallel (in accordance with Hill's model [27]), in series with the Achilles tendon, representing the elastic behavior of the muscle.

While the Achilles tendon is assumed to have a constant value of the stiffness ( $k_t$ ), the soleus is assumed to have a stiffness ( $k_m$ ) which is proportional to the load that is being transferred by the system. Thus, the total stiffness of the soleus ( $k_{ss}$ ) can be obtained from a unitary stiffness value ( $k_{ss}$ ) multiplied by the total load ( $f$ ) passing through the MTU:

$$k_m = k_{ss} \cdot f \quad (18)$$

The relationship between the equivalent global stiffness ( $k$ ) and the individual stiffnesses ( $k_{ss}$  and  $k_t$ ) can be easily obtained from **Figure 3**, with two springs in series. On the one hand, for the apparent TS system, the elongation and the associated force are related by the equivalent stiffness  $k$  by means of

$$u(k) = \frac{f}{k} \quad (19)$$

On the other hand, for the components in series, both with the same force ( $f$ ), the following relation applies:

$$u(k_t, k_m) = u(k_t) + u(k_m) = \frac{f}{k_t} + \frac{f}{k_m} \quad (20)$$

Identifying displacements in Eqs. (19) and (20), in order to have an equivalent behavior, an expression of  $k$  in terms of its individual constituents ( $k_m$  and  $k_t$ ) can be easily obtained, and using Eq. (18) in terms of ( $k_{ss}$  and  $k_t$ ),

$$k = \frac{k_t k_m}{k_t + k_m} = \frac{k_t k_{ss} f}{k_t + k_{ss} f} \quad (21)$$

In Eq. (21)  $k$  and  $f$  are considered known. The unknowns in Eq. (21) are the stiffness of the Achilles tendon and the unit stiffness of the soleus, respectively,  $k_t$  and  $k_{ss}$ . They will be evaluated by means of least squares fitting between experimental data and Eq. (21).

It is important to stress that  $k_t$  and  $k_{ss}$  are unknowns of a different nature. While in the Achilles tendon,  $k_t$  is a real stiffness value (measured in force/length, e.g.,  $kN/m$ )  $k_{ss}$  represents the stiffness per unit load in the soleus (e.g.,  $(kN/m)/kN$  or simply  $m^{-1}$ ). For a better physical understanding of the parameters  $k_t$  and  $k_{ss}$ , **Figure 7** represents Eq. (21), showing the nonlinear dependence of the total stiffness of the MTU ( $k$ ) on the total load ( $f$ ) and a saturation level for high values of  $f$ .

The stiffness per unit load in the soleus  $k_{ss}$  represents the slope of the curve at the origin

$$k_{ss} = \lim_{f \rightarrow 0} \frac{dk(f)}{df}, \quad (22)$$

while the Achilles tendon stiffness  $k_t$  represents the horizontal asymptote of the curve for high values of the total force  $f$ :

$$k_t = \lim_{f \rightarrow \infty} k(f) \quad (23)$$

With the two elastic elements (Achilles tendon and soleus) in series, the lower stiffness is the one that controls the apparent stiffness of the TS. With low values of the total transmitted force  $f$ , the stiffness of the soleus ( $k_{ss} \cdot f$ ) is lower than the stiffness of the Achilles tendon  $k_t$ . Thus, the total stiffness  $k$  is controlled by the stiffness of the soleus. By contrast, at higher values of  $f$ , the stiffness of the soleus ( $k_{ss} \cdot f$ ) is much higher than the constant value of the Achilles tendon  $k_t$ . Thus, the total stiffness  $k$  is now controlled by the stiffness of the Achilles tendon  $k_t$ . Finding experimental results close to the horizontal asymptote depends on the total stiffness that can be developed by the soleus. This fact is important as an accurate determination of both values ( $k_t$  and  $k_{ss}$ ) should be carried out, having experimental results for low and high values of  $f$ .

### 3. Application (adjustment procedure)

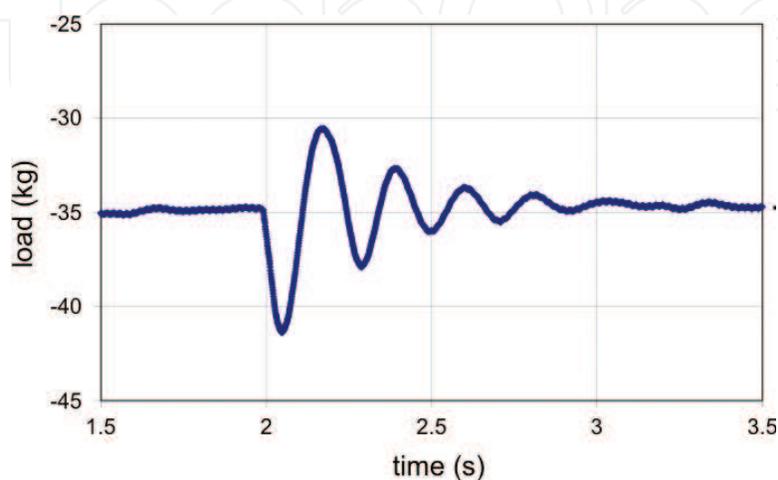
As an example of the whole procedure of measurement and postprocessing of the raw experimental data, a brief description of the process is summarized in what follows from a practical point of view.

First of all, using the device manufactured for the test (see **Figure 4**) applying different external loads on the knee of the subject and after an impact of controlled energy, a load versus time record is obtained. Different loads (placed on the knee of the subject) are used to obtain the full response of the MTU. A typical record is shown below (see **Figure 5**). Notice that the oscillation of the perturbation is only around 1 second in length. This is the value recorded at the load cell, which is placed under the second metatarsal of the foot.

The equation representing the one-degree-of-freedom system, previously introduced in Section 1 (Eq. 11), can be used to fit the experimental part of the curve (in the oscillation range) by



**Figure 4.** Pictures of the measuring device that integrates the different elements (regulation systems, load cell, and integrated software) to obtain the mechanical properties of the plantar flexors by means of the free vibration technique.



**Figure 5.** Force versus time record for a single test.

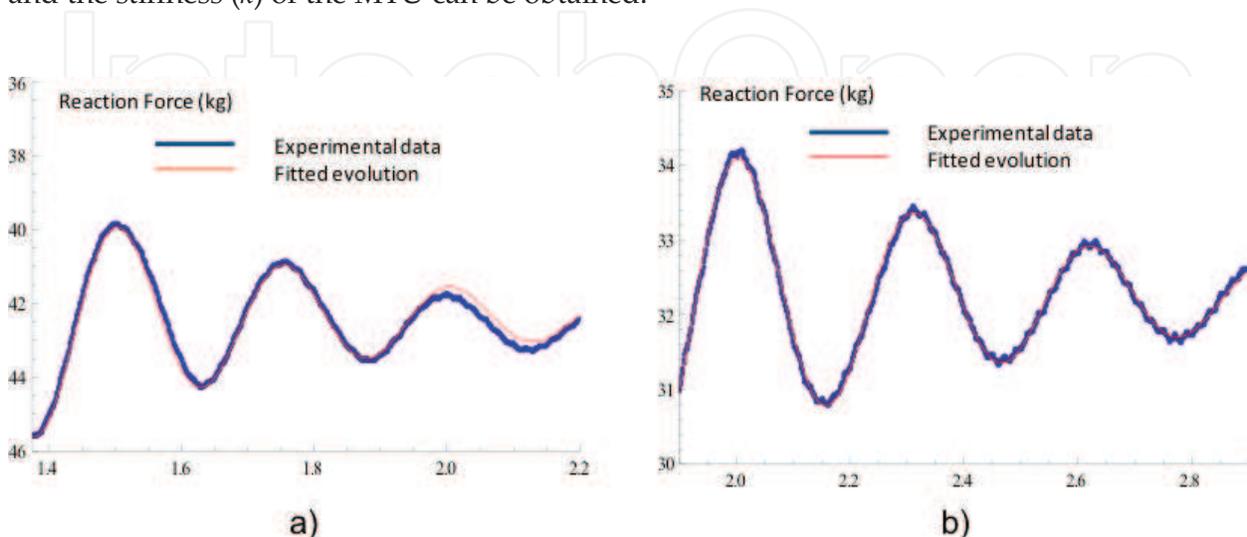
means of a least squares procedure to determine the five parameters included in this equation ( $A_F, B_F, \omega_D, M, \gamma$ ).

This fitting procedure is not straightforward, and several aspects should be taken into account for an accurate determination of the parameters. Three of the five parameters ( $\omega_D, M, \gamma$ ) can be easily initially estimated by simple observation of the curve, because they are associated with the frequency, stationary mass, and damping coefficient, respectively. Once this initial estimation is done, the precise determination of the complete set of values is easily carried out by any standard multivariable least squares routine. For further details of the procedure, see [20], where sensitivity analysis was also performed to check the robustness of the fitting procedure to avoid local minimum solutions. In this first step of the complete procedure, some care has also to be taken when selecting the range of the record to be fitted. The longer the record, the better, but data just after the initiation of the oscillation (just after the impact) have to be discarded and also data near the end of the oscillation need also to be discarded, as the noise is of the same amplitude that the amplitude of the oscillation.

The software, implemented by the authors for the automatic processing of the complete procedure, takes into account all these details for a robust determination of the set of parameters. The software implementation will be briefly summarized at the end of this section.

The experimental behavior of the lower limb of the leg can be mathematically modeled as a single-degree damped system; some examples of the fitting procedure are shown in **Figure 6**. In **Figure 6b**, a secondary higher frequency can be observed, which is due to the natural frequency oscillation of the metallic structure of the device used to carry out the test. Care must also be paid to this fact to ensure that this natural frequency is, at least, one order of magnitude higher than the natural frequency of the physiological part of the oscillation.

For each one of the curves, a pair of values ( $F, K$ ), representing the average force and the global stiffness of the oscillating system (the whole lower limb and accessories), can be, respectively, determined using Eqs. (11) and (13). Using the moment arm lengths of the forefoot ( $R$ ) and rearfoot ( $r$ ) and the moment equilibrium equations given in (Eqs. (16) and (17)), the force ( $f$ ) and the stiffness ( $k$ ) of the MTU can be obtained.



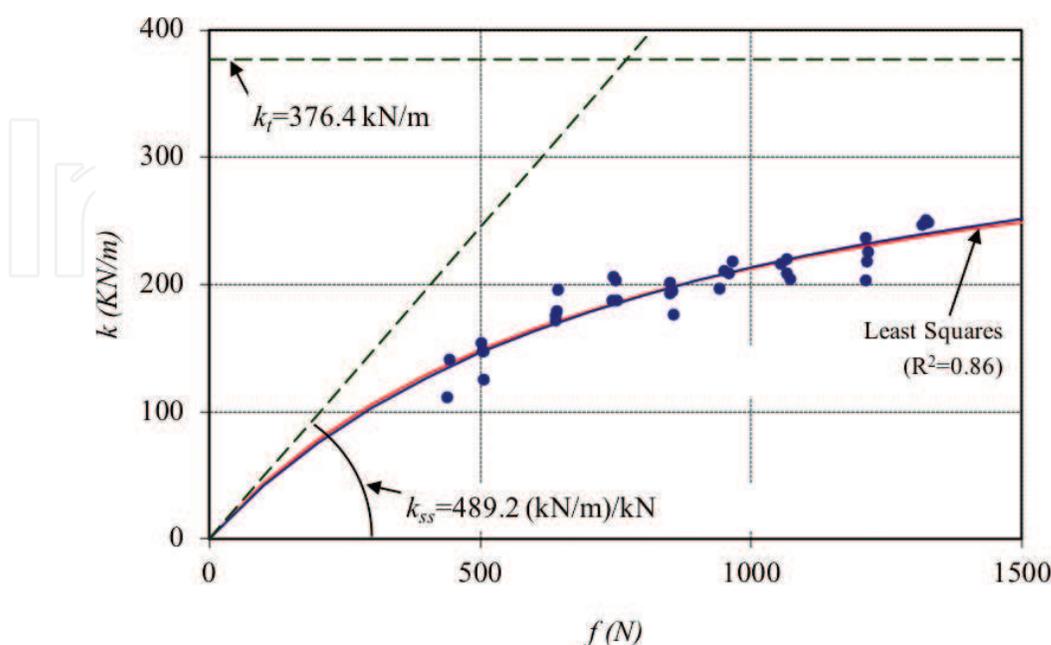
**Figure 6.** Examples of fitted curves with different applied loads. (a) Applied load 30 kg. (b) Applied load 20 kg.

Taking into account that the stiffness of the Achilles tendon can be assumed to remain almost constant with the load passing through it, and that the stiffness of the soleus is a function of the load passing along the muscle, several different weights have to be used to completely determine the mechanical behavior of the system. This behavior, together with the fact that each load is tested more than once (to avoid high dispersion of the results), generates several curves that have to be postprocessed.

This process is tedious, and error-prone, if made manually. The developed software takes all curves and processes all of them in a single click, including quality evaluation of the fitting procedure by means of the regression coefficient parameter ( $R^2$ ) arising from the least squares procedure.

Once the complete set of tests are performed and each one has been processed to obtain the five parameters, a new cloud of processed data ( $f,k$ ) is available, as schematically observed in **Figure 7**. A new mathematical model (see Eq. 21) describes the mechanical behavior of the triceps surae varying with the load passing along it. A new least squares fitting can be done to determine the stiffness of the Achilles tendon ( $k_t$ ) and the unitary stiffness of the soleus ( $k_{ss}$ ). This time, only two parameters are involved in the fitting process, but it is important to notice that each pair of data ( $f,k$ ) is the result of a previous fitting procedure, including some uncertainty in their determination, as the regression coefficient is less than one ( $R^2 < 1$ ).

With respect to this quality indicator ( $R^2$ ), it is also important to stress (see **Figure 7**) that the value of  $R^2$  changes if all individual values are taken into account ( $R^2 = 0.86$ ) or the fitting process is carried out with the average values for each load ( $R^2 = 0.95$ ), although the fitted curve is almost coincident using both alternatives. In both cases the regression coefficient is very close to unity, which means that the mathematical model in this second step of the process also accurately reproduces the physical behavior of the physiological system.



**Figure 7.** Example of the second fitting procedure.

The analyses of the quality of the adjustment procedure for different subjects and different days have been carried out in the previous work [24].

The Achilles tendon stiffness value is associated with the horizontal asymptote of the curve, while the unitary stiffness of the soleus is given by the slope of the curve when  $f \rightarrow 0$  (see **Figure 7**).

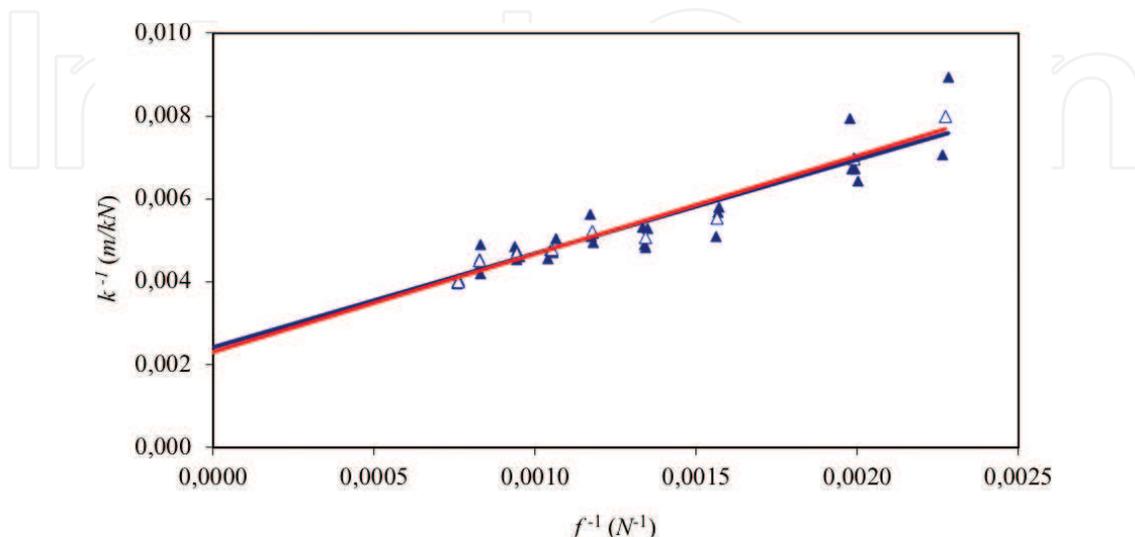
The relationship between the stiffness of the triceps surae ( $k$ ) and the stiffness of the Achilles tendon ( $k_t$ ) and the unitary stiffness of the soleus ( $k_{ss}$ ) can be easily manipulated (taking the compliance ( $k^{-1}$ ) instead of the stiffness ( $k$ ) to give a linear relationship, as can be observed (see **Figure 8**).

With this linear relationship, different robust procedures for linear regression can be used. In [28], the authors used up to four different procedures for linear regression, to check the influence of the fitting method on the results.

This second adjustment step, done by the cloud of points coming from the results of the first test (and first fitting procedure), finally gives the desired mechanical properties: stiffness of the Achilles tendon ( $k_t$ ) and the unitary stiffness of the soleus ( $k_{ss}$ ).

As a summary of the complete experimental process, and previously to introduce the implementation of the developed software to automatize the postprocessing of data, the steps of the process are:

1. The subject is placed in the testing device, and with a certain weight on the knee, and with an impact of controlled energy (on the knee), the lower limb oscillates, and a force versus time record is obtained (see **Figure 5**).
2. This record is repeated for each weight (four or five times).
3. The weight on the knee is changed, and four or five individual tests are repeated.
4. For each record, a first fitting procedure is carried out (see **Figure 6**), obtaining the five parameters ( $A_F$ ,  $B_F$ ,  $\omega_D$ ,  $M$ ,  $\gamma$ ).
5. With these parameters the pairs ( $F$ ,  $K$ ) are obtained, and using the moment arm distances of the foot ( $f$ ,  $k$ ) are finally calculated, one pair ( $f$ ,  $k$ ) for each test.



**Figure 8.** Linear dependency of compliances.

6. With all pairs ( $f, k$ ), a second fitting adjustment is carried out, using stiffness (see **Figure 7**) or using the compliances (see **Figure 8**). This second least squares fitting procedure gives the desired values of the Achilles tendon stiffness ( $k_t$ ) and the unitary stiffness of the muscle ( $k_{ss}$ ).

All this experimental procedure and the associated postprocessing of results are tedious, as different hardware and software are involved in the different steps:

- For the initial record of the force versus time data, an acquisition software (e.g., LabVIEW) and hardware need to be used, taking into account all aspects previously mentioned to try to guaranty the quality of the record.
- These data have to be prepared and introduced in a mathematical software (e.g., Mathematica, MATLAB, etc.) to perform the first fitting adjustment.
- After the manual postprocessing of all tests, a second file is prepared, which also needs to be fitted with the second adjustment procedure.
- All manipulations between software are being done manually with a certain probability of errors.

#### 4. Software FLEXOR

To avoid all these technician manipulations, a software has been implemented to perform the acquisition and postprocessing of data in a single software.

**Figure 9** shows an example of the first-phase fitting (red curve is the experimental data and blue line the representation of the mathematical equation), which is the most time-consuming. This window has all control parameters needed to optimize the time for data processing. The software can read online the load cell (little window at the top right part of the panel software) (see **Figure 9**) and saves all records with a format which allows to recover the different projects to perform additional calculations.

All tests for the different load range used (0–50 kg) in one project are displayed on the left side of the screenshot. All of them have the approval criterion (green check symbol) of the software, which means that the fitting parameter, the regression coefficient,  $R^2 > 0.9$ . The details of the window range adjustment and the details of the parameters used are shown on the right side of the screenshot. In the present case, all tests carried out give rise to excellent adjustments. This procedure, which was previously carried out manually, is now performed with a single mouse click, the software analyzing all introduced records sequentially.

As an example of the potential of the software, the analyst can change manually one or more parameters of the adjustment. In some cases, when the experimental data have not enough quality, the analyst can manually modify any of the five parameters or even the window of adjustment to obtain the best result ( $R^2$  values close to the unit).

Finally, the second phase of the adjustment is displayed, in the same window (not shown in **Figure 9**), and the cloud of points generated from the first phase of the adjustment and final results for the Achilles tendon stiffness and soleus unitary stiffness can be displayed.



Figure 9. Software view.

After performing all calculations, the software makes a report with all data and stores the files for future manipulations.

The software has a flexible design, in order to be adaptable to future requirements, adding new functionalities through a software component-based approach, as well as to provide the information generated in the different studies in an interoperable way, using JSON technology. This enables multi-stakeholder scenarios, where different types of analysts with different FLEXOR installations (e.g., different application servers, platforms, and technologies) are able to conduct joint studies in a remote way, just exchanging FLEXOR proprietary files which represents studies (\*.fxs, \*.fxp).

So far, several studies have been conducted in the lab to analyze the reproducibility of the results comparing the one used in previous procedures and the results obtained by the software. To carry out this comparison, load cell raw data (files) from previous studies conducted using the former procedure have been used. Each file contains the list of points (force, time) that comes from the load cell. When the file is imported in the new software, automatically, it parses the file into a sample and calculates the first-phase adjustment. Importing the list of files that corresponds to a project obtained from one subject, the same study can be simulated (and the procedure) through the software. This approach allows to validate that the software works in a proper way, checking if the results are equal to the previous studies conducted.

The software and the mechanical device conceived for the tests are a practical tool for technicians, researchers, and professionals to make quick and reliable evaluations of the mechanical properties of the muscle-tendon units of subjects. These mechanical properties have a direct influence on the sport performance and thus becoming a very interesting tool to be used in different applications, from rehabilitation to evaluation of sport performance, etc. The software is prepared to be used in any platform in a simple way.

## 5. Conclusions

In the present work, the mechanical properties (mainly stiffness) of the musculo-articular stiffness (MAS) have been obtained by means of the free vibration technique.

For this objective, a measurement device has been conceived and manufactured. This device allows to carry out the test for subjects with different anthropometric characteristics and ensures a high level of reproducibility and reliability of the tests.

The complete postprocessing of the experimental data has been done according to the physiological models of the muscle-tendon unit (MTU). The whole process has been deeply analyzed and implemented in the calculations since this is very sensitive to the influence of other variables being a key point for the calculation of the desired parameter. On the one hand, the reproducibility and the reliability of the MAS depend on the measurement device developed to ensure always the same response of the subject. On the other hand, the high levels of precision in the adjustment process are required as well.

Finally, the study of the mechanical response of the MTU to evaluate the functional response of the subject depends on the applicability of the parameter. In this sense, to facilitate the postprocessing of the experimental data, a software (FLEXOR) has been implemented allowing the adjustment of all variables and an agile data management.

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