Universidad de la Republica

Licenciatura en Biología Humana

Informe de Pasantía

Design and construction of a device for evaluating the contractile capacity of human triceps surae

Diseño y construcción de un dispositivo para la evaluación de la capacidad contráctil del tríceps sural humano

(Manuscrito en elaboración para ser presentado en Journal of Biomechanics)

Estudiante: Sabrina Cervetto

Tutor: Richard Fariña

Orientador de Pasantía: Angel Caputi

Lugar de realización: Departamento de Neurociencias Integrativas y Computacionales. IIBCE

Fecha de presentación: Abril de 2015

Abstract.

The ability of a muscle to develop tension is one of the most important biomechanical properties. The estimation of this parameter through non-invasive techniques remains to be a challenge. Here we present a novel device to estimate the active tension exerted by the triceps surae, in humans. The basis of our method is the deformation of a load cell, firmly kept over the Achilles tendon, caused by the change in the line of action of it. This change is observed during its tightening while triceps surae contracting. In addition, a device was constructed to measure ankle torque and angle. Both, “Achilles sensor” signal and plantiflexor torque were compared while a subject performed triceps surae involuntary contractions evoked by electrical stimulation and voluntary efforts. There was a good linear correlation between these variables, in the stimulation condition. The parameters of the function allow to estimate the mechanical contribution of the triceps surae to the joint torque. The experiments performed in voluntary effort condition supported repeatability of the data and no effect associated to the collocation of the device. In conclusion, the developed system allows to study basic and clinical aspects of the triceps surae contractile capacity.

Resumen

La capacidad de un músculo para desarrollar tensión es una de las propiedades biomecánicas más importantes. La estimación de este parámetro utilizando técnicas no invasivas, continúa siendo un desafío. En este trabajo presentamos un novedoso dispositivo que permite medir la contribución mecánica del tríceps sural al torque articular, en humanos. La base de nuestro método es la deformación que sufre una celda de carga puesta sobre el tendón de Aquiles, causada por el cambio de la línea de acción de este tendón. Dicho cambio es observado cuando el tendón es estirado por el acortamiento del tríceps sural. Un segundo dispositivo fue construido para medir el torque y ángulo de tobillo. Las señales de torque y las obtenidas del “sensor de Aquiles” fueron comparadas mientras un sujeto realizaba contracciones involuntarias provocadas por estimulación eléctrica del nervio tibial y durante esfuerzos plantiflexores voluntarios. En la condición de estimulación, se observó una buena correlación lineal entre ambas variables. Los parámetros de la función obtenida permiten estimar la contribución mecánica del tríceps sural al torque articular. Los experimentos realizados con esfuerzos voluntarios muestran que los resultados obtenidos son repetibles y que no se ven afectados por las condiciones particulares de colocación del dispositivo. En conclusión, el sistema desarrollado permite estudiar aspectos clínicos y básicos de la capacidad contráctil de tríceps sural.

1. Introduction.

The main function of skeletal muscle is to contract, transmitting torques by the tendon to the joint and allowing the movement and the maintenance of postures. The capacity of a muscle to develop tension, or contractile capacity, is one of the most important biomechanical properties (Ravary *et al*., 2004; Đorđević *et al.* 2011). Quantification of muscle tension is important in the fields of motor control, biomechanics and robotics, as well as in those that require an understanding of human motor behavior, as sport and physiotherapy (Komi, 1990; Bouillard *et al*., 2011; Đorđević et al. 2011; Bey & Derwin, 2012). The estimation of muscle tension remains to be a challenge (Bouillard et al., 2011) and many human studies use the net joint torque as alternative. However, the joint torque results of the sum of the synergistic and antagonistic muscles acting on the joint.

The "gold standard" device for force estimation in animals is the buckle force transductor implantable at the tendon (Weytjens *et al*., 1992). This kind of devices consists of a metal closure with stress sensors in which the tendon is bonded. As the sensors are part of a Wheastone bridge, the change in resistance due to the buckle deformation is detected by a change of electric potential proportional to the tendon force.

In the past 30 years several techniques for directly or indirectly measuring muscle strength in humans have been developed (Fleming & Beynnon, 2004). Implantable devices, similar to the buckle force transducer, allow to measure the deformation of an elastic structure standing along the tendon (Komi *et al*., 1987, Komi *et al*., 1990; Ravary *et al*., 2004). To calibrate this type of device Komi *et al*. (1990) compared the recorded force with the muscle torque exerted by the ankle joint.

Other kinds of direct measuring devices are fiber optic transducers. The fiber optic technology was developed by Komi *et al*. (1996) for the purpose of generating a less invasive action. The transducer consists of a fiber optic probe inserted perpendicularly into the tendon. As the intensity of light traversing through the probe is modulated by the lateral compression exerted by the tension of the tendon, continuous recording of light intensity variation allows to estimate the tension of the tendon. The calibration of both types of devices is performed in a similar way relating the output of the transducer with the moment generated by the joint during maximum voluntary muscle contractions and electrical stimulation. Sensors are generally well tolerated by the subject except for buckle transducer which can produce pain dependent on the size of the device.

There are several limitations associated with the sensors described above that most often preclude their use in human research (Ravary *et al*., 2004). First, they are invasive methods requiring small surgery under local anesthesia. This implies a reduction of potential volunteers who consent to this maneuver and also research centers allowed to perform these studies. Second, a post-operative period is necessary before conducting the experiments. Third, the presence of a mechanical device may potentially cause lesions that can not only affect inevitably the results of the experiment but also surpass the clinical threshold.

Non invasive methods of force measurement based on evaluation of the stiffness of the muscle belly were more recently developed. Boulliard *et al*. (2011) used an elastography technique called supersonic shear imaging (SSI) to study two muscles of the hand. This technique consists of calculating shear elastic modulus by measuring the local shear wave velocity propagation. Their results were compared with electromyographic data. According to Bay et al. (2012) although there are preliminary reports on the use of sonoelastography in human Achilles tendon, the accuracy and reliability of this technique has not been rigorously evaluated. Another introduced non invasive procedure is the tensiomyographic measuring technique (Dahmane *et al*., 2001, Đorđević *et al*., 2011). This technique is based on the selective tensiomyographic measurement of muscle belly displacement, in which muscle belly displacement (and the pressure that it generates over the skin) is proportional to muscle force. Based on this technique, Dordevic *et al*. (2011) have developed a “muscle contraction sensor”. Such sensor is attached to the skin applying pressure on it and ultimately on the measured muscle. During skeletal muscle activity, the tension of that muscle changes, changing the recorder pressure.

The objective of this work is to introduce a novel non-invasive device based on the measurement of the deformation of a load cell firmly attached to the Achilles tendon. This technique combines the ideas of measuring the deformation of an external device with the direct measurement related to the tension suffered by the tendon.

2. Materials and methods

2.1. Foundations of the method.

The ("Achilles") sensor was designed to measure the isogonic active torque of the triceps surae, in a non invasive way. Active refers to the generated torque above its passive tension at the rest. Several studies have shown that the line of action of Achilles tendon significantly moves during an isogonic plantiflexor effort (Maganaris *et al.*, 1999) from de the rest to 30% of the maximal voluntary effort (Hashizume *et al.*, 2014) (Fig. 1). This can be easily perceived by touching the skin over the tendon while contracting the plantiflexor muscles.

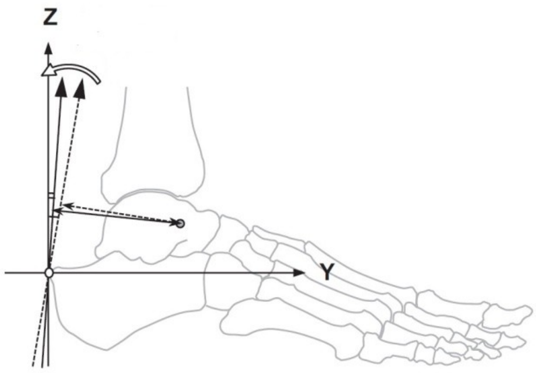

Fig 1. Orientation of the Achilles tendon action line significantly changes respect of the z axis during the contraction of the Triceps surae, from rest (dotted narrow) to 30% of the maximal voluntary contraction (solid narrow). Figure extracted from Hashizume, S., Iwanuma, S., Akagi, R., Kanehisa, H., Kawakami, Y., & Yanai, T. (2014). The contraction-induced increase in Achilles tendon moment arm: A three-dimensional study. *Journal of biomechanics*, 47(12), 3226-3231

The basis of our method is the change in the deformation of a load cell, kept in contact with the skin over the Achilles tendon, caused by the movement of the line of action during the triceps surae contraction (Fig. 2). At the resting state, the external load cell is placed in such way that it causes a small deformation of the tendon (baseline of the sensor signal). Muscle contraction increases the tightening along the tendon causing a monotonical increment of the cell deformation. The cell was used in combination with an instrumental amplifier to amplify and filter the voltage signal, an A/D conversor and a PC (with data acquisition interface) to store the data. All components will be described in more detail below.

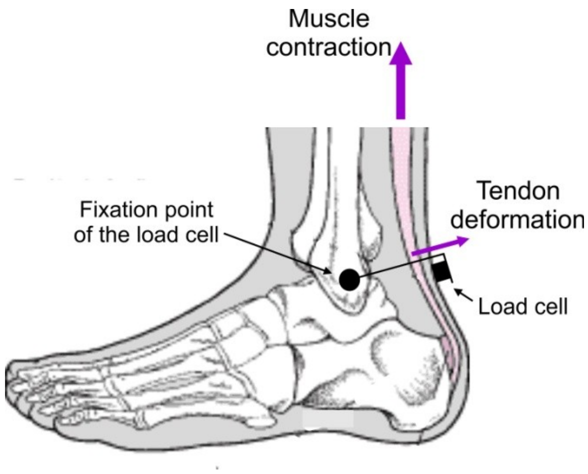

Fig 2. Schematic representation of the Achiles sensor. A load cell is attached over the Achilles tendon, using the melleoli as the fixation points. When triceps surae contracts, the tendon moves increasing a pressure on the cell. The output signal of the sensor is monotonically related with the triceps surae effort

2.2. Force transducer and hardware implementation.

The force transducer of the Achilles sensor consists of a strain gauge load cell obtained from a commercial scale (55x20mm, range 0-2Kg, resolution 0.1gr). As one extreme of the cell is in contact with the tendon, it exerts a pressure over the cell while triceps surae contracting. The force being sensed deforms the strain gauge and it measures the deformation (strain) as a change in electrical resistance (Ravary *et al*., 2004, see Fig, 3A). Strain gauges are arranged in a full bridge (Wheatstone) configuration. A Wheatstone bridge is an electronic circuit commonly used to measure an unknown resistance. We connected the bridge output with the input of a signal conditioning analogic home made circuit (Fig 3B, C) but maintaining the original connection with a scale display circuit. This allowed us to directly calibrate the deformation of the device. It was performed increasing the load in steps of 300gr and recording the sensor voltage output. The circuit was designed (see scheme in Fig. 3B) to low pass filter the signal at 40 Hz, 18 dB/octave, amplify 8 times and adjust the gain and the offset. The signal was digitally sampled at 1kHz (1322 Axon instruments).

Additionally, we constructed a joint device to measure ankle torque and angular position. We used a load cell and a linear rheostat, respectively, connected to conditioning circuits similar to that mentioned above. Mechanical supplement is described below.

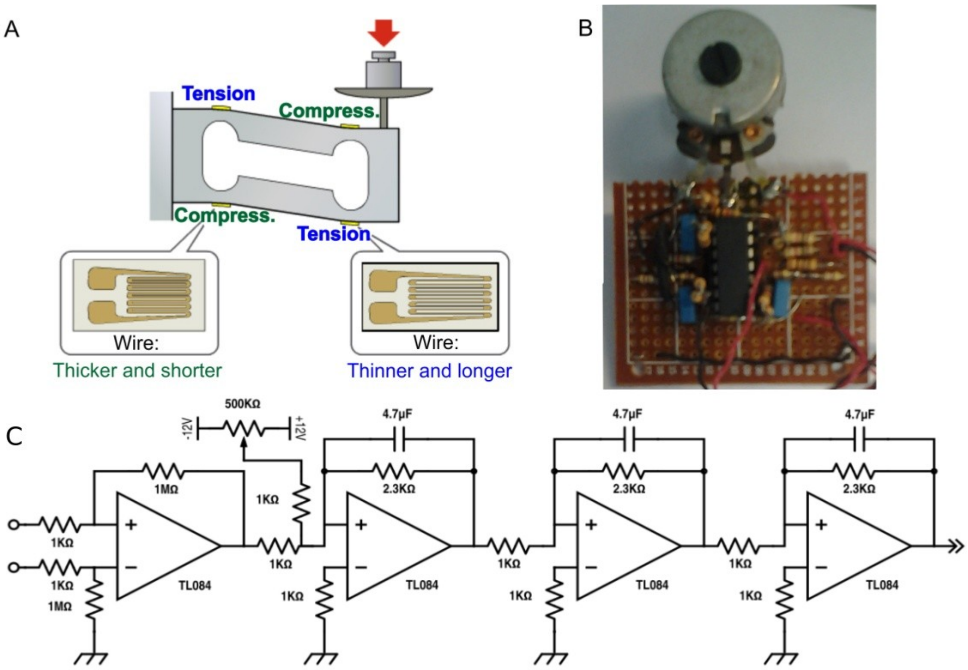

Fig 3. A) Scheme of a strain gauge load cell. The strain gauges placed at the extremes of the cell are deformed by the imposed load and its conductivity changes. The change in the voltage output is proportional to the load. B) Implementation of one of the conditioning electronic circuits of the angle, torque and Achilles tendon signals. C) Design of the conditioning circuit.

2.3. Mechanic model of the Achilles sensor and joint device

The transducer is mounted in a rigid U-shaped structure that is firmly attached to both malleoli by two adjustable arms (Fig. 4A). To secure adaptation, the end of each arm has a foam cover cup adapted to the shape and size of the malleolus. The distance between both arms is controllable with a double screw. The intermediate portion of the U-shaped structure has a mobile bar (125x45mm) in which the load cell is firmly attached. On the side of the cell that makes contact with the skin there is an aluminum buffer stop of round edges to assure a proper contact between the cell and the tendon. In order to apply the desired basal pressure on the tendon, the bar containing the cell can be moved with the aid of fine thread screws.

The joint device consists of two wooden pieces: a footplate and a vertical support, articulated by an axis rotation (Fig. 4B and scheme in Fig.5). The vertical support is in contact with the anterior side of the leg and the footplate is in contact with the sole of the foot. The rotation axis consists of a wooden mobile piece attached to the vertical support with double screw. This piece can be moved vertically in order to align it with the bimaleolar line in the horizontal plane. One of a pair of rheostats serving as hinges of the rotation axis is used to measure the ankle angle (indicated in Fig. 5 as “A”). A semicircle was centered with the axis to make the calibration of the sensor. It was performed turning the footplate in steps of ten degrees and recording the sensor voltage output. Torque measurement was obtained by another load cell attached at the end of the footplate (indicated in Fig. 5 as "T") and perpendicularly connected to a fix point (to prevent footplate moved when subjects exert forces). Torque value was calculated by multiplying the lever arm of the footplate times the normal force recorded by the sensor.

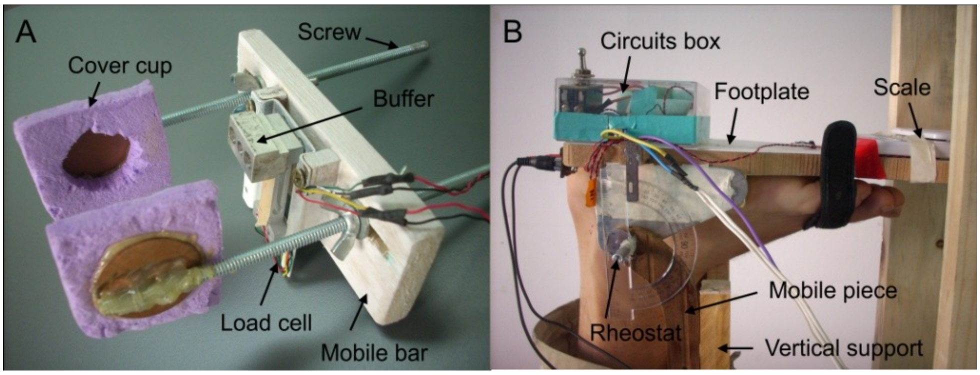

Fig 4. A) The Achilles sensor consists of a load cell attached to a bar which is crossed by two screws. At the end of each screw, a cover cup helps to attach the device to the ankle by the malleoli. The distance between the malleoli and the load cell is adjusted by the double nut. B) The joint device consists of a footplate and a vertical support articulated by a rotation axis. The rotation axis contains the rheostat that serves as ankle angle sensor. It is aligned to the inter-malleoli line through a mobile piece. Over the footplate a scale is placed with its load cell uncovered. A rigid rod (not showed in the photo) is connected to the cell (through a hole in the footplate) and a fix point

2.3. Tests and Data analysis

We tested the recording system under conditions of voluntary activations and evoked contractions by electrical stimulation, in one healthy human subject. The experiment was approved by the ethical committee of the *Instituto de Investigaciones Biologicas Clemente Estable*, conforming the standards set out in the Declaration of Helsinki.

Data were obtained with subjects in prone decubitus with the tibia vertically aligned (right knee angle of 90°) as is showed in Fig. 5. The prone position was chosen to minimize trunk and hip rotation and limb displacements. The right leg was firmly attached to a vertical support fixed to the table. The foot sole was in contact with a footplate and "Velcro"® straps were used to immobilize it. The rotation axis of the footplate-support and the inter-malleoli line were aligned.

Torque, angular position and Achilles sensor signal were recorder simultaneously in two different conditions: 1) A subject was trained to control the exerted torque with a visual feedback of the torque signal displayed in an oscilloscope with a high sweeping speed. A target torque reference line was presented in another channel of the oscilloscope. The subject was instructed to steadily match the target torque value generating sustained plantiflexor effort levels from low to moderate, for a two second period. 2) Transcutaneous electrical stimulation of the tibial nerve was performed in a subject, using bipolar configuration electrodes, with the cathode over popliteal space and the anode on the patella. The stimuli were square-wave pulses of 500 μs duration delivered via a stimulus isolation unit. The amplitude of the stimuli was from the minimum to generate a single twitch to the level in which the subject started to feel discomfort.

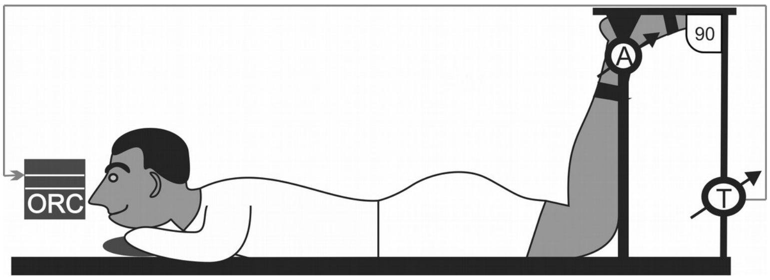

Fig 5. The subjects were lying in prone position with the forearms under the chin and the knee angle at 90°. The leg was firmly attached to a vertical pole. The ankle angle (A) and normal torque to the pedal (T) were measured. The subject was asked to match a target signal controlled by the experiment. A constant target and the generated torques signal were displayed together on the screen of an oscilloscope at high sweeping speed (ORC) and showed to the subject

We determined the relationship between the Achilles sensor signal and plantiflexor torque in an ankle angle of 0° (perpendicular position of the footplate respect to the tibia). From the electrical stimulation condition, a correlation analysis was performed during one single stimulus to evaluate the capacity of the sensor to detect temporal changes in muscle tension. In addition, a response curve was constructed by plotting the peak value of both signals obtained from each stimulus. This curve was fitted by the best-fitting model.

From the voluntary efforts condition, we tested data repeatability by performing 3 successive trials and repeating the experiment in 2 different days, with the same subject. The graph obtained from the whole set of data was fitted by a non-linear function and residual analysis was performed.

3. Results and Discussion

Figure 6A shows superimposed torque (black line) and Achilles sensor (red line) recordings during the application of a single stimulus. The correlation analysis between both signals showed a good linearity with a Determination coefficient of 0.99 ($p<0.0001$, significance level of 0.05). However, the correlation graph (Fig. 6B) evidences a slight differential behavior during muscle activation and de-activation, like an hysteresis process. It can be clearly viewed in the de-activation phase of both superimposed curves (Fig. 6A) in which the torque signal declines more smoothly. This difference could be explained by a difference in the filtering circuit responses of these signals.

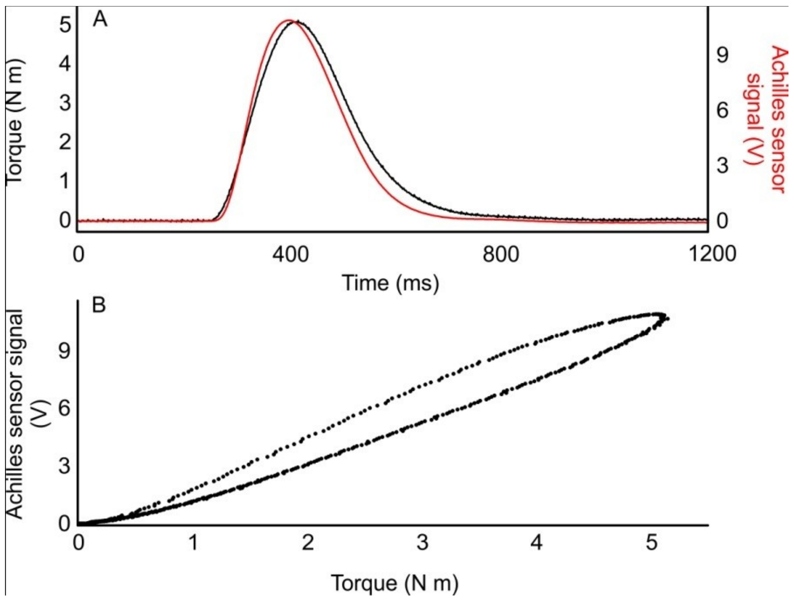

Fig 6. A) Plantiflexor torque (a twitch, black line) was recorded while the subject involuntarily pressed the footplate by a single electric stimulus. Simultaneously, the Achilles sensor signal (red line) was recorded. B) A correlation analysis was performed with the data showed above. There is a good association between variables (Determination coefficient of 0.99) indicating that our device is able to detect temporal changes in muscle tension. The graph evidenced a differential behavior during muscle activation and de-activation.

In order to estimate the mechanical contribution of the triceps surae to the net joint torque, we analyzed the function between the torque and Achilles sensor signals during plantiflexor contractions evoked by electrical stimulation of the tibial nerve. Although the tibial nerve also innervates an accessory plantiflexor muscle (plantaris) at the level of the popliteal fossa, we can consider it as negligible. Then, the triceps surae is the responsible of the joint torque in these conditions. This allows us to calibrate the system. Figure 7A shows the peak values of the Achilles sensor signal as a function of the peak values of the torque, obtained from each stimulus. Data showed a good linear relationship in the studied range, with a Determination coefficient of 0.96 ($p<0.0001$, significance level of 0.05). The studied range was limited by the tolerance of the subject to the electric stimuli. In these conditions, the parameters of the linear function can be used to predict the triceps surae torque from de Achilles sensor signal, independently of potential changes in the other involved muscles, as for example the antagonist contribution.

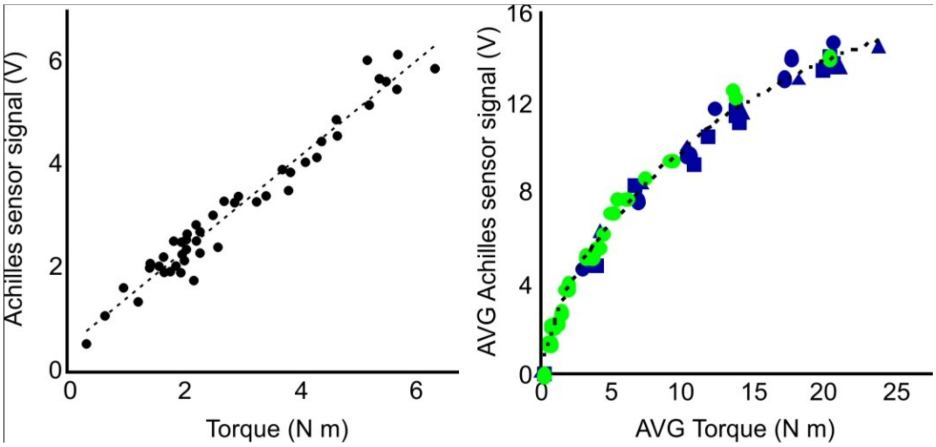

Fig 7. A) Torque and Achilles sensor signals were recorded while superficial tibial nerve electrical stimulation was applied in a subject. The peak torque was obtained from each stimulus. The relationship was fitted by a linear function (dotted line, R^2=0.96). The parameters of this function allow to predict the triceps surae torque from the Achilles sensor signal. B) Torque and Achilles sensor signal were recorded while the subject was asked to maintain a series of sustained plantiflexor efforts. Two second period of each level of effort was averaged. The graph shows the relationship between the torque and Achilles sensor signal average, for the same subject in three consecutive trials (blue symbols, each symbol corresponds to one trial) and in a different day (green circles). Data were well fitted to an exponential function (regression coefficient of 0.98)

To evaluate repeatability of the data, voluntary sustained efforts from low to moderate intensities were performed. Torque and Achilles sensor signals were recorded during three consecutive trials and (blue symbols in Fig. 7B, different shapes indicate different trials) and in a different day (green symbols in Fig. 7B), with the same subject. Considering all data together, the relationship between torque and tendon signal was fitted to an exponential model by a non-linear regression method. Residual analysis indicated that data were well fitted with a Determination coefficient of 0.98 (Chi square test to evaluate significant differences between observed values and theoretical values from the model, $p=0.3$), indicating that the whole set of data belong to the same population. There was no effect by the collocation of the device in different moments. The shape of the curve is consistent with the fact that while muscle contraction increases, the ability of deformation diminishes because the tendon is tightening.

3.3. Conclusions

A novel device to measure the mechanical contribution of the triceps surae to the plantiflexor torque was designed and constructed. This system allows to quantify potential changes in the triceps surae contractile capacity in healthy and pathological subjects, by applying the same electrical stimulus, before and after the exposition to a certain condition and comparing the developed tension. It could be used, for example, to evaluate the acute effect of pharmacological treatment, sportive or therapeutic muscle training session, fatiguing protocol, etc.. In addition, clinical or basic aspects of the contractile capacity can be studied through the electromyogram/force relationship during voluntary isogonic efforts.

The more important features of the Achilles sensor are its non-invasive character, low cost and easy application. As disadvantages, it cannot be used in the wide range of the subject force because of the changes in the deformation of the Achilles tendon finished when it is completely tightened. Moreover, the ability of deformation is probably dependent on the ankle angle and it is possible that this device cannot be used in the wide range of angles. Functional movements cannot be studied using our device.

An improvement of the electronic design could lead to a better performance of the device. Also, it is necessary to increase the number of the population sample to reinforce the results of this study.

References

Bey, M. J., & Derwin, K. A. (2012). Measurement of in vivo tendon function. *Journal of Shoulder and Elbow Surgery*, 21(2), 149-157.

Bouillard K, Nordez A, Hug F (2011) Estimation of Individual Muscle Force Using Elastography. PLoS ONE 6(12): e29261. doi:10.1371/journal.pone.0029261

Dahmane, R., Valenčič, V., Knez, N., & Eržen, I. (2001). Evaluation of the ability to make non-invasive estimation of muscle contractile properties on the basis of the muscle belly response. *Medical and biological engineering and computing*, *39*(1), 51-55.

Đorđević, S., Stančin, S., Meglič, A., Milutinović, V., & Tomažič, S. (2011). MC Sensor-A Novel Method for Measurement of Muscle Tension. *Sensors*, *11*(10), 9411-9425

Fleming, B. C., & Beynnon, B. D. (2004). In vivo measurement of ligament/tendon strains and forces: a review. *Annals of biomedical engineering*, 32(3), 318-328.

Hashizume, S., Iwanuma, S., Akagi, R., Kanehisa, H., Kawakami, Y., & Yanai, T. (2014). The contraction-induced increase in Achilles tendon moment arm: A three-dimensional study. *Journal of biomechanics*, 47(12), 3226-3231.

Komi, P. V., Salonen, M., Järvinen, M., & Kokko, O. (1987). In vivo registration of Achilles tendon forces in man. *International journal of sports medicine*, *8*(S 1), S3-S8.

Komi, P. V. (1990). Relevance of “in vivo” force measurements to human biomechanics. *Journal of Biomechanics*, *23*, 23-34.

Komi, P. V., A. Belli, V. Huttunen, R. Bonnefoy, A. Geyssant, and J. R. Lacour. (1996) Optic fiber as a transducer of tendomuscular forces. *Eur. J. Appl. Physiol.* 72:278–280.

Maganaris, C. N., Baltzopoulos, V., & Sargeant, A. J. (1999). Changes in the tibialis anterior tendon moment arm from rest to maximum isometric dorsiflexion: in vivo observations in man. *Clinical biomechanics*, 14(9), 661-666.

Ravary, B., Pourcelot, P., Bortolussi, C., Konieczka, S., & Crevier-Denoix, N. (2004). Strain and force transducers used in human and veterinary tendon and ligament biomechanical studies. *Clinical biomechanics*, *19*(5), 433-447.

Weytjens, J. L. F., Viberg, D. A., Caputi, A. A., Kallesøe, K., & Hoffer, J. A. (1992). New transducers for measuring muscle length in unrestrained animals. *Journal of neuroscience methods*, 45(3), 217-225.