ULTRASONIC MEASUREMENT OF THE TENDON STRESS DURING LOADING: PRELIMINARY RESULTS

<u>Marielle Defontaine</u> [+], Philippe Pourcelot[#], Mickaël Lemâtre [+], Bérangère Ravary[#], Catherine Roux[+], Nathalie Crevier-Denoix [#]
[+]GIP Ultrasons/LUSSI FRE 2448, 37044 Tours, France
[#]Laboratoire de Biomécanique et Pathologie Locomotrice du cheval, UMR INRA-ENVA, Ecole Nationale Vétérinaire d'Alfort, 94704 Maisons Alfort, France
defontai@med.univ-tours.fr

Abstract

Our purpose is to present a non-invasive ultrasound method for measuring the level of stress in the tendon during effort. This work is the result of a collaboration between the Laboratoire de Biomécanique et Pathologie Locomotrice du Cheval, and GIP Ultrasons. This method consists in transmitting an ultrasonic pulse along the tendon and to receive at 5 linearly spaced transducers the corresponding echo signals. First results were obtained on an equine ex-vivo forelimb, in the metacarpal area (digital flexor tendon), which was submitted to loading/unloading test (0-6000 N). The first echo arrival in the RF signal was treated to estimate two parameters: attenuation and transient time (or celerity). The transient time estimation is based on a cross-correlation algorithm between adjacent signals from the receivers or also from different loading levels. The results obtained presented important changes with loading, which makes the method sensitive to stress in the tendon.

Introduction

Tendon pathology diagnosis and/or elasticity recovering follow-up after partial or total injuries cover a wide range of invasive or non-invasive investigation methods. Most of the non-invasive ones (MRI, Echography) are not quantitative, and often bed rest will be the base treatment. The measurement of the essential parameter that is the level of stress inside the tendon during effort is not available at the moment. In the domain of ultrasonic tendon characterization, only in-vivo (in a static situation) or in vitro measurements have been performed. Celerity, attenuation or backscattering coefficient (in the range of 3-10 MHz) parameters have been measured [1]. Coefficient of elasticity (in vitro only and in static situation) have also been calculated [2] [3]. Classical B-scan ultrasound images are used to measure the level of strain simultaneously with the recording of the isometric stress applied at the tendon [4].

In 2001, the Laboratoire de Biomécanique et Pathologie du Cheval (LBPLC) at Ecole Vétérinaire de Maisons-Alfort who was studying the possibilty to estimate by ultrasound the level of stress in the equine tendon. We have developed, in a very close collaboration with the LBPLC, a non-invasive ultrasonic method. The region of interest was the digital flexor tendon of an equine forelimb (fig. 1), and particularly the Superficial Digital Flexor tendon (SDF), in red on the picture below.

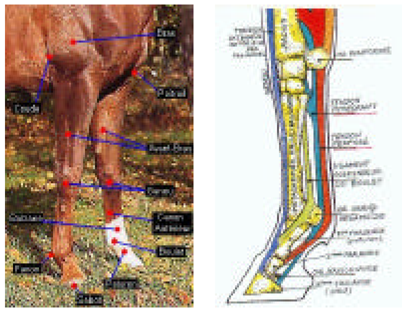


Figure 1 : Equine forelimb, Superficial Digital Flexor (SDF, in red) and Deep Digital Flexor (DDF, in blue) Tendons.

In this paper we first describe the feasibility technique tested and validated in-vitro on an ex-vivo equine forelimb placed into a traction machine. Then we developed with Ultrasons Technologies and Vermon societies a dedicated instrumentation for in-vivo measurements, which offers the possibility to estimate the celerity in the tendon during effort.

Methods

Experimental setup

Considering the anatomy of the region to be investigated, we have proposed to use an axial transmission technique (fig.2). In this configuration, an ultrasonic pulse emitted at 1MHz central frequency is recorded 40 mm further in the axial direction of the tendon. An angle of incidence, corresponding to the critical angle (evaluated at 35°-45°), was applied in order to generate a longitudinal wave propagating inside the tendon. The RF signal acquired from the receiver transducer was treated to extract the first echo, which was related to the faster tendon propagation medium. The emitter/receiver distance was evaluated with respect to the order of time arrival of the different echoes. Particularly we wanted the skin echo to arrive later.

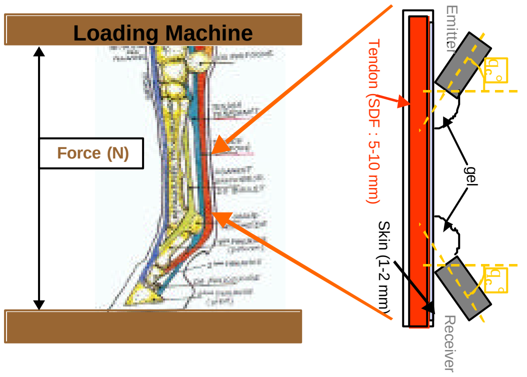


Figure 2: Experimental set-up.

The forelimb ex-vivo member was placed in a testing machine that allows loading values from 0 to 6000 Newton. For each loading level (by steps of 500 N) an ultrasonic acquisition was realized.

Algorithms

The signal analysis was rather simple and divided into following steps:

- *Signal Windowing in the region of interest of the RF signal:*
  - Detection of the first echo arrived;
  - Truncation of this echo with a window duration, roughly equal to impulse response duration (5µsec, in red on fig. 3);
  - Use of a Hamming filtering to smooth the edges of the truncation at the end of the window.
- *Time of flight changes estimation:*
  - Crossed-correlation of adjacent signals during compression and extension cycle;
  - Parabolic fit and FFT fit to estimate the maximum value of the crossed-correlated function;
  - Cumulated summation of adjacent estimated delays in order to provide the estimated time-of-flight as a function of the loading.
- *Attenuation changes estimation:*
  - Estimation of the slope of the transfer functions ratio in order to provide the attenuation coefficient normalised by the maximum loading value:

$$attenuation = slope_{[400-800kHz]}\left(20Ln\left(\frac{|S_{load\_i}(f)|}{|S_{load\_max}(f)|}\right)\right)$$

  - Estimation of the maximum of the FFT module which is of course correlated to the previous one.

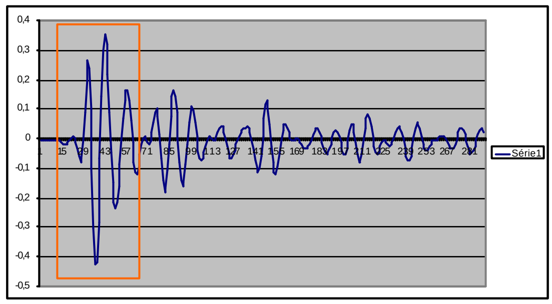
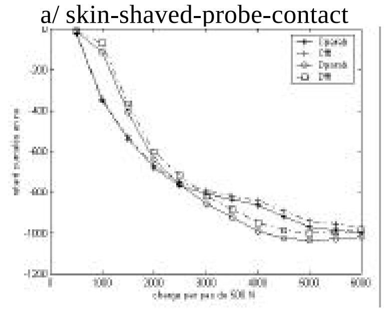


Figure 3 : Experimental RF axial transmission signal.

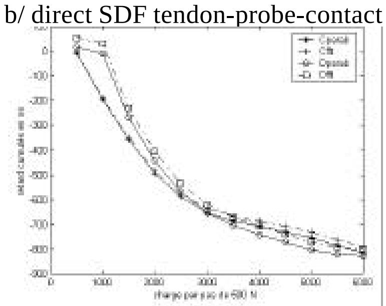
Feasibility results

In order to validate this technique and to make sure that the parameters were really related to the ultrasonic propagation inside the tendon, we performed several experiments (figure 4).

a/ skin-shaved-probe-contact



b/ direct SDF tendon-probe-contact



c/ direct SDF tendon-probe-contact + paper of sheet placed between SDF and deeper DDF tendon.

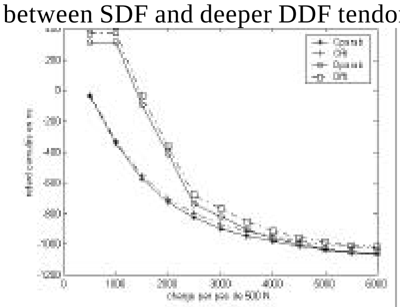


Figure 4 : Three preliminary experiments results: time delays changes curves as a function of loading (0-6000N) and unloading (6000-0N) cycle of an ex-vivo equine forelimb.

The results obtained clearly demonstrate that the changes we observe in the estimation of the first time-echo arrival were actually related to the SDF tendon. It is not related to the skin, neither to the deeper DDF tendon. So this parameter is a good candidate for the estimation of the level of stress in such tendons.

The energy curves (fig. 5) presents also changes with the loading: the transfer function modules are increasing with the loading values, i.e. the level of stress in the tendon. In addition, the transfer function curves (fig. 5 left) clearly show the viscosity absorption effect, as the curves are shifted to low frequency values when loading is decreasing.

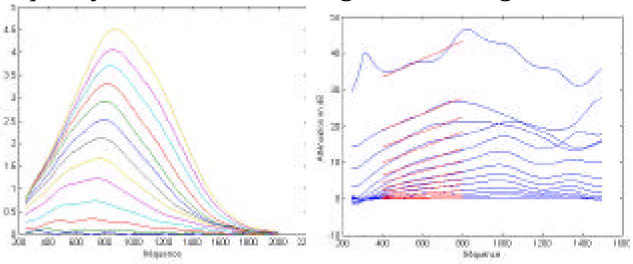


Figure 5: Transfer function modules and attenuation curves for regularly spaced loading values (0-6000N, steps of 500N).

Both parameters the attenuation coefficient and the maximum of the spectrum amplitude were decreasing and increasing respectively with increasing loading values (increasing level of stress), such as expected (fig. 6). With increasing loading values, the tendon structure becomes more and more stiff and less attenuating medium. This observation is also available for the first time-echo arrival, which is decreasing with increasing loading values, the medium becoming also faster and faster.

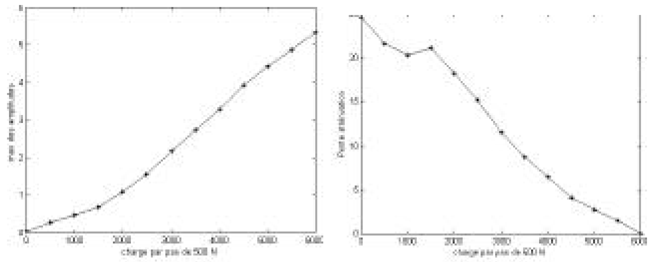


Figure 6 : Maximum of the transfer function modulus and attenuation coefficient as a function of loading.

Because of the widening of pulse response with increasing loading values (transfer function shifted to lower frequencies), the cross-correlation method for delay estimation, is not adapted to separate absorption from pure elasticity effects. That is why for the dedicated in-vivo instrumentation we have designed a probe with 5 receiver elements in order to estimate the celerity between receivers 1 and 5, and so being less dependent of visco-elasticity effects.

Dedicated in-vivo ultrasound instrumentation

The practical interest of this new instrumentation is related to the in-vivo facility of use. It is a stand-alone equipment functioning with batteries, composed of a small and light electronic module. The Ultrasonic data are recorded in situ and later transferred to a PC for analysis via an Ethernet link. The scientific interest resides in the possibility to estimate the celerity in an assumed homogeneous section of the tendon (18 mm long in the case of the equine application).

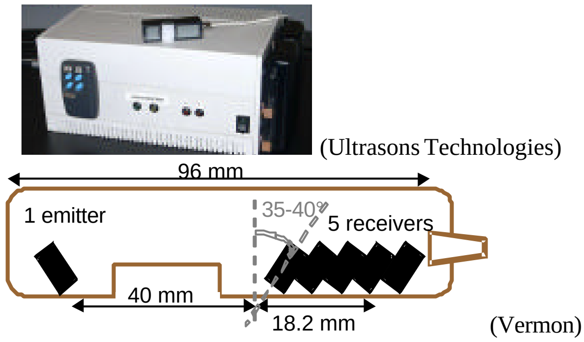


Figure 7: In-vivo Ultrasonic instrumentation.

Celerity results & discussion

The data analysis was the same as the previous one, we have only added the estimation of celerity between R1-R5. For this parameter we have been calculated the time delay difference between adjacent receivers using the same cross-correlation method. Figures 8 and 9 are presenting the (x,t) diagrams for different loading values on an ex-vivo equine forelimb. Two experiments were performed: direct SDF tendon-probe contact and shaved-skin-probe contact, respectively in figures 8 and 9.

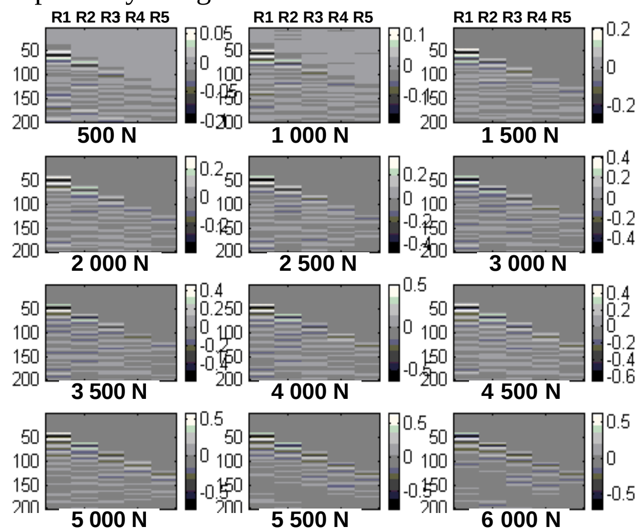


Figure 8: (x,t) diagrams for different loading values (probe directly in contact with the SDF tendon).

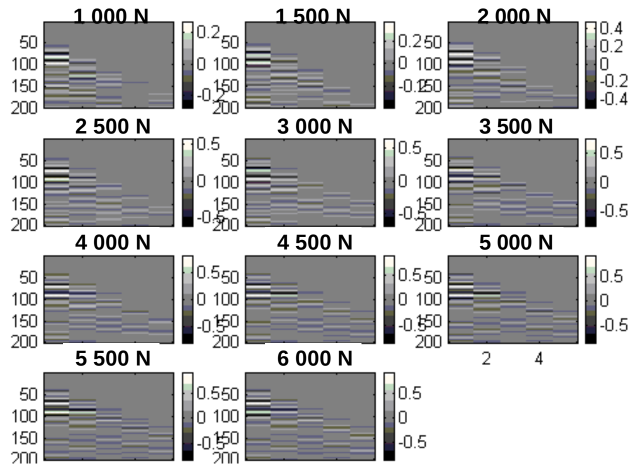


Figure 9: (x,t) diagrams for different loading values (probe in contact with the shaved skin).

R1-R5 celerity have been estimated in both cases, and plotted as a function of the loading/unloading values (graphs 10 and 11).

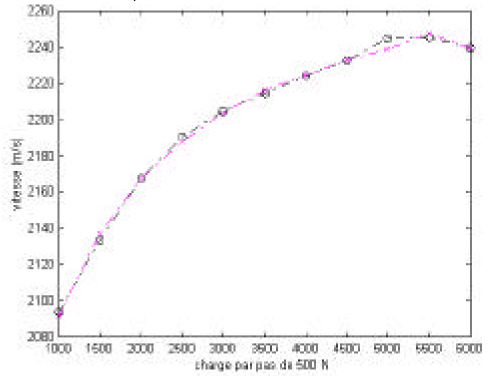


Figure 10: Celerity measured between R1 and R5 on an ex-vivo forelimb placed in the testing machine. The transducer was placed directly in contact with the SDF tendon.

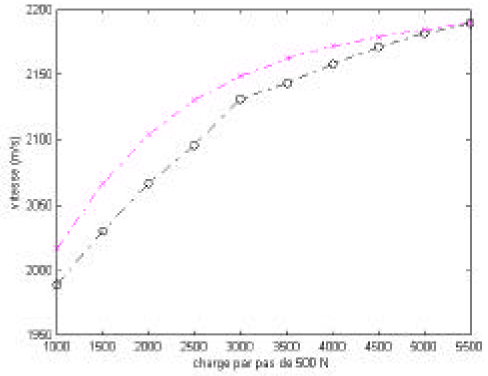


Figure 11: Celerity measured between R1 and R5 on an ex-vivo forelimb placed in the testing machine. The transducer was placed in contact with shaved skin in regard to the SDF tendon.

In both experiments important changes in celerity (Δc_{R1-R5}) have been obtained, more than = 130 m/s and 200 m/s in the first and second case respectively. A small hysteresis value is appearing in the shaved-skin-probe configuration. The use of a differential method to estimate the transient time makes the method less sensitive to visco-elasticity effects due to different loading values. So this parameter should be more relevant to estimate true tendon stiffness changes.

Perspectives and conclusion

Equine application

This technique has been developed in the context of the equine application (fig. 12), which was the problematic of the LBPLC. They have performed several in-vivo protocols in order first to evaluate the precision of the technique and then to evaluate different clinical studies. The results are to be submitted in a review and will not be provided in this paper. During this work a patent has been recorded on this method by INRA-ENVA.

Figure 12: Equine in-vivo experimental set-up.

Human application

The problematic of LUSSI concerns the human application, and we have adapted this instrumentation for the in-vivo estimation of the level of stress in the Achilles tendon (fig. 13). A Ph.D. student is continuing this work. She will particularly study the acoustical propagation model in the tendon, in order to approach an estimation of elasticity coefficients.

Figure 13: Human in-vivo experimental set-up.

Acknoledgements

The research team of GIP Ultrasons/LUSSI and the LBPLC-UMR INRA/ENVA express their gratitude to the anonymous human and equine subjects who participated in the in vivo experiments, to Professor Fouquet from Tours Hospital for his device, to the electronic study team of ULTRASONS TECHNOLOGIES, to the transducer study team of VERMON, to Claude Dardillat from LBPLC UMR INRA and to Danny Carré from GIP Ultrasons/LUSSI, for their participation in this program.

References

[1] C.A. Miles, “Ultrasonic properties of tendon: Velocity, attenuation, and backscattering in equine digital flexor tendons,” J. Acoust. Soc. Am, vol. 99 (5), pp. 3225-32, 1996.

[2] B.K. Hoffmeister, S.M. Handley, E.D. Verdonk, S.A. Wickline, and J.G. Miller, “Estimation of the elastic stiffness coefficient c13 of fixed tendon and fixed myocardium,” J. Acoust. Soc. Am, vol. 97 (5), pp. 3171-76, 1995.

[3] P.L. Kuo, P.C. Li, and M.L. Li, “Elastic properties of tendon measured by two different approaches,” Ultrasound in Med. & Biol., vol. 27 (9), pp. 1275-84, 2001.

[4] K. Kubo, Y. Kawakami, H. Kanahisa, and T. Fukunaga, “Measurement of viscoelastic properties of tendon structures in vivo,” Scand. J. Med. Sci. Sports, vol. 12, pp. 3-8, 2002.