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NeuroMusculoSkeletal Lab

Quantifying Passive Stiffness in Human

From the Lab to the Track:

Application to the runner's lower leg

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"If I have seen farther than others, it is because I was standing on the shoulders of giants."

Sir Isaac Newton

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List of abbreviations

ANOVA	Analysis of Variance
AT	Achilles Tendon
BW	Body Weight
C	Creep (myotonSAS definition)
CC	Contractile Component
CEP	Elastic Parallel Component
CES	Elastic Serial Component
COM	Center of Mass
D	Decrement (myotonSAS definition)
DE	Distal Extensor
DF	Distal Flexor
EL	Elastic Stiffness
EMG	ElectroMyoGraphy
EOD	Electromechanical Oscillation Device
ES	Effect Size
F	Frequency (myotonSAS definition)
FFT	Fast Fourier Transformation
I	Inertia
Kjoint	Joint stiffness
Kleg	Leg stiffness
Kvert	Vertical stiffness
LG	Lateralis Gastrocnemius
LRA	Locoregional anesthesia
MAS	Maximal Aerobic Speed
MDC	Minimal Detectable Change
MG	Medialis Gastrocnemius
MTJ	Muscle Tendon Junction

NMC	Net Metabolic Cost (of running)
PE	Proximal Extensor
PF	Proximal Flexor
PF	Plantar Fascia
PMAS	Passive MusculoArticular Stiffness
PT	Patellar Tendon
R	Relaxation time (myotonSAS definition)
ROM	Range of Motion
S	Stiffness (myotonSAS definition)
SD	Standard Deviation
SEM	Standard Error of measurement
SMD	Spring Mass Damper (model)
SSC	Stretch Shortening Cycle
SWE	Shear Wave Elastography
VI	Viscous Stiffness
VLR	Vertical Loading Rate
VO2max	Maximal oxygen uptake

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Figure 1. Illustration of a spring at rest (upper left) undergoing an external traction force (bottom left) **F** in Newton, inducing a change in length (or deformation) **Δx** in [m]. This relationship between the force and the deformation is represented in the graph (upper right) with the slope of the linear relationship indicating the stiffness (k) in [N/m].

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Figure 33. Evolution of the lower limb stiffness measurement point throughout the 10k run. White dot represents significant difference value compared to baseline, i.e., plantar fascia measurement at the 5th and the 10th kilometer.

Figure 34. The observation of the elastic stiffness of the Achilles tendon in a healthy 20-year-old male can be accurately assessed through ultrasonography. This method allows for the visualization of the Muscle-Tendon Junction (MTJ) of the medial gastrocnemius. During the

assessment, the subject's ankle is maintained in a neutral position. When the subject performs an isometric contraction, a force F in Newtons is generated, which can be recorded by a force plate positioned under the subject's foot. Concurrently, the displacement Δx, measured in millimeters, can be measured using ultrasonography. According to Hooke's Law, as mentioned in the mechanical engineering section, the elastic stiffness of the Achilles tendon (k) can be calculated using the formula: $k = \frac{F}{\Delta x}$ where k, expressed in Newtons per millimeter (N/mm), provides a quantifiable measure of the tendon's ability to resist deformation under load.

Figure 35. Prototype of the measurement device for recording ankle amplitude and torque. An electronic goniometer can record the position of the ankle. This position can be either locked or left free to observe PMAS in static or dynamic conditions. A torquemeter can be placed on the axis of the device to record the pressure force applied by the resistive movement during dynamic assessments, or by the ankle plantar flexors, predominantly the triceps surae, in static conditions.

Figure 36. Representation of biomechanical stiffness in running.

Figure 37. Representation of stiffness interest in running performance. Green arrow represents positive correlation with running performance. Yellow arrow represents negative correlation with running performance. Grey cross represent no correlation with running performance.

Summary

The assessments of passive musculoarticular stiffness (PMAS) in the lower limbs during running represents a complex challenge, owing to the difficulties associated in accurately quantifying and providing meaningful evaluation of these outcomes. The objectives of this thesis were centered on three axes: 1. to enhance our understanding of PMAS behavior in healthy individuals, 2. to elucidate a specific technique for measuring PMAS, specifically myotonometry, aiming to better accommodate the limitations encountered in field setting with runners. Finally, 3. we undertook a comprehensive analysis of PMAS in runners, which included a running economy testing trial, an incremental running trial, and a constant velocity run.

In this thesis, we have provided an extensive discussion on the viscoelastic stiffness in the human body. This investigation into viscoelastic properties offers normative data obtained from a diverse sample that includes individuals ranging from young children to elderly. This information could assist in evaluating the optimal range of PMAS in healthy individuals and in making comparisons across various athletic or pathological conditions. Additionally, this section has enhanced our understanding of PMAS and the challenge of evaluating viscoelastic stiffness in clinical settings. More specifically, we have confirmed the possibility to quantify PMAS without any voluntary contraction in patients.

Furthermore, the research provides a comprehensive explanation of the principles underlying myotonometry and the use of the MyotonPro device. Although the equipment and technology are currently utilized in research, there are limited studies that conduct fundamental assessments to determine the potential measurements and their practical utility. The MyotonPro device was validated for its stiffness outcome in comparison to a gold standard rheometer. However, other parameters of the MyotonPro device have not been reported as either reliable or sensitive enough for the elastic stiffness component. Moreover, there was no concrete correlation found between gold standard electromechanical oscillation methodology and myotonometry concerning viscous stiffness. In this thesis, we also provided recommendations for using myotonometry to measure stiffness in the lower leg.

In the final part, this thesis reported a clinical observation regarding the use of myotonometry measurement in running. According to the spring mass damper model, studying the stiffness and rheology of lower limb tissues could offer valuable insights into enhancing performance, as well as a better understanding and managing of running-related injuries. While they are initial observations, they open the possibility and reflection to further develop methodologies to support healthcare professionals and practitioners.

1. General Introduction

1.1. Terms and definitions

In the literature and across various fields of practice, such as engineering, biomechanics, and clinical profession including physicians, physiotherapists, and podiatrists, there exists a range of terminologies or synonyms for different terms. The concept of stiffness is no exception. Therefore, this introduction highlights the conceptual terms and definitions used across mechanical engineering, materials engineering, biomedical engineering and biomechanics.

1.1.1. Mechanical engineering

In mechanical engineering, the concept of stiffness is commonly illustrated by a spring (Figure 1). When a spring undergoes an external force ($F_{traction}$), measured in Newton [N], such as traction, it undergoes a proportional change in length (Δx), measured in [m], that we will name in the present manuscript: a deformation. The stiffness is defined as the slope of the linear relationship between the force applied to a structure and the associated deformation. This relationship can be represented by the following equation:

$$F = k \cdot \Delta x \quad \text{(Equation 01)}$$

This constitutive law is commonly used in the field of mechanical engineering and is recognized within this discipline of physics.

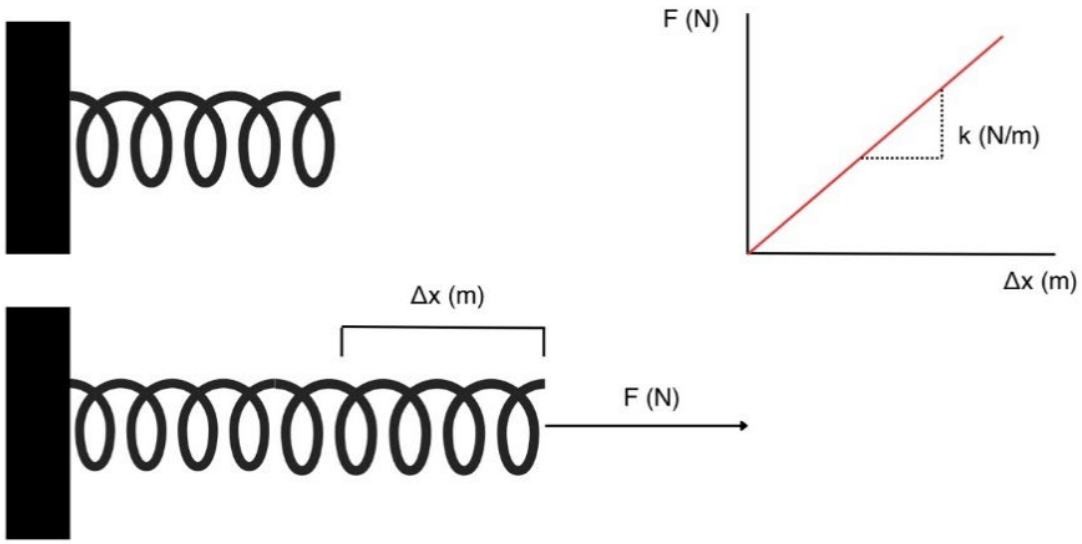


Figure 1. *Illustration of a spring at rest (upper left) undergoing an external traction force (bottom left)* ***F*** *in Newton, inducing a change in length (or deformation)* ***Δx*** *in [m]. This relationship between the force and the deformation is represented in the graph (upper right) with the slope of the linear relationship indicating the stiffness (k) in [N/m].*

In this field, we can also describe the damping effect, commonly known as viscosity. Similarly, to stiffness, viscosity can be defined as the ratio (or the slope) between an external force ($F_{damping}$) expressed in *[N]* and the speed of deformation, i.e., the time-rate of the damper length, L, $\dot{\mathbf{x}} = \frac{\boldsymbol{dL}}{\boldsymbol{dt}}$) expressed in [m/s] (Figure 2). Viscosity is quantified by the damping coefficient (D) expressed in [Ns/m], using the following equation:

$$F_{damping} = D \,.\, \dot{x} \quad \text{(Equation 2)}$$

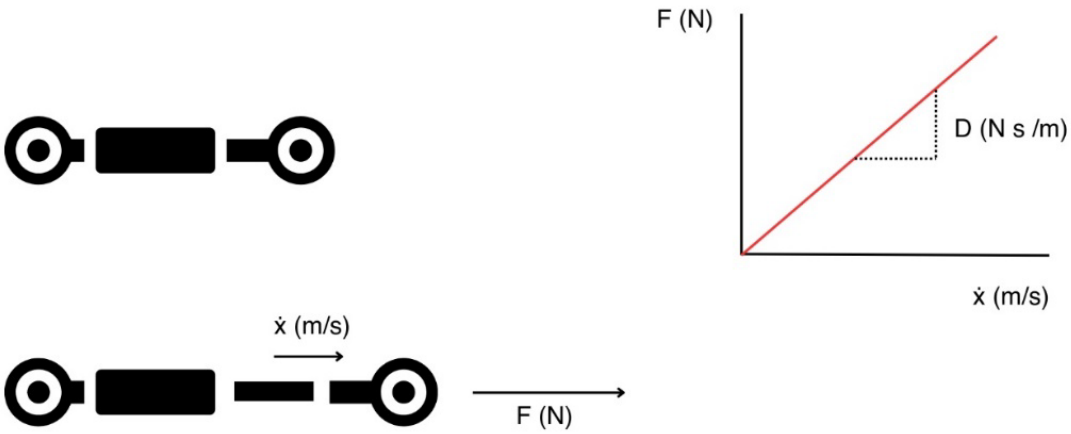


***Figure 2**. Viscosity can be illustrated by a damper at rest (upper) and a damper in movement (bottom). This linear relationship can be depicted in the graph in the upper right. The Force (in [N]) is related to the damping coefficient (D) expressed in [Ns/m]. This coefficient depends on the material damping properties or on fluid viscosity in case of a hydraulic damper for instance. D is not dependent on the system position, in contrast to stiffness.*

We can also describe friction forces, which are represented by the following equation:

$$F_{friction} = -f \,.\, F_n \,.\, sign\, \dot{x} \quad \text{(Equation 3)}$$

where, $F_{friction}$ is the dry friction force expressed in [N], f represents the dynamic friction coefficient between the two bodies sliding over each other, F_n is the normal force (in [N]), and the sign of ẋ indicates the direction of the relative velocity of the body on which the force is applied; indeed, force is opposed to speed (Figure 3).

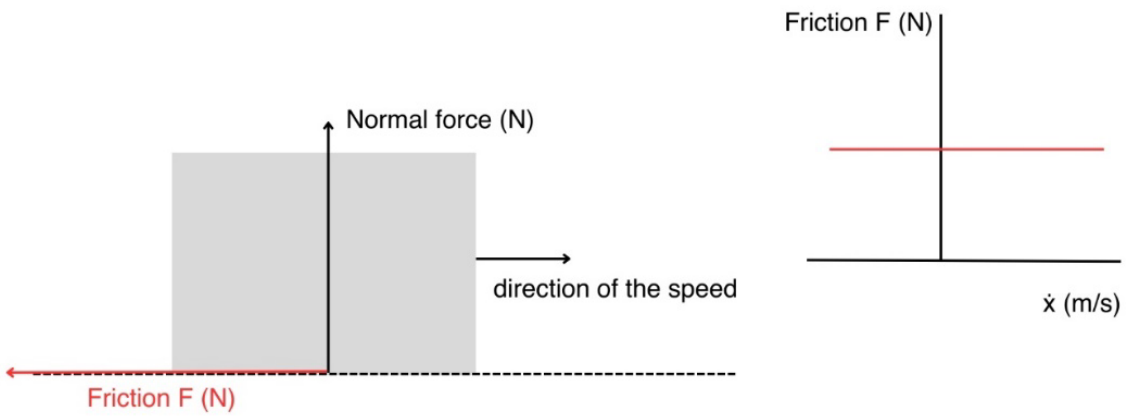


Figure 3. *Representation of a bloc that is moving to the right. The Friction force (N) is opposed to the direction of the speed of the box. The normal force is indicated in (N). the graphic representation of the situation is represented on the upper right.*

1.1.2. Materials engineering

The stiffness of a specific material is governed by Hooke's law and depends on the elastic modulus also known as Young's Modulus. The elastic modulus of a material defines the extend of the elastic properties. A material's elastic modulus measures its ability to withstand deformation, in a so-called elastic manner. When materials are subjected to an elastic stress (σ) (in [N/mm^2], a strain (ε) (or relative deformation = Δl/L$_0$) in [%] arises, but the material returns to its original state once the stress is released. Once materials transition from the elastic region to the plastic region reach the yield point, they experience permanent deformation that remains even after the tensile stress is removed. The Young modulus is defined as the slope of the elastic region, that is the straight-line portion of the stress (σ) and strain (ε) curves (Figure 4, green zone) and is calculated according to the Hooke law:

$$\sigma = E \,.\, \varepsilon \quad \text{(Equation 4)}$$

where σ (the stress) is expressed in [N/mm^2] and is the ratio of the external force to the cross-sectional area of the structure, in [mm^2]. **E** represents the elastic modulus, also expressed in [N/mm^2]. ε (the deformation) is dimensionless and represents the relative change in length, calculated as the ratio of the difference in length (ΔL) to the original length (L$_0$) of the system, expressed as a percentage.

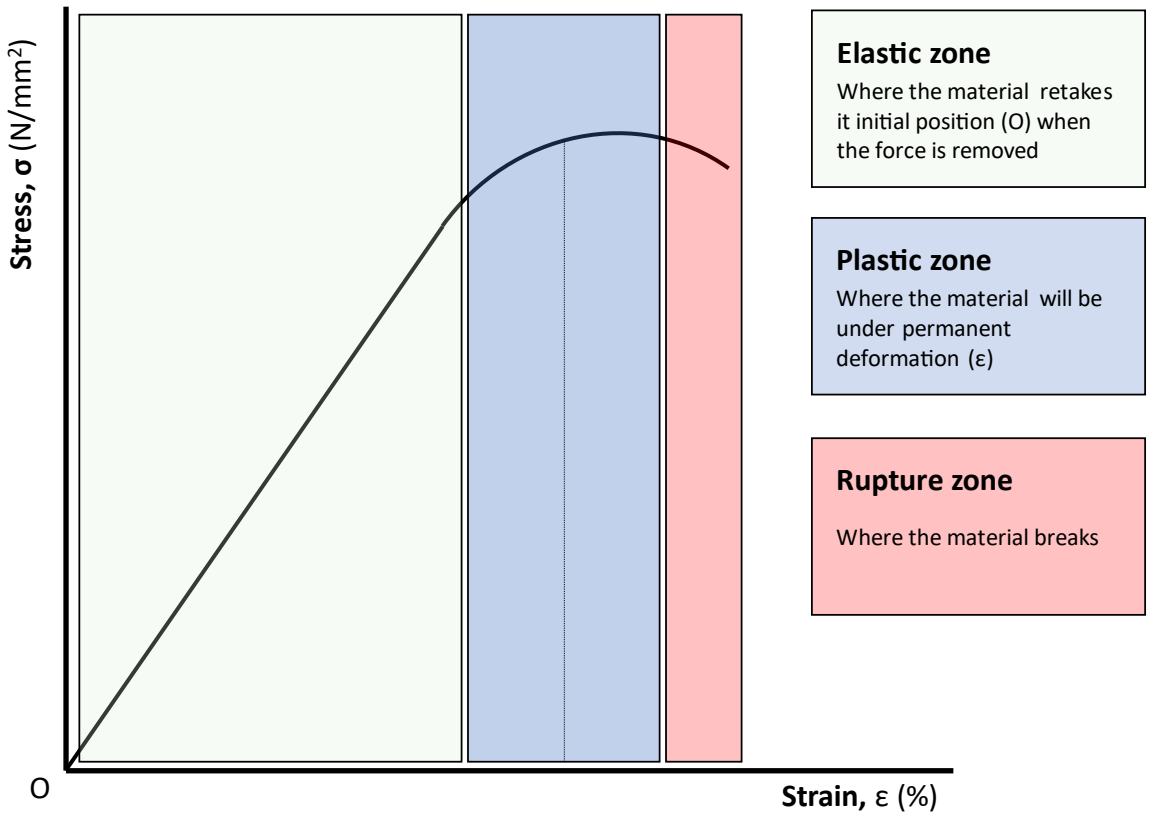


Figure 4. *Stress in N/mm2 expressed as function of strain in %.*

Since both stress and strain are normalized values, the Young modulus is a constant material property that allows for comparison across different systems and structures. For instance, a large piece of steel and a small chunk of steel have the same elastic modulus, but deforming the large piece requires a greater absolute force. Brittle materials, such as cast iron, typically have a higher elastic Young modulus than ductile materials like aluminum or rubber.

1.1.3. Biomedical engineering

In the field of biomedical engineering, the term "elastic stiffness" is frequently used to designate “stiffness”. Furthermore, one distinguishes “elastic stiffness” from the concept of "elasticity," which here pertains to damping. Thus, elasticity can be described as the manner

in which energy dissipation affects the oscillatory motion of a system. The damping ratio of an elastic structure (or tissue) goes hand in hand with the amount of energy dissipated during an oscillatory motion of a system, which returns to its original state prior the application of an external force (Figure 5).

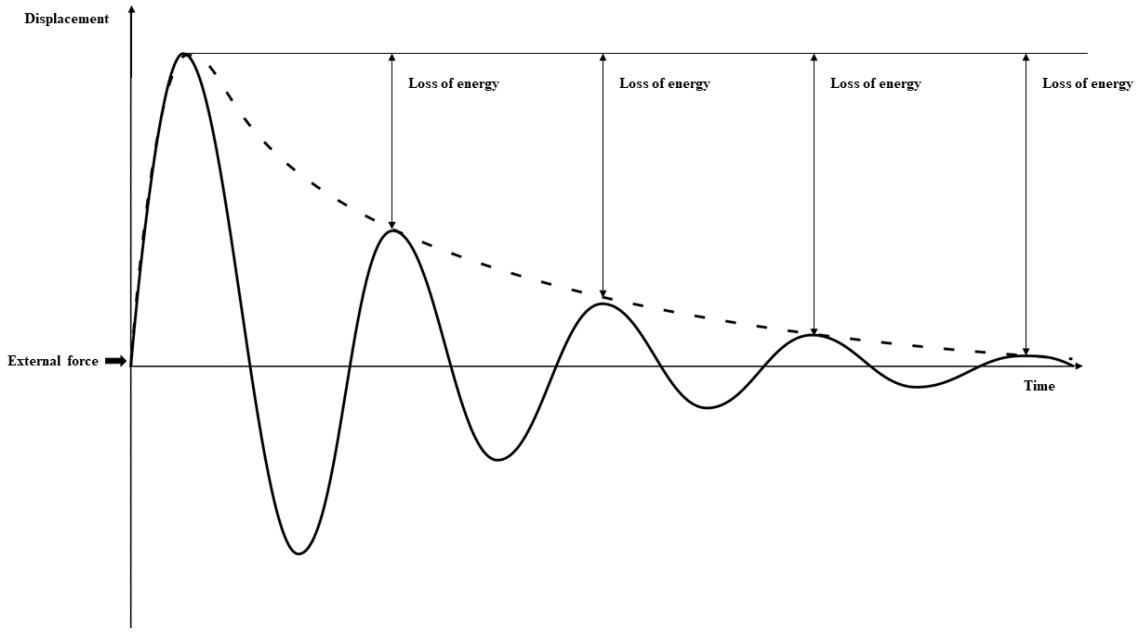


Figure 5. *Representation of the "elasticity" and its damping effect on a system demonstrating oscillatory decay after an initial external force application.*

In biomedical engineering, elastic stiffness, also represented by **k** (cfr. Mechanical engineering section), expressed in [N/m] is related to the elastic modulus through the following equation, rather similar to Equation 1:

$$k = F/\delta \text{ (Equation 5)}$$

Combining all these features/parameters and laws plays a crucial role in both human health and sports performance, particularly in relation to the concept of viscoelasticity. The viscoelastic properties of human tissues result in specific stress and strain behaviors over time, including phenomena such as time relaxation, creep or fluage, as well as the hysteresis effect.

Time relaxation. The time relaxation characteristics is observed when a material is stretched and held a constant length. Over time, the tension or external force required to maintain this range of motion decreases (Figure 6, left). For example, this viscoelastic property is the reason why when you stretch a muscle, the more you maintain the stretched position, the easier it gets.

Fluage (or creep). The condition known as the "fluage" effect, often referred to as "creep", occurs when a consistent external force applied to a viscoelastic structure causes a gradual deformation over time (Figure 6, middle). For example, if a cardboard box is placed at the edge of a table, it will slowly deform as time progress.

Hysteresis. Finally, the phenomenon of hysteresis helps to explain why human tissues are not perfect springs. Hysteresis refers to the energy dissipation that occurs within a viscoelastic material during the processes of loading and subsequent unloading. This phenomenon can be visualized as a loop on a stress-strain curve (Figure 6, right). Hysteresis is commonly observed in everyday situations, such as when applying and then releasing pressure on a sponge. After the pressure is released, the sponge does not immediately return to its original shape; instead, it slowly resumes its initial form over time. This behavior indicates that the energy is dissipated during both the deformation process and the subsequent recovery phase.

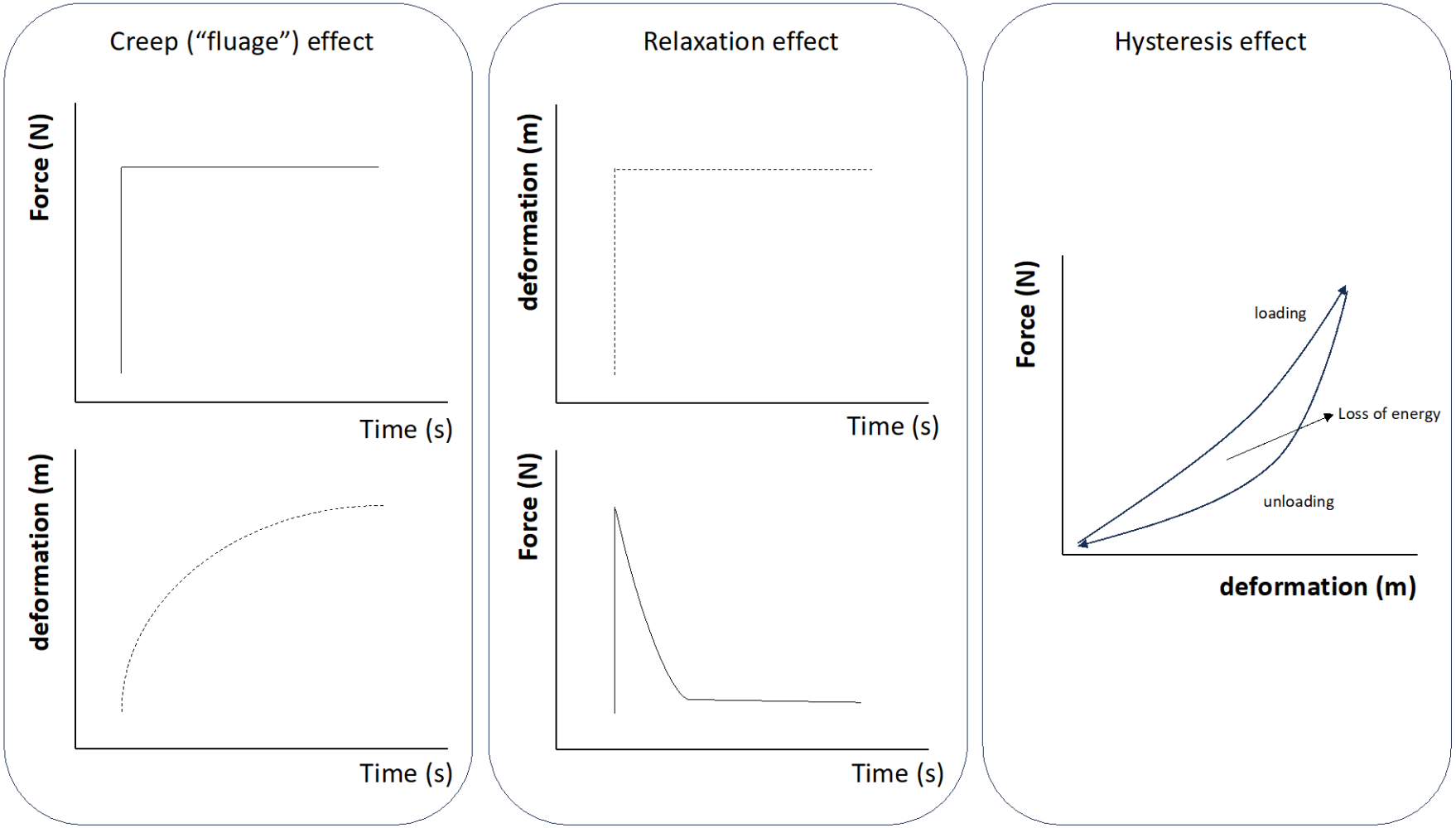


Figure 6. *Viscoelastic properties - Left: Creep/fluage effect, Middle: relaxation effect, Right: hysteresis effect..*

1.1.4. Viscoelasticity in Human

Exploring the rheological properties of muscles and tendons (*rheology being the branch of physics that studies how structures deform or flow under stress or forces)* could offer better insight and relevant guidelines in both rehabilitation and sports performance. Hill et al. [6] described the muscle-tendon unit with a specific mathematical model. The musculotendinous system can be depicted as a configuration of contractile elements that generate force (GF). Those contractile component rheological properties are also responsible for the viscous stiffness of the muscle. The contractile components are arranged in a series connection with a complex of elements with/producing an elastic stiffness (CES), i.e., the tendon and the titin protein, as well as connected in parallel with another elastic stiffness component (CEP), i.e., the fascia and connective tissue (Figure 7).

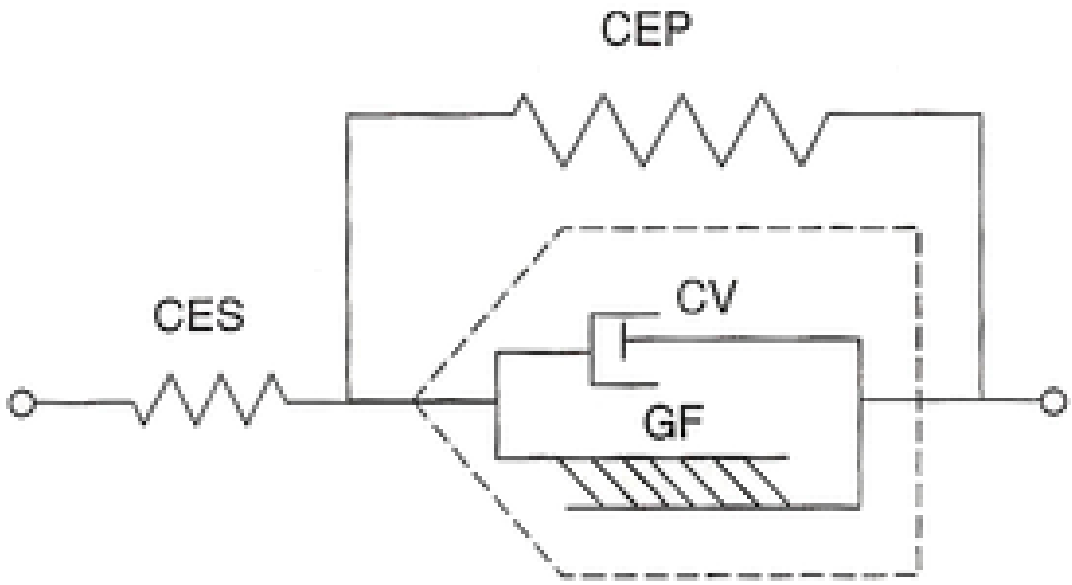


Figure 7. *Hill's model. GF represents the generator of Force, or contractile tissue, CV is the viscous component, CES represents the elastic serial component while CEP represents the elastic parallel component.*

Similar to the principle of materials engineering, we could explain myofascial elastic stiffness properties using a stress and strain curve. This approach allows us to understand how myofascial tissues behave under various levels of stress and how they deform (or strain) in response. By analysing the stress-strain relationship, we can gain insights into the elastic and plastic properties of these tissues, which is relevant for understanding their function and behaviour under physiological and pathological conditions.

1.2. The challenge of viscoelastic, elastic and viscous stiffness measurement in human

By definition, a measurement is quantifiable entity. In terms of viscoelastic stiffness, its quantification can be qualitative, like the Ashworth scale in neurology [7] or quantitative [8]. The purpose of measuring is to facilitate the comparisons, either against normative values or self-reported benchmark. This evaluation can then be interpreted to inform decisions in relevant areas, such as recommending specific therapy or exercise routines. Therefore, providing reliable and pertinent measurements of viscoelastic stiffness is crucial in the fields of sports or rehabilitation.

In the human body, viscoelastic stiffness is divided in two component, active stiffness that is influenced by the contractile element of the muscle and passive stiffness that can be connected to rheological properties of non-contractile element such as tendon, ligament, aponeurosis, … [9,10] Furthermore, when a joint was passively moved, and muscle stretched, it is possible to measure the muscle passive stiffness explained by conjunctive tissue into the muscle and the titin protein. [11] A this approach focuses on understanding the inherent properties of these tissues when they are not actively contracting, offering insights into how their passive characteristics contribute to human movement and health. Indeed, when a joint undergoes passive movement and the associated muscle is stretched, it is possible to measure the muscle rheology. The type of stiffness is attributed to the connective tissue within the muscle and the titin protein. [12] Given that various tissues can influence this type of stiffness, we refer to it as passive musculoarticular stiffness (PMAS). This term encompasses the combined stiffness properties of muscles, tendons, ligaments, and joint capsules when they are not actively contracting.

1.2.1. The Electromechanical Oscillation Device (EOD)

Building upon the established passive sinusoidal oscillation methodology for PMAS assessment, as outlined by Rack & Lehman [13,14] and further developed by Detrembleur in 2000 [8] , our research team has created the Electromechanical Oscillation Device (EOD) (Figure 8). This device is designed with minimal inertia, enhancing its portability and making it user-friendly for wrist applications. The device mobilized the wrist at frequencies ranging from 3 Hz to 12 Hz, increasing in 1 Hz increment. The movement was confined to a range of 10° of flexion and 10° of extension, resulting in a total of 20° of movement around the neutral position. A torquemeter attached to the gripping handle, and a potentiometer located at the center of rotation simultaneously recorded the torsional force and the displacement. Angle displacement and torque data were recorded at frequency of 1024 Hz and subsequently analyzed using a Fast Fourier transformation. The torque amplitude response that is in phase with the displacement indicates the elastic (stiffness) torque, while the torque amplitude that is 90° out of phase corresponds to the viscous (stiffness) torque. The effect of the handle and hand inertia was computed and subsequently subtracted, yielding the net elastic stiffness values. The net viscous stiffness values were determined by subtracting the device friction. Elastic stiffness is expressed in N m rad^{-1} and viscous stiffness in N m s rad^{-1}.



Figure 8. *Electromechanical Oscillation device of first generation for the ankle (left) and second generation for the wrist (right).*

The detailed methods for calculating PMAS, and more specifically elastic and viscous stiffness are outlined as follows:

The angular wrist displacement (θ in radian) was recorded by a digital coder.

Mr, the resisting moment (in N m) was recorded using a torquemeter.

All signals, including displacement and torque were recorded at 1024 Hz.

The equation for the resisting moment (M_r) can be writing as follows:

$$M_r = k.\theta + \eta.\frac{d\theta}{dt} + \eta'.\frac{d\theta}{dt}. + I.\frac{d^2\theta}{dt^2} \quad \text{(Eq. 6)}$$

$$M_r = k.\theta + (\eta+\eta').l.\frac{d\theta}{dt} + I.\frac{d^2\theta}{dt^2} \quad \text{(Eq. 7)}$$

where k represents the elastic stiffness coefficient in N m rad^{-1}; θ is the angular displacement in radian; η is the viscous stiffness coefficient in N m s rad^{-1} and η'is viscous stiffness (friction

component); $\frac{d\theta}{dt}$ represents the angular speed in rad s^{-1}; $\frac{d^2\theta}{dt^2}$ is the angular acceleration in rad s^{-2}, and I is the Inertia of system in kg m^2.

As the motion is sinusoidal, we can express the equations as:

$\theta = A\sin(\omega t)$, where θ is the angular displacement in radian

$\frac{d\theta}{dt} = A\,\omega\cos(\omega t)$, which represents the angular speed or first derivative of θ, in rad s^{-1}

$\frac{d^2\theta}{dt^2} = -A\omega^2\sin(\omega t)$, which is the angular acceleration or second derivative of θ, in rad s^{-2}

When substituting the sinusoidal motion terms into your equation, you can replace the angular displacement (θ), angular speed (dθ/dt), and angular acceleration ($d^2\theta/dt^2$) terms with their sinusoidal expressions. This will allow you to express the resisting moment (Mr) in terms of the amplitude (A), angular frequency (ω), and time (t).

$$M_r = k.A\sin(\omega t) + (\eta+\eta').A\cos(\omega t) + I.-A\omega^2\sin(\omega t) \quad \text{(Equation 8)}$$

where each term is now expressed in terms of the sinusoidal motion components. This substitution helps in understanding how the resisting moment changes with time in a sinusoidal motion.

When highlighting the terms involving sine and cosine in the equation for the resisting moment (Mr), it helps to clearly identify the contributions of the different components of the motion. Here's the revised equation with the sine and cosine terms highlighted:

$$M_r = (k. -I\omega^2)A\sin(\omega t).1 + (\eta+\eta').A\,\omega\, t\cos(\omega t).1 \quad \text{(Equation 9)}$$

where $(k. -I\omega^2)A$ represents A_1, or total elastic stiffness, and $(\eta+\eta').A\,\omega$ represents A_2, or total viscous stiffness. A Fast Fourier Transformation was applied to the angular wrist displacement to calculate A (amplitude).

A Fast Fourier Transformation was applied to the torque curve to calculate A1 and A2.

Then, A_1 and A_2 were divided by A. The result of A_1/A represents the total elastic stiffness, which can be mathematically expressed as $(k. -I\omega^2)$.

A graph was plotted of A1/A as a function of the square of the oscillation frequency (ω^2). A linear regression was then fitted through the points to calculate the system's inertia (I). Finally, the elastic net stiffness, at each oscillation frequency (i), can be calculated using the following formula:

$$k_i = A_{1i} - I\,\omega^2 \quad \text{(Equation 10)}$$

The frictional moment of the device (η') is determined by operating the device without any load, thereby capturing the inherent viscous component of the system. This value is then subtracted from total viscous stiffness. Subsequently, the viscous stiffness, at each *i* frequency, can be calculated using the following formula:

$$A_2/A = (\eta+\eta').\,\omega \quad \text{(Equation 11)}$$

$$\eta_i = A_{2i} - \eta' \quad \text{(Equation 12)}$$

To calculate the elastic stiffness, we constructed a graph plotting the net elastic stiffness as function of the oscillation frequency. A linear regression was then applied through the data points. The intercept of the fitted line on the vertical axis represents EL, which is the elastic stiffness coefficient. Similarly, to determine the coefficient of viscous stiffness, we plotted a graph of the net viscous stiffness against the oscillation frequency. A linear regression was then applied through the data points. In this case, VI is represented by the slope of the fitted line (Figure 9).

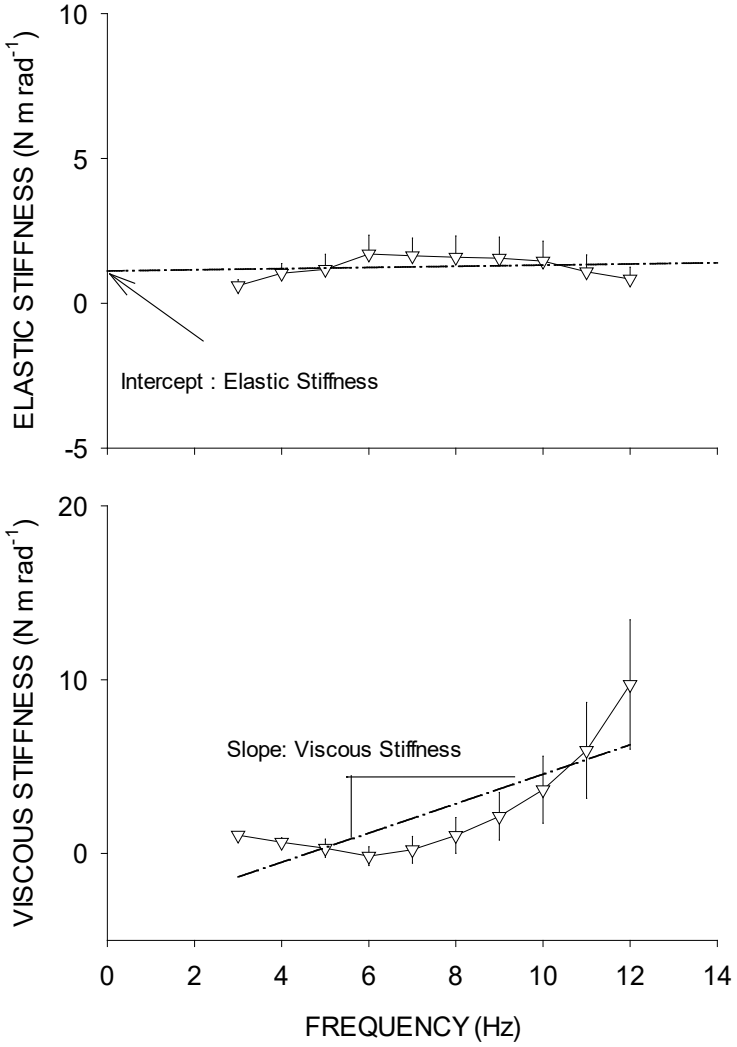


Figure 9. *The figure displays the values of elastic stiffness (top) and viscous stiffness (bottom) at varying frequency (plotted on the x-axis) for healthy 20 years subjects. To determine the final value of elastic stiffness, a linear regression is applied to the elastic stiffness values across the different frequencies. The intercept of this regression line on the y-axis represents the elastic stiffness value (EL). Similarly, the final viscous stiffness value is derived from the slope of the linear regression line plotted against the viscous stiffness values at each frequency. This approach enables the quantification of both the elastic and viscous components of stiffness in the study population. (adapted from Nguyen et al. 2020)*

1.2.2. Elastrography

Elastography is a medical imaging technique that uses ultrasonography principles to measure varying degrees of elastic stiffness. Elastography operator sometime use the term “elasticity” to refer to elastic stiffness. They use this vocabulary to distinguish between “stiff” and “elastic” structures. Ideally, a perfectly “elastic” structure would show minimal “stiffness”, and vice-versa. In the context of this terminology, 'stiffness' in elastography could be analogous to the viscous stiffness we discuss in the previous section. However, Elastography devices measure “elastic stiffness” per se. Indeed, elastography has become a relevant diagnostic and assessment tool in various medical conditions, particularly in examining the breast, liver, and muscle tissues. [15] Elastography can be categorized into two main types: strain elastography and shear wave elastography (Figure 10). Each type operates on distinct principles and has specific practical applications.



Figure 10. *Shear Wave Elastrography device*

In strain elastography, the operator applies physical pressure using the ultrasound probe, causing tissue deformation. This deformation is typically measured in a two-dimensional (2D) zone beneath the transducer that covers the entire field of view. The principle behind this technique is that softer tissue will deform or strain more than stiffer tissue when subjected to the same amount of applied stress. This variance in tissue strain is visualized on the elastogram, typically represented through color or gray scale variations. In these images, differences in colours or shades correspond to varying degrees of tissue stiffness, with softer areas showing more deformation (higher strain) and stiffer areas showing less deformation. [16] Despite the limitations of using strain elastography to evaluate "stress" (understand here, elastic stiffness or k) distribution within tissue, there are methods for transforming qualitative data into semiquantitative data, which can be particularly useful in clinical diagnostics and assessments.

Shear wave elastography (SWE) represents an advanced form of elastography that like strain elastography assesses tissue elastic stiffness. However, SWE employs a different principle: it utilizes acoustic radiation force to generate shear waves within the tissue. These shear waves then propagate perpendicular to the ultrasound beam, allowing for the measurement of tissue elastic stiffness. The key distinction between SWE and strain elastography lies in their approach to measuring stiffness: strain elastography primarily measures elastic stiffness longitudinally through manual compression with the probe, whereas SWE evaluates elastic stiffness perpendicularly to the ultrasound beam without the need for manual compression. This capability to measure elastic stiffness perpendicularly gives SWE a significant advantage in providing objective and quantifiable assessments of tissue elastic stiffness. [17] To visualize how SWE works, imagine dropping a stone into a still pond. The ripples that radiate outward from the point of impact are like the shear waves generated in tissue during an SWE procedure (Figure 11).

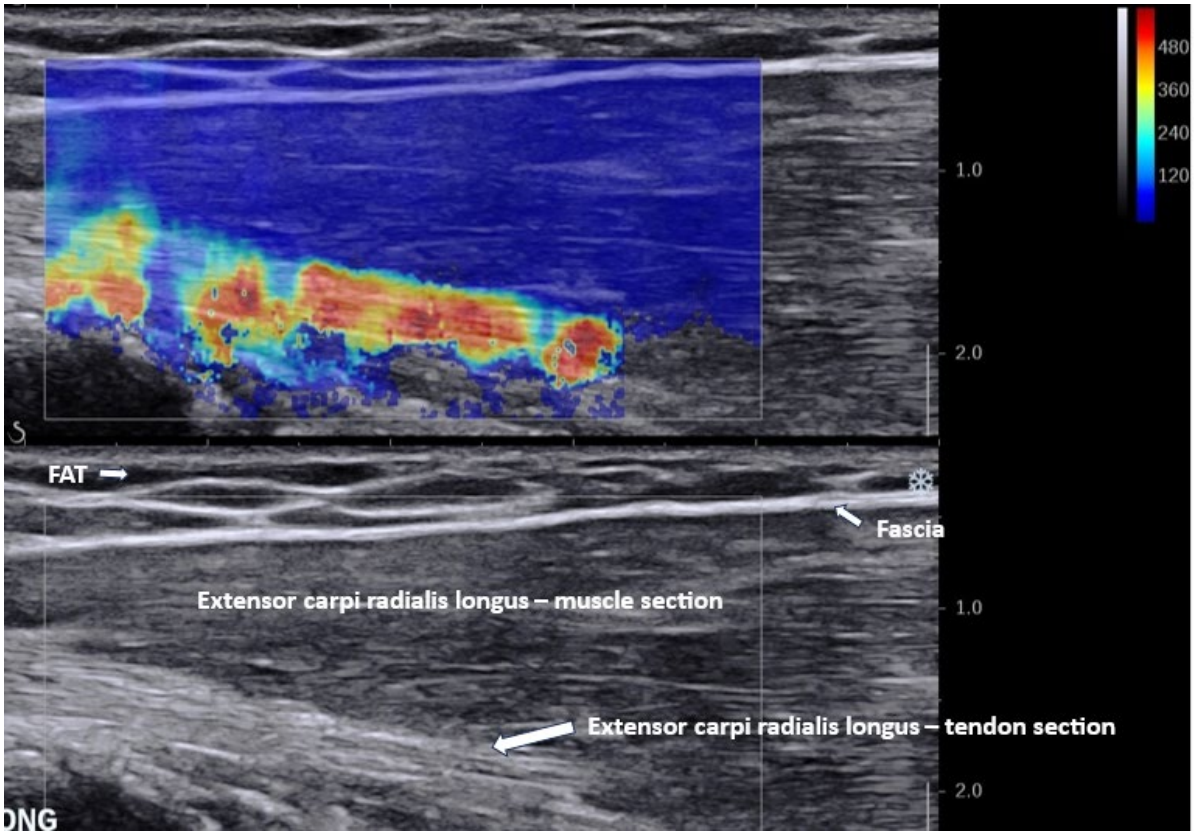


Figure 11. *The image at the top shows a shear wave elastography of the extensor carpi radialis longus muscle, with the elastic stiffness values displayed in Pascal (as indicated by the scale on the right). The elastography image demonstrated that the muscle section exhibits lower elastic stiffness (indicated by the blue color) compared to the tendinous section (indicated by the red color). The bottom image is a B-mode ultrasound of the same area, providing a structural view of the muscle and tendon. This combined representation allows for a detailed assessment of tissue elastic stiffness within the muscle-tendon unit, highlighting the differences in mechanical properties between the muscle and tendinous sections.*

1.2.3. Myotonometry

The Myotonpro (Figure 12) is a handheld device that use myotonometry to apply a rapid (0.015 s), single mechanical impulse of 0.4 N under a constant compression force of 0.18N to the subcutaneous tissue layer. [18] It is thought to be able to report viscoelastic stiffness in human tissue.



Figure 12. *MyotonPro device. The properties of the sensor are follows: amplitude, range of about 8g in full range ; resolution of <0.001g ; output data rate and bandwidth 3.200 Hz ; sensitivity of <20mg/LSB for each axis ; sensitivity about 0.1% due to the temperature change ; bias level of each axis 100mg ; noise performance of each <1.5 LSB rms ; and oscillation acceleration signal.*

The MyotonPRO technique use viscoelastic principle and more specifically the gradual elongation of tissue over time under consistent tensile stress. [5] The single external mechanical shock delivered by the device induces a serie of damped oscillations which is captured by an acceleration sensor (Figure 13). Typically, the MyotonPRO probe is positioned perpendicular to structure being evaluated. The total force applied to the structure is 0.58 N. After delivering a single external mechanical stimulus, the device records 400 ms of free oscillations in the tissue. It then takes 10 ms for the device to process the raw signals, followed by an additional 5 ms to calculate all the relevant biomechanical parameters. This rapid processing enables efficient and timely data collection during assessments. The average of the five sequences conducted with the Myotonpro corresponds to a single evaluation. This averaging process enhances the accuracy and reliability of the data. From this evaluation, it is possible to calculate five key biomechanical parameters: Frequency (F) represents the oscillation frequency and

indicates muscle tonus or tension when the muscle is contracted: F [Hz]=Fmax (Equation 13). Stiffness (S) denotes the elastic stiffness of the structure and is calculated using the following equation:

S [N/m]=(a_max.m)/l (Equation 14), where a is probe acceleration in m s^{-2}, m is probe mass in kg, and l is displacement in m. The logarithmic decrement, which represents the rate of decrease in oscillation amplitude between the first and second cycles, is defined as the decrement (D). It quantifies energy loss, characterizing the system's elasticity: D=ln (a_1/a_3) (Equation 15) where a_1 and a_3 are the peak accelerations in m s^{-2} of the first and second cycles. A decrement value of 0 indicates perfect elasticity, with no loss of elastic energy (Figure 5). The relaxation time is the duration between the peak value and the end of the first cycle. It indicates the time taken for the structure to return to its initial position. Creep represents the Deborah number, a dimensionless value characterizing a material's fluidity under specific conditions. [19] A higher creep (or Deborah number) signifies a more solid-like behavior of the material.

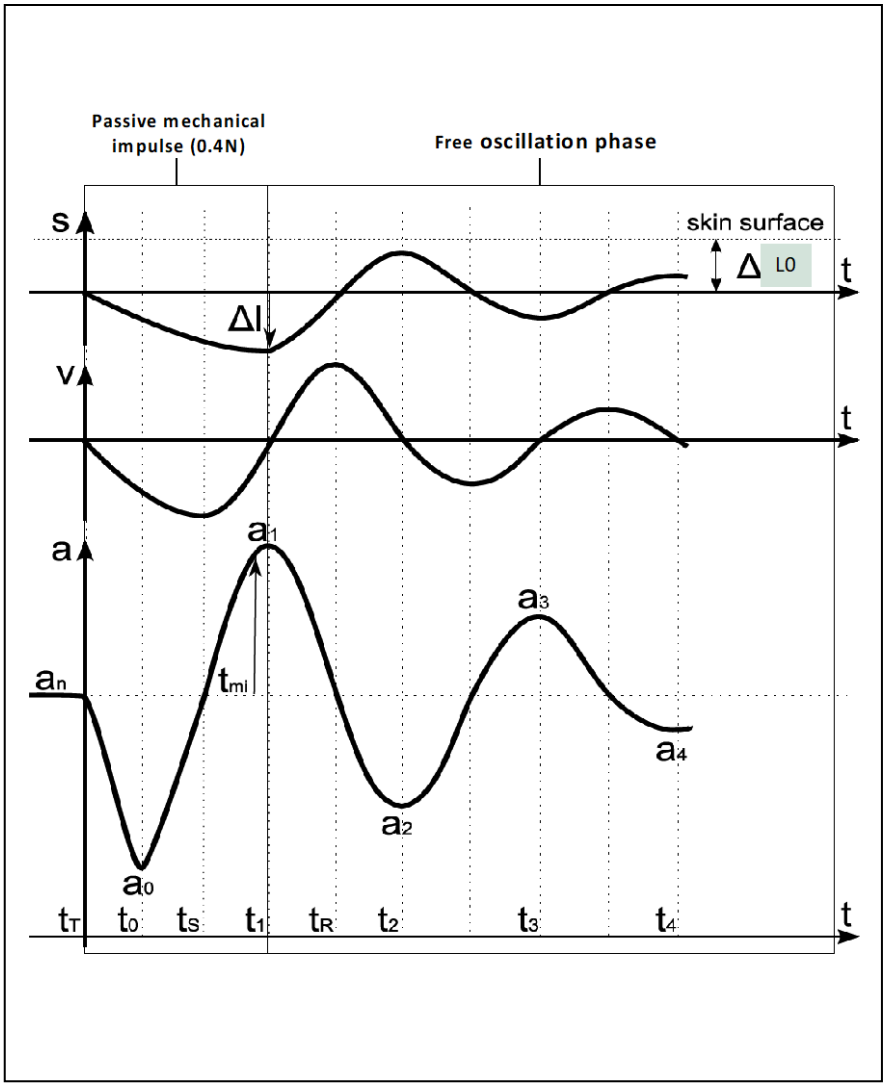


Figure 13. *From Nguyen et al.* [20] *The description of the induced mechanical impulse and the free oscillation phase in a system, like the one used in device such as MyotonPro, can be understood examining the relationship between displacement (l), velocity (V) and acceleration (a) (Top to bottom curves). Adapted from the MyotonPRO user manual (MyotonPRO AS, London, United Kingdom) and previously described by Schneider et al. in 2015.* [5]

1.3. Running: an interesting laboratory for stiffness exploration

1.3.1. Running

To better understand stiffness, we can illustrate its impact on various daily life activities. One activity that offers relevant opportunities to study stiffness is running, and consequently, its influence on running performance. According to evolutionary biologists, efficient endurance running played a crucial role in the survival and expansion of the Homo sapiens. [21,22] It is believed that early human ability to run contributed to their survival, leading to modifications in our physical form such as longer legs and a sturdier upper body. [21] Further along the timeline, running was celebrated as a sport by the Ancient Greeks starting in 776 BC during the first Olympic game, and the first modern marathon took place during the Athens Olympic game held in 1896. Running has had a significant societal influence and has been utilized to bring about social change, as exemplified in the women's rights movement. In 1967, Kathrine Switzer challenged the all-male tradition of the Boston Marathon. [23] Running also fosters community development, as evidenced by the proliferation of running clubs and events worldwide. To underscore this, it is estimated that there are more than 20 million “regular” runners in the United States of America, [24] and in 2023, over 40 000 participants registered for the 20 km race in Brussels, Belgium.

Nowadays, running is easily recognized as one of the most popular recreational sports activities. This popularity can be attributed to its accessibility, as well as its potential health benefits, which include improvements in musculoskeletal and cardiovascular health [25] and psychological well-being. [26] Regular participation in running has been linked to enhanced cardiovascular health, effective weight management, and potentially a longer lifespan. [27] Running is acknowledged as beneficial a non-pharmacological approach for addressing mental health conditions, in addition to its physical benefits. It has been shown to be an effective therapeutic intervention for conditions such as depression and anxiety. [28,29] Furthermore, running physiologically triggers the release of endorphins, a class of chemicals that function as natural painkillers in the brain and nervous system. This release leads to sensations of exhilaration and a significant reduction in stress levels, a phenomenon commonly referred to as the "runner's high". [30]

External factors also play a significant role in motivating individuals to run. Organized running events can encourage to challenge themselves, set and achieve defined goals, and experience a

sense of achievement. Additionally, the camaraderie formed through training together and sharing experiences can foster a sense of friendship and community, further augmenting the social and psychological benefits of running. [31]

Given running's widespread appeal, scholars have developed an interest in it from a locomotion perspective to a multidisciplinary approach, covering fields like sports science, health, psychology, and sociology. The growing number of studies featuring "runner" AND "running" published on PubMed from 1954 to 2022 shows a marked increase in academic interest in the field since the mid-20th century (Figure 14). This surge not only follows the trend of running's growing popularity but also reflective of the overall expansion in research across various topics.

In the world of running, every stakeholder – be it athletes, coaches, medical professionals, researchers, or spectators – shares common goals: improving running performance and recovering from or preventing running-related injuries.

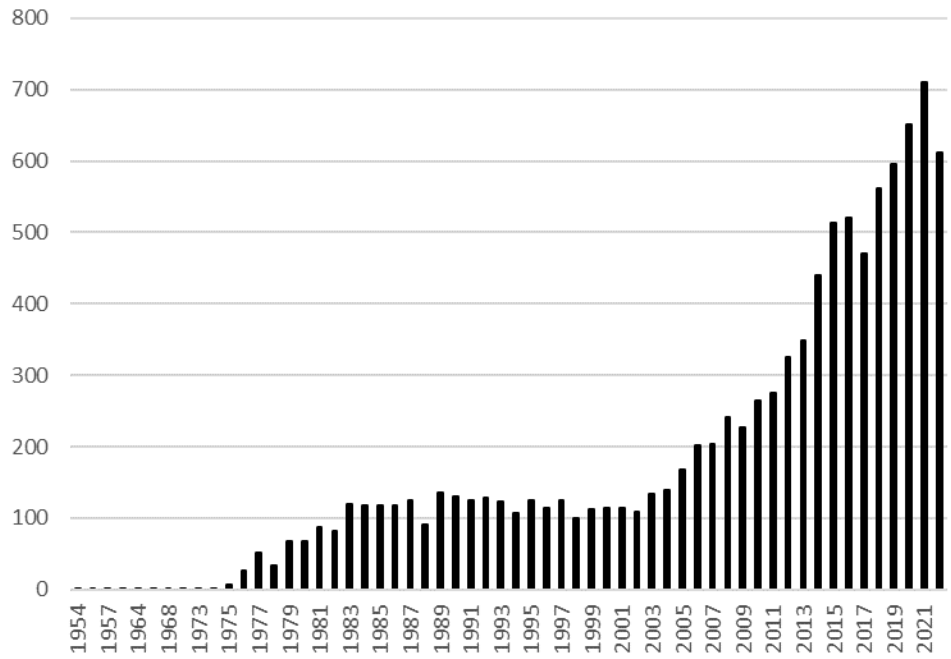


Figure 14. *Histogram displaying the annual number of references identified in Medline/Pubmed using the term: "runner" OR "running".*

Given the available data, the practice of running, particularly in context of runners, presents significant opportunities for investigating stiffness and the effect of muscle rheological properties within the realms of sports and rehabilitation. Various models have been proposed to explain the mechanism of running, with the most straightforward and widely used being the spring mass damper (SMD) model[32] (Figure 15). The model can be presented as a passive one-body system [33]or as a multibody model [34] to elucidate the mechanical behavior of the runner's leg. Unlike those passive models, active SMD models include muscle contraction in their analysis. [32] However, this thesis will focus exclusively on the **passive aspect** of the model.

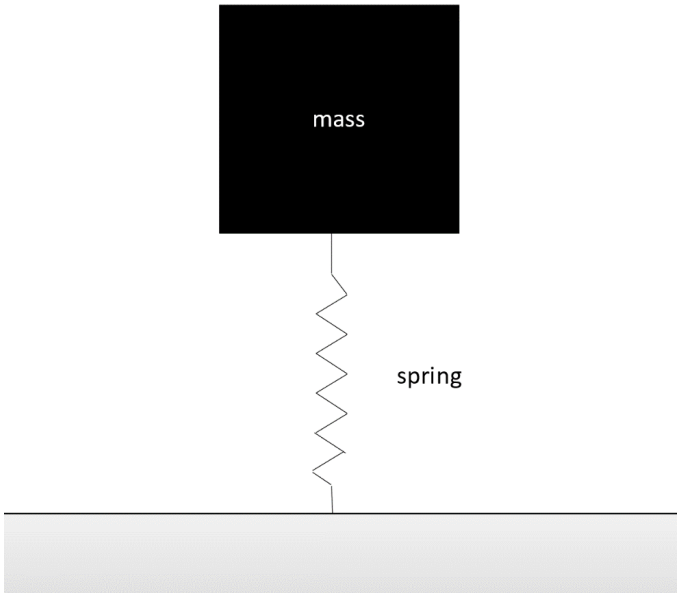


Figure 15. *The one body spring mass model representing the leg of the runner as a spring and the body of the runner as a uniform mass.*

1.3.2. Why is it important to measure elastic stiffness in athletes?

In the field of sports, understanding tissue stiffness is indeed a crucial factor in enhancing athletic performance and reducing the risk of injury. The rheological properties of tissues, particularly tendons, play a significant role in the overall biomechanics of movement. The mechanical characteristics of tendons are crucial in facilitating the passive transmission of forces from muscles to bones, providing resistance against abrupt stretching, and serving as a mechanical buffer that enhances the efficiency of muscle function. [35]

The rheological properties of muscle tissue can vary based on several factors, such as genetic predisposition and environmental influences. Understanding the biomechanical characteristics of muscular tissues is greatly influenced by muscle fiber type. [36] It is well established that long distance runners, particularly those competing in events over 5000 meters, tend to have a higher proportion of slow twitch muscle fibers in comparison to short distance runners. [37] This type of muscle fiber composition plays a significant role in their performance and endurance during long-distance events. The characteristics of different fiber types can influence various aspects of running. These include the running technique, the running regimen, and physiological factors like running economy. [38] Bellinger et al. have suggested that the rheological characteristics of slow twitch fibers may serve as a protective mechanism against certain types of injury. Slow twitch fibers are distinguished by their more substantial anatomical structure, which included characteristics such as thicker Z-line and M-line. These features make athletes with a predominance of slow twitch fiber more resistant to fatigue. [38] In contrast, fast-twitch fibers are reported to be five times more susceptibility to injury compared to slow-twitch fibers. [39] However, the vulnerability to injure is highly dependent on the specific sport or physical activity in which the athlete is involved. Athletes with a predominance of fast-twitch fibers generally engage in activities that are characterized by shorter durations, increased speed, and greater explosiveness. These activities often include sprinting or jumping, where the demands on the muscles are intense and immediate.

In general, athletes who specialize in various forms of running, including short distance and long-distance events, may exhibit distinct rheological properties. These properties can significantly influence the stiffness characteristics of their muscle tissues while running. To gain a deeper understanding of muscle adaptation in this context, numerous studies have focused on investigating how running affects the stiffness of the Achilles tendon. [40–43] Houghton et al. (2013) observed that either maintaining a consistent running speed or engaging in a 60-minute running session did not significantly affect the properties of the Achilles tendons. [44] Conversely, Fletcher et al. (2010) discovered that the stiffness of the Achilles tendon decreased after a 90-minute run performed at or below the lactate threshold. [45] Struzik et al. (2021) reported that tendons with greater stiffness have the capacity to store more energy, which can lead to reduce energy expenditure during physical activities. [46] Individuals who do not participate in running activities typically show lower levels of Achilles tendon stiffness compared to athletes. Numerous studies have demonstrated that engaging in plyometric, isometric, and dynamic heavy resistance training regimens is associated with significant

improvements in both muscular strength and stiffness. [18,35,40,47–49] Research has shown that tendons with a larger cross-sectional area demonstrated increased force and enhanced storage of strain elastic energy. [50] As a result, a tendon that is thicker and stiffer can store a greater amount of elastic energy. [51]

Numerous studies reported that individuals with Achilles tendinopathy often exhibit a decrease in the stiffness of their Achilles tendons. [52–54] A reduction in tendon stiffness could indeed lead to increase muscle shortening as a means of compensating for the increased compliance of the tendon. [55] A decrease in stiffness would impede the ability to generate quick movements and negatively affects the rate of force production. [56] The cumulative effect of these mechanical changes could compromise overall functional performance. Furthermore, a tendon with reduced stiffness, typical in diseased conditions, undergoes greater strain. This increased strain can elevate the risk of developing small tears within the collagen fibers. Over time, the cumulative effect of these microtraumas may make the tendon more prone to injuries and prolong the healing process. While further clinical research is necessary to confirm the relationship between tendon stiffness and soft tissue injuries, it may be beneficial to routinely assess tendon stiffness to monitor potential improvements, maintenance, or decline in tendon function. The ability of the Achilles tendon to efficiently store and release elastic energy during locomotion and explosive activities, such as jumping, is compromised in individuals diagnosed with Achilles tendinopathy. As a result, there is a reduced capacity for energy conservation and an increase dissipation of energy as heat in case of Achilles tendinopathy.

1.3.3. Running on Springs

Numerous studies have described the relationship between excessive or insufficient level of elastic or viscous stiffness in the lower limb and a potential higher risk of injury. [52,57,58] Given the multifaceted nature of lower limb injuries, it can be inferred that an optimal level of lower limb stiffness is necessary for both performance enhancement and injury prevention. A high level of the lower limb stiffness is associated with increased peak forces of ground reaction force and amplified loading rates on the lower limb's structures, elevating the risk of bone injuries and stress fractures. [59] On the other hand, a low level of lower limb stiffness may lead to soft tissue injuries. [57]

During the stance in running, the legs compress similarly to a spring under the body's mass (Figure 15), facilitating the investigation of the storage and release of potential elastic energy utilized for propelling the runner forward. This understanding allows researchers and practitioners to explore how various types of footwear or running biomechanics might influence overall running performance and efficiency. The SMD model, extensively utilized in this context, enable the examination of multiple variables including leg stiffness, vertical stiffness, and joint stiffness.

As outlined in the SMD model, running stiffness can be categorized into passive and active stiffness. Active stiffness involves the calculation of the muscle contraction unit. Specifically, this model considers the pre- landing activity of the lower limb muscles, making the SMD sensitive to external factors such as the footwear, the ground compliance, or other external stimuli. [3] This active model enhances our understanding of running biomechanics by considering how muscle activity and external factors can influence the stiffness of the lower limb tissues. [32] On the contrary, passive models do not account for contractile activity. Although dynamics stiffness has been extensively studied [60], investigating the rheological properties of musculoskeletal tissue, including muscles, tendons, or entire joint complexes, is potentially significant in the context of running and for overall human well-being [61,62] (Figure 16).

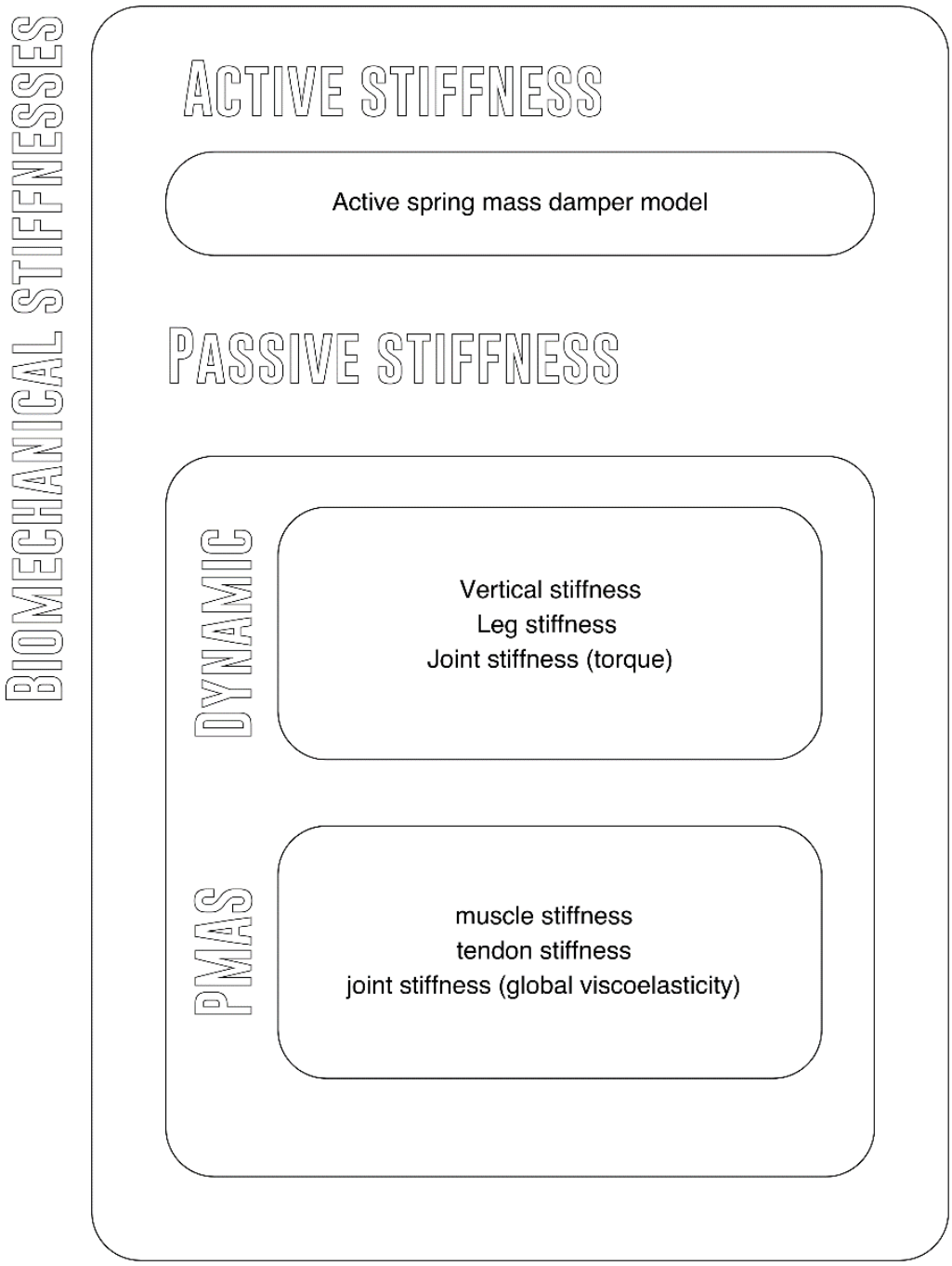


Figure 16. Summary of biomechanical stiffnesses in running. Active stiffness: this type of stiffness involves measurements taken when muscle contraction is actively occurring and is being measured. Dynamic stiffness represents Kleg, or leg stiffness, which is the ratio of maximal force to maximal compression.[1] Linear vertical motions like jumping and hopping are referred to as "vertical stiffness" (Kvert). The most widely used technique is McMahon and Cheng's [2] vertical stiffness technique. It considers the maximum vertical force and maximum

vertical displacement of the center of mass (COM), assuming they peak during mid-stance. Vertical stiffness is computed as the peak vertical force divided by the greatest vertical displacement. Joint stiffness (Kjoint) measures a joint's angular displacement in response to a moment. It is calculated by comparing the joint moment to its sagittal plane angular displacement. [3,4] Finally, PMAS is Passive musculoarticular stiffness or rheological stiffness reported by musculoskeletal tissues.

2. Objectives of the thesis

The EOD has been validated in previous studies [63,64] and is proficient in evaluating both the muscle-tendon unit and the PMAS of distal joints, including the wrist, ankle, or finger. This capability makes the EOD a valuable tool for assessing the stiffness characteristics of these specific joints. However, it is important to note that, to the best of our knowledge, there is no available device to measure PMAS in proximal joints (closer to the body's center, such as the shoulder or hip) with the same level of accuracy as the Rack & Lehmann device, which is recognized for its precision in such measurements. The Rack & Lehmann device remains the golden standard for PMAS assessment in distal joints. Alternative methods currently available, such as magnetic resonance elastography [65], ultrasound palpation system transient elastography, and SWE [66] have proven inadequate in recording the viscous stiffness in humans. [8,67] Indeed, these devices are either expensive, heavily dependent on the operator skill, or lack portability, which limits their practicality for routine clinical use. Implementing these technologies requires considerable financial resources and careful logistical planning, posing significant challenges in both research and clinical environments. For instance, both SWE and ultrasonography require an experienced operator and a stable electricity-equipped environment for continuous operation. Additionally, data analysis demands expertise. These requirements restrict the feasibility of including a large cohort of subjects for PMAS analysis in humans. Conversely, EOD technology facilitates the precise measurements of PMAS and offers a user-friendly interface, making it accessible even to individuals without specialized training. This advantage broadens the scope for extensive human in PMAS research. This opportunity permits the exploration of how age and human characteristics influence PMAS, thereby offering valuable statistical insights for a deeper understanding of PMAS.

Consequently, the **first objective** of this thesis was to investigate **the effect of age and anthropometric factors on PMAS in healthy population**. This research aimed to establish normative data on PMAS across a diverse age range (from 3 to 90 years), contributing to a more comprehensive understanding of PMAS in various demographics. Additionally, this thesis represented the first investigation to reveal the true accuracy of PMAS measurements using the EOD in conjunction with axillary block injection on the upper limb. This approach aimed to validate the EOD's capability in accurately assessing PMAS, ensuring that the

measurements are reflective of the intrinsic stiffness properties of the musculoskeletal system, unaffected by voluntary or reflexive muscle contractions.

Despite the portability of EOD technology, it necessitates access to electricity and secure environment for operation. As a result, alternative technologies are gaining attention in the scientific community. One such example is the MyotonPro, an innovative handheld diagnostic tool that captures parameters related to the tension state, biomechanical characteristics, and viscoelastic properties of superficial structures by recording and analyzing an acceleration signal. [5] While the MyotonPro offers promising capabilities, researchers and clinicians should approach its use with caution due to the current lack of extensive published validation studies. The MyotonPro device claims to record both biomechanical stiffness – referred to as elastic stiffness - and viscoelastic properties, including viscous stiffness. However, the reliability and accuracy of these measurements still require further validation through rigorous scientific research to ensure their effectiveness and applicability in clinical and research settings.

The second objective of this thesis was to ascertain **the validity of MyotonPro technology in measuring viscoelastic properties in both artificial and human tissues**. To fulfill this aim, we conducted a comparative analysis of measurements obtained using the MyotonPro and the EOD in a sample of healthy individuals. Furthermore, this section of the manuscript detailed the application of the MyotonPro device on the ankle plantar flexor muscles. This aspect is particularly noteworthy as the MyotonPro offers the capability to directly measure PMAS in runners in a field setting, thereby providing valuable insights in sports and rehabilitation contexts.

The third objective of this thesis was **to investigate PMAS measurements in healthy runners under various running conditions**. Specifically, we focused on three distinct running conditions following specific objectives. The first condition involved analyzing the correlation between PMAS of ankle plantar flexor muscles and net metabolic cost of running during a submaximal trial. The second condition entailed examining the change in ankle plantar flexor muscle PMAS during an exhaustive incremental run. Finally, the third conditions consisted of a 10 km running session that allowed us to better understand PMAS during a longer run. This could be relevant as we were able to compare those results with the two previous investigations and see adaptations in PMAS related to exhaustion level, speed, and duration. The methodology of this thesis was novel in that it included measurements taken during the run

(per-protocol) and before and after the run. This approach allowed for an in-depth observation of the dynamic changes in PMAS throughout several running conditions.

In order to address all the objectives, the present thesis is divided in three parts.

2.1. Unravelling Passive Musculoarticular Stiffness in Human.

In this section, we conducted multiple studies to deepen our understanding of PMAS in healthy human. Specifically, we aim to address the **first objective** with the following questions:

- How does age and various anthropometric characteristics influence PMAS in a healthy population?

  Nguyen, A. P., Herman, B., Mahaudens, P., Everard, G., Libert, T., & Detrembleur, C. (2020). Effect of age and body size on the wrist's viscoelasticity in healthy participants from 3 to 90 years old and reliability assessment. *Frontiers in Sports and Active Living*, *2*, 23.

- Does EOD trully measure passive stiffness?

  Libert, T., Detrembleur, C., Melebeck, F., & Nguyen, A. P. (2024). Validating the measurement of passive Musculo-articular wrist stiffness without intentional or reactive contraction using axillary plexus block. *Clinical Biomechanics*, 106190.

By answering these questions, we hope to provide a more comprehensive understanding of PMAS and its variability among different individuals, as well as to evaluate the effectiveness of the tools used to measure it.

2.2. Validation of MyotonPro technology for passive musculoarticular stiffness measurement.

While the EOD approach may not be practicable for use in therapeutic setting, myotonometry is increasingly being utilized in clinical and ecological environments. This section of the thesis focus on **the second objective**, i.e. evaluating the validity of MyotonPro in comparison to the established and validated method of EOD. Additionally, it aims to identify the most suitable locations for stiffness measurements in runners using Myoton. This section will address the following key questions:

- How does the MyotonPro device's measurements of viscoelastic properties compare to those obtained by the EOD in terms of accuracy and reliability?

  Nguyen AP, Detrembleur C, Fisette P, Selves C, Mahaudens P. MyotonPro Is a Valid Device for Assessing Wrist Biomechanical Stiffness in Healthy Young Adults (2022). *Front Sports Act Living. 2022 Feb 21;4:797975. doi: 10.3389/fspor.2022.797975.*

  - What are the optimal sites on a runner's body for stiffness measurements using the MyotonPro device to obtain the most relevant and reliable data?

  Nguyen AP, Verdebout J, Fontaine A, Detrembleur C, P Mahaudens.(2024) Recommendation on myotonometry stiffness measurement for gastrocnemii and Achilles tendon at rest and in load conditions. European rehabilitation Journal, Accepted

The answers to these questions will provide valuable insights into the practical applications of the MyotonPro device in sports science and rehabilitation, especially for runners, and will help to advance our understanding of how to effectively measure and interpret PMAS in different populations.

2.3. Passive musculoarticular stiffness of the lower limb in runners.

The accessibility of MyotonPro allows for testing of PMAS before, after, and even during a run. This capability enables a more comprehensive understanding of the PMAS of ankle muscles in runners. In this section, we aimed to address the **third objective** of this thesis with the following questions:

- Is there a relation between Achilles tendon stiffness and running economy?

  AnhPhong Nguyen, Laurent Bosquet, Rayan Belaoued, Philippe Mahaudens, Christine Detrembleur (2024). The impact of Plantar Flexor stiffness on Running Economy and Biomechanics. In review process, Sport Biomechanics

- Changes of lower leg stiffness properties during an incremental running protocol

  AnhPhong Nguyen, Laurent Bosquet, Laureen Cugnet-Evans, Christine Detrembleur, Philippe Mahaudens (2024). Observation of the Achilles tendon and gastrocnemii

muscle's passive stiffness during an incremental running protocol. In review process, International Journal of Sports Physiology and Performance

- Changes of lower leg stiffness properties in a 10 km run.

  AnhPhong Nguyen, Christine Detrembleur, Philippe Mahaudens. Variation of lower limb tendon and plantar fascia in a constant velocity 10k run. *In submission process in Journal of Sport Biomechanics*

Overall, the findings from these studies underscore the importance of considering PMAS in human with an illustration on running. These studies enhance the title part of this thesis "from the lab to the track" and could hopefully offer further reflexion to better comprehend the impact of PMAS on running performance and rehabilitation potential.

3. Unravelling Passive Musculoarticular Stiffness in human.

3.1. Preamble

The EOD is recognized as the most reliable and efficient tool currently available for measuring stiffness. To our knowledge, the EOD is unique in its ability to quantify viscous stiffness, thereby offering the potential to record viscoelastic characteristics accurately. Additionally, using and interpreting EOD measurements does not require extensive experience or advanced programming skills. The EOD has been validated for its accuracy and reliability in measurements. It is adaptable for use on both the wrist and the ankle. However, due to the ankle EOD's limited portability and the primary goal of this thesis to explore PMAS in a large population, this chapter will focus on utilizing EOD to provide comprehensive insights into PMAS in humans.

3.2. How does age and various anthropometric characteristics influence PMAS in a healthy population?

Nguyen, A. P., Herman, B., Mahaudens, P., Everard, G., Libert, T., & Detrembleur, C. (**2020**). Effect of age and body size on the wrist's viscoelasticity in healthy participants from 3 to 90 years old and reliability assessment. *Frontiers in Sports and Active Living*, *2*, 23.

WHAT IS ALREADY KNOWN ON THIS TOPIC:

- Electromechanical Oscillation methodology is valid and reliable to assess PMAS in the ankle.
- Younger and older individuals have demonstrated contrasting values for Passive Musculoarticular Stiffness.
- Passive stiffness seems to be linked to body size characteristics rather than gender.

WHAT THIS STUDY ADDED:

- The current study provides the first reliability measurements of wrist PMAS measured with EOD in a large sample (n = 458) of individuals aged between 3 and 90 years.
- Elastic stiffness is primarily related to body size characteristics, while viscous stiffness is also influenced by aging.
- The current results may serve as normative values for future reference.

HOW THIS STUDY MIGHT AFFECT RESEARCH, PRACTICE OR POLICY:

- Older individuals exhibit less stiffness, particularly in terms of viscous stiffness, which should be considered for prevention and rehabilitation purposes.
- Viscoelastic models utilizing stiffness data obtained from the EOD may be employed to simulate soft tissues in virtual and robotic surgery applications.

3.2.1. Effect of age and body size on the wrist's viscoelasticity in healthy participants from 3 to 90-years-old and reliability assessment.

Abstract

Excessive or insufficient levels of passive musculoarticular stiffness (PMAS) can lead to joint impairment or instability. Quantifying the PMAS may provide a better understanding of neurological or musculoskeletal disorders. Aims of present study were multiple: first, to assess the reliability of quantifying PMAS and to collect normative data on the wrist in healthy participants, and second, to assess the effect of age and body size on PMAS. For this purpose, 458 participants from 3 to 90-year-old were analyzed with an Electromechanical Oscillation Device (EOD). Passive sinusoidal movements were induced in a flexion/extension pattern in the participants' wrists enabling an objective measurement of elastic stiffness (EL) and viscous stiffness (VI). Both the dominant and non-dominant wrists were assessed. Two-way repeated measures ANOVA revealed a sex differentiation from puberty (12-18-year-old) and an increase of EL and VI from childhood to adulthood and a decrease of stiffness at old age. EL and VI values were associated to body size characteristics and age. After body size normalization, EL was no longer influenced by the variables measured. On the other hand, VI remained moderately influenced by age and body size. The current study was able to provide normative data of PMAS in the wrist of healthy participants.

Key terms

Biomechanics, Stiffness, Elasticity, Viscosity, Normative Data.

Introduction

Assessment of passive musculoarticular stiffness (PMAS) is of a growing interest in clinical practice and sport to diagnose, prevent or treat musculoskeletal injuries. Inadequate levels of PMAS have been associated with the occurrence of movement dysfunction, development of pathologies and reduction in performance.

PMAS can be defined as the ratio between the load applied and the deformation that occurs in a structure [68], i.e., the greater the PMAS, the fewer applied load impacts the deformation. PMAS is often differentiated from active musculoarticular stiffness that is generated by contractile structures of the muscles and reflex pathways. [8] For its part, PMAS is due to the rheological properties of non-contractile elements such as muscle-tendon units, aponeurosis and capsuloligamentous complex [61,69]. Furthermore, Riemann et al. described that PMAS included the contribution from all the tissues located around and within the joint, such as muscle-tendon units, skin, subcutaneous tissues, facias, ligaments, joint capsule and cartilage. [70] Those passive components may be influenced by several factors including anatomical properties of muscular tissues, age, gender, body size characteristics but also the surrounding temperature, hour of the assessment, strain rate and amplitude, history of previous pathologies or immobilization. It is known that men develop higher PMAS than women, but PMAS values become nonsignificant between gender after body size normalization. [68] Several authors demonstrated the correlation between body size characteristics and PMAS values, e.g., in the knee or elbow. [71–73] Ditroilo et al. (2012) showed that when the normalized results were compared between older (65 yrs) and younger subjects (22 yrs), no differences were found in PMAS. [74] Kubo et al. (2001) observed that the tendon structures were more compliant in young boys (11 yrs) than those in older boys (15 yrs) and young men (25 yrs). [75] However, no studies described the relationship between all ages of life and PMAS.

PMAS has already been objectively and quantitatively measured distally by inducing sinusoidal oscillations as demonstrated by Rack or Lehmann [13], and adapted by Detrembleur. [8] The distal unit, as ankle or wrist joint, can be simplified in a mechanical model like a torsional spring, a viscous torsional damper and a rotatory mass. These torsional elements seem appropriate as the system is being rotated around the joint. The Electromechanical Oscillation Device (EOD) can measure the resistance of the joint under passive sinusoidal displacements at different frequencies of oscillation. EOD can provide a characteristic torque response, which is dependent on the particular mechanical properties of the model's elements. [8,63] Torsional

spring will correspond to a torque response in phase with the displacement but independent of the frequency while the torsional viscous damper is related to the frequency of the sinusoidal displacement. Torque contribution of the mass can be computed and subtracted from the total response and provides net PMAS composed of elastic stiffness and viscous stiffness. [8] This quantitative device can therefore assess the PMAS of a distal joint such as the ankles, fingers or wrists. [63]

The primary aim of the current study was to investigate the intra- and inter-rater reliability of the EOD in healthy participants. Our secondary aim was to observe the relationship between age and body size characteristics on the wrist's PMAS in a healthy population (in 458 participants from 3 to 90-year-old). To our knowledge, no normalized data of PMAS in the wrist has been acquired thus far. This study could offer such data on PMAS and provide better understanding concerning the assessment and treatment of musculoskeletal disorders or neurological pathologies such as forearm spasticity, tremor in Parkinson disease, wrist arthritis or inflammation.

Material and Methods

Population

The study was performed between July 2018 and December 2018 in Université catholique de Louvain (UCLouvain) in Brussels and Louvain-la-Neuve, Belgium. Healthy participants of both genders aged between 3 to 90 years old were recruited by word of mouth, advertisements in local schools, at the university and surrounding or investigators' acquaintances. Four hundred and fifty-eight participants were enrolled in the current study. Seven were removed due to technical problems during the data collection. Among the 451 remaining participants, 148 children (72 girls and 76 boys) and 303 adults (162 women and 141 men) constituted our sample (see Table1). Childhood was defined by chronological age. No participants reported any pain or inconvenience during or after the data collection. To participate in the study, participant had to meet the inclusion criteria, i.e., no history of injury or pathology in the upper limb, having no pain or any condition that might cause discomfort during the study. The current study was approved by the Local Ethics Committee: "*Comité d'éthique Hospitalo-Facultaire de l'UCLouvain" (CEHF)* (Belgian registration number: B403201523492) in March 2017. All participants and parents of children were informed about the protocol before any measurement and signed a written informed consent.

Bodysize measurements

For all participants, age, height, body mass, forearm length (from the head of the radius to radial styloid process with a ruler), forearm circumference (at the largest part of the forearm) was recorded. Wrist range of motion was measured with a goniometer in full flexion and full extension. All the measurements were taken for both sides while the participants sat with the forearm on the examination table. The dominant side was defined as the writing hand of the participant.

Electromechanical Oscillation Device (EOD)

Inspired by the work of Rack and Lehmann and adapted by Detrembleur et al. in the ankle [8], we developed in collaboration with the Institute of Mechanics, Materials and Civil Engineering of the University of Louvain (UCLouvain), an electromechanical device applying sinusoidal rotatory movements of varying frequencies to objectively assess passive wrists' musculoarticular stiffness. We were inspired by existing devices to build one with as little

inertia as possible, portable, easy to use in a clinical context, safe, reliable and reproducible. The participants' wrists were mobilized from 3 to 12 Hz in 1 Hz increment. [14]The movement was driven by the gripping handle and oscillation movements were performed from 10° of wrist flexion to 10° of wrist extension for a total of mean amplitude of 20° with a precision of 0.8°. Neutral wrist position at 0° of flexion/extension was chosen. [76] Pilot protocol showed no significant difference in starting position in extension or flexion. In addition, if needed, a piece of foam was placed into the opencast to limit the movement of the forearm. A torquemeter (Sensy 62200 of 75 Nm) placed on the gripping handle axis measured torsion forces, while a potentiometer (Codeur SICK absolute 4096 p/t) mounted on the axis of rotation recorded angular displacement. The forearm was fixed by strap in an opencast and the hand was stabilized by four straps around the hand and fingers (Fig. 17).

Participants ideally sat with the forearm relaxed and the elbow flexed between 120°-135° and the shoulder at 45° of flexion. Angle and torque were recorded in 1024 Hz and analyzed by a Fast Fourier Transformation (FFT). Only the first harmonic, corresponding to the gripping handle oscillation frequency was preserved. The amplitude and phase shift of the torque signal relative to the displacement signal were computed. The amplitude of the torque response that is in phase with the wrist displacement is called the elastic torque and the one that is 90° out of phase with the displacement is called the viscous torque. The influence of combined gripping handle and hand inertia (I) on the amplitude of the elastic torque and device friction on the amplitude of the viscous torque were subtracted, respectively, from raw amplitude values. Inertia was calculated as slope of the regression line from total elastic stiffness values *vs* frequency squared. Net elastic stiffness at each frame was obtained by subtracting inertia from total stiffness. The amplitude of the elastic (EL) and viscous (VI) torque components was expressed in terms of stiffness (N m rad^{-1}) and computed as the ratio of net torque amplitude (raw torque amplitude without the effects of inertia or friction) to wrist displacement. EL and VI results were expressed as the mean computed from the three measures recorded at each frequency. Elastic stiffness was represented by the intercept of the regression line fitted through points (Elastic stiffness *vs.* Frequency) and expressed in N m rad^{-1}. Viscous stiffness was represented by the slope of the regression line fitted through points (Viscous stiffness *vs.* Frequency) and expressed in N m s rad^{-1} (Fig. 18).

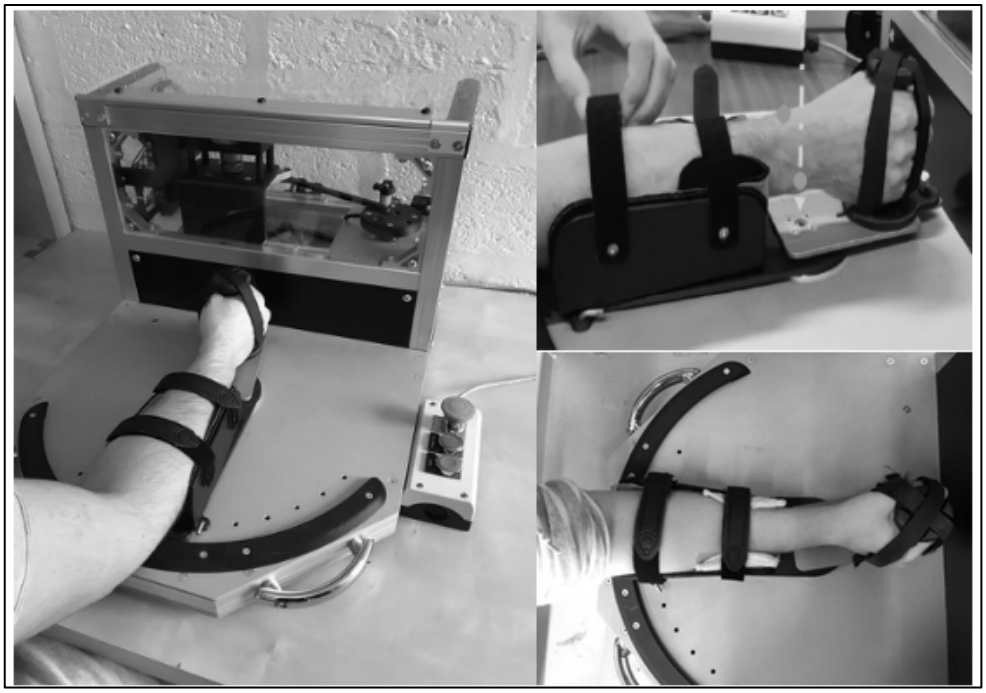


Figure 17 *Electro-mechanical oscillation device adapted by Detrembleur & Herman. Arrow shows the rotatory axe. Dots represent the radial and ulnar styloid process of the forearm.*

Experimental procedure

Reliability study

Intra- and inter-reliability were tested by two operators assessing each participant twice in one session (T_1-T_2-T_3-T_4). All measurements were performed on the dominant limb, defined as the writing hand of the subject.

PMAS study

Participants were assessed twice in one session, one assessment per wrist. Each assessment consisted of 30 trials of 10 different frequencies of oscillation (3 Hz to 12 Hz). Order of frequencies was randomized to avoid pattern prediction. Total duration of one assessment lasted 195 seconds with one minute of rest between each assessment.

Statistical analysis

Reliability study

Data were analyzed using SPSS 16.0. A two-way random intraclass correlation coefficient (ICC 2,1) was used to assess inter-operator reliability and a two-way mixed ICC (3,1) was used to calculate intra-operator reliability. Following the Shrout & Fleiss classification, ICCs > 0.75 were considered as excellent, between 0.4-0.75 as moderate and < 0.4 as weak. [77] Standard errors of the mean (SEM) were calculated using the following Equation 16:

$$SEM = SD_x\sqrt{1 - R_x} \text{ (Equation 16).}$$

where SD represents the standard deviation, x represents the session and R represents the ICC (3,1). The minimal detectable change (MDC) was then calculated as follows (Equation 17):

$$MDC = 1.96\sqrt{2\,SEM} \text{ (Equation 17)}$$

where 1.96 is the two-sided table z value for the 95% CI and √2 is used for the variance of two measurements. [78] SEM and MDC were also expressed as percentages.

PMAS study

Statistical analysis was performed by SigmaPlot version 14 software. Wilcoxon signed rank test (as normality test failed) was used to establish dominance effect on PMAS in all participants.

To eliminate the body size effect, EL and VI values were also normalized following Hof equation (Eq. 3 and Eq. 4) [79].

$$\text{Normalized EL} = \frac{\text{EL}}{\text{Body mass} \times \text{g} \times \frac{\text{Forearm length}}{\text{Height}}} \times \alpha \quad \text{(Equation 18)}$$

where EL represents Elastic stiffness values, *g* represents gravitational acceleration and α the angular acceleration.

$$\text{Normalized VI} = \frac{VI}{Body\ mass \times g \times \frac{Forearm\ length}{Height} \times \omega} \times \alpha^2 \text{ (Equation 19)}$$

where VI represents de Viscous stiffness values, g the gravitational acceleration, ω the angular velocity and α the angular acceleration.

Next, seven groups of ages were determined: 3–6, 6–12, 12–18, 18–25, 25–45,45-65, 65-90-years-old for the following statistical analysis. Descriptive analysis was used on the dominant limb, and results expressed as mean ± standard deviation (SD). Two-way ANOVA was used to establish the effect of age and sex on stiffness.

To characterize the association between age, height, body mass, sex, forearm circumference, forearm length, ROM flexion, ROM extension (X_i variables) and stiffness variables (Y_i), a polynomial regression (first or second order) was adjusted. The choice of order was verified by the highest r and significant p value. The coefficients r were classified as follows: 0-0.25 = very low, 0.26-0.49 = low, 0.5-0.69 = moderate, 0.7-0.89 = high and 0.9-1.00 = very high. The significance level was fixed at p=0.05.

Results

Reliability of EOD

Eighty-six participants (46 female), ranging in age from 5 to 90 years old (29.58 ± 16.8 years) were included in the reliability study. Intra-operator reliability (Table 1) was excellent for all parameters (ICCs ranged from 0.87 to 0.93). Inter-operator reliability showed similar results, with an ICC>0.93 for all stiffness components. The intra-operator SEM values for ES and VI were respectively 0.22 N m rad^{-1} and 0.18 N m s rad^{-1}. Inter-operator SEM values were lower than intra-operator values for ES (0.14 N m rad^{-1}), VI (0.17 N m s rad^{-1}). All the MDC values are shown in Table 1.

Table 1: Reliability of EOD in wrist's stiffness.

Intra examiner			**Reliability**	**Standard Error of Mean**		**Responsiveness**
	MEAN± SD	**ICC (95% CI)**	**SEM**	**SEM%**	**MDC**	**MDC%**
Elastic stiffness (N m rad^{-1})	1.716 ± 0.64	0.878 (0.836-0.91)	0.223	34.9	0.619	36.1
Viscous stiffness (N m s rad^{-1})	1.922 ± 0.68	0.931 (0.907-0.949)	0.181	26.3	0.500	26.0
Inter examiner			**Reliability**	**Standard Error of Mean**		**Responsiveness**
	MEAN ± SD	**ICC (95% CI)**	**SEM**	**SEM%**	**MDC**	**MDC%**
Elastic stiffness (N m rad^{-1})	1.742 ± 0.60	0.941 (0.909-0.961)	0.154	24.3	0.428	25.0
Viscous stiffness (N m s rad^{-1})	1.966 ± 0.67	0.931 (0.895-0.955)	0.178	26.3	0.495	25.9

ICC: intra class correlation coefficient; SEM: Standard error of mean; MDC minimal detectable change

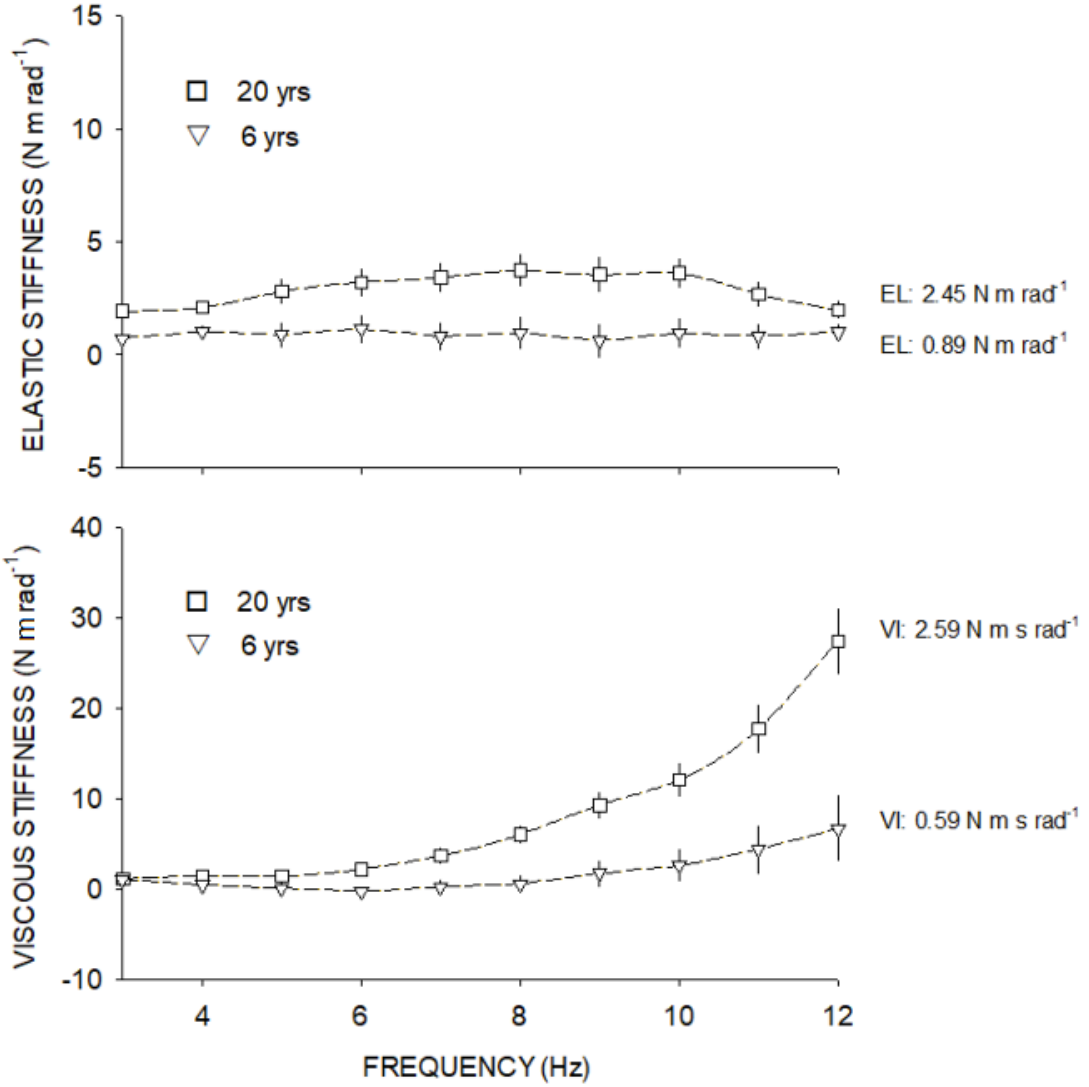


Figure 18. *Elastic stiffness (above) and viscous stiffness (below) at each frequency from 3 Hz to 12 Hz. Square symbols showed the elastic and viscous stiffness from a 20-year-old adult and triangle symbols showed the elastic and viscous stiffness of a 6-year-old child.*

PMAS and wrist dominance

No significant difference was observed between dominant and non-dominant wrist in EL ($p=0.069$; Dominant (Median(1st quartile-3rd quartile)) = 1.33 (0.85-1.96) N m rad^{-1}; Non-dominant = 1.30 (0.84-1.89) N m rad^{-1}) and VI ($p=0.47$;Dominant (Median (1st quartile-3rd quartile)) = 1.57 (0.77-2.23) N m s rad^{-1}; Non-dominant = 1.59 (0.90-2.11) N m s rad^{-1}). As dominance showed no significant interest, further analysis was only made on dominant side.

PMAS and sex difference

Two-way ANOVA showed a significant difference between sex and age in both EL and VI components. Sex difference only appeared after the 12-18-year-old group while 3-6-year-olds ($p=0.893$) and 6-12-year-olds ($p=0.951$) showed no significant differences between males and females concerning EL and VI values. The 12-18, 18-25, 25-45, 45-65 and 65-90-year-old group showed higher EL and VI values in males than females (see Table 2).

Table 2. Descriptive statistics on body size variables and stiffness

Age group (years)	Sex	n	Height (m)	Body mass (kg)	Forearm circumf. (m)	Forearm length (m)	ROM Flex (°)	ROM Ext (°)	EL (N.m Rad^{-1})	VI (N.m.s Rad^{-1})
3-6	F	16	1.06 ±0.08	18.05 ±3.45	0.17 ±0.01	0.16 ±0.01	107.63 ±6.81	93.81 ±5.31	0.41 ±0.13	0.13 ±0.14
	M	23	1.07 ±0.08	17.49 ±3.35	0.18 ±0.01	0.16 ±0.02	108.3 ±12.41	92.44 ±6.42	0.45 ±0.23	0.16 ±0.15
6-12	F	22	1.37 ±0.12	32.48 ±7.49	0.2 ±0.02	0.21 ±0.02	107.59 ±10.9	90.73 ±4.65	0.69 ±0.17	0.67 ±0.27
	M	14	1.38 ±0.12	33.19 ±7.48	0.21 ±0.02	0.22 ±0.02	102.5 ±9.35	90.43 ±11.09	0.71 ±0.25	0.66 ±0.35
12-18*	F	34	1.65 ±0.07	53.53 ±7.53	0.25 ±0.01	0.27 ±0.02	87.06 ±3.22	83.29 ±5.39	1.43 ±0.33	1.7 ±0.39
	M	39	1.72 ±0.09	58.6 ±11.67	0.26 ±0.02	0.29 ±0.02	86.08 ±4.72	80.08 ±8.35	1.75 ±0.48	1.99 ±0.57

Table 2 (continued)

18-25*	F	44	1.68 ±0.06	61.85 ±9.34	0.24 ±0.02	0.25 ±0.02	88.07 ±7.82	67.25 ±12.17	1.4 ±0.41	1.64 ±0.45
	M	50	1.81 ±0.06	75.19 ±10.98	0.28 ±0.02	0.28 ±0.02	89.16 ±9.83	72 ±9.39	2.07 ±0.35	2.41 ±0.41
25-45*	F	37	1.68 ±0.06	66.24 ±12.10	0.26 ±0.02	0.23 ±0.02	89.89 ±9.87	68.27 ±15.03	1.64 ±0.49	1.89 ±0.45
	M	41	1.80 ±0.07	79.36 ±11.76	0.28 ±0.02	0.26 ±0.03	85.29 ±9.59	68.19 ±12.99	2.22 ±0.44	2.56 ±0.43
45-65*	F	38	1.68 ±0.07	72.01 ±13.58	0.25 ±0.02	0.26 ±0.01	93.37 ±11.26	87.76 ±16.26	1.69 ±0.43	1.92 ±0.49
	M	33	1.8 ±0.07	87.94 ±11.02	0.29 ±0.02	0.29 ±0.02	89.24 ±14.74	85.49 ±13.07	2.25 ±0.51	2.45 ±0.51
65-90*	F	44	1.61 ±0.06	61.61 ±11.67	0.24 ±0.02	0.25 ±0.02	63.29 ±10.28	65.46 ±11.19	1.09 ±0.43	0.97 ±0.43
	M	18	1.69 ±0.07	70.17 ±13.82	0.26 ±0.02	0.26 ±0.02	72.33 ±14.16	63.89 ±12.43	1.65 ±0.83	1.59 ±0.84

Results are provided in (mean ±SD). Circumf: circumference; ROM: Range of Motion; Flex: flexion; Ext: Extension; EL: Elastic stiffness; VI: Viscous stiffness; F: Female; M: Male; Na: missing values. * indicated significant difference between male and female for EL and VI values.

PMAS and aging

Results showed an increase of EL stiffness during the lifetime with similar strategy between males and females with the male group increasing further after 12-18-year-old group and both male and female values dropping after 65-year-old (18). Normalized EL values for both male and female showed an equalizing of values from the 3-6-year-old group to the 65-90-year-old group (Figure 18).

VI stiffness showed similar pattern than EL with an increase of values from childhood to adulthood and a decrease of values after the 45-65 years old group. However, normalized VI values kept the same evolution after removing the height, body mass and forearm length influences. Also, normalized VI values didn't show any sex differences ($p=0.056$). All age group relationships are illustrated in Figure 19.

PMAS and body size characteristics

Polynomial regressions (first or second order) were adjusted between normalised and not normalised stiffness and body size characteristics (Table 3).

Table 3: Polynomial regression (first order: $Y_i=Y_0 + b_0.X_i$ or second order: $Y_i = Y_0 + b_0 . X_i + b_1 . X_i ^2$) on elastic and viscous stiffness variables (Y_i) (not and normalized). X_i are independent variables characterizing body size effect and age.

EL	Order	Equation	r^2	r
Age (years)	2	0.445 + (0.0688 * Age) - (0.000706 * Age^2)	0.42	0.65
Sex (1=woman)	1	0.82 + (0.479 * Sex)	0.12	0.35
Height (m)	1	-2.407 + (2.411 * Height)	0.57	0.75
Body mass (kg)	1	0.0335 + (0.0244 * Body mass)	0.57	0.75
Forearm circumference (m)	1	-2.030 + (14.273 * circumference)	0.58	0.76
Forearm length	1	-1.390 + (11.672 * Forearm length)	0.46	0.68
ROM flexion (°)	2	-3.546 + (0.130 * ROM flex) - (0.000794 * ROM flex^2)	0.19	0.44
ROM extension (°)	1	2.183 - (0.00848 * ROM ext)	0.04	0.20

Table 3 (continued)

VI	Order	Equation	r^2	r
Age (years)	2	0.324 + (0.0888 * Age) - (0.000938 * Age^2)	0.47	0.69
Sex (1=woman)	1	0.87 + (0.536 * Sex)	0.09	0.30
Height (m)	1	-3.513 + (3.181 * Height)	0.71	0.84
Body mass (kg)	1	-0.189 + (0.0304 * Body mass)	0.59	0.77
Forearm circumference (m)	1	-2.847 + (18.143 * circumference)	0.63	0.79
Forearm length	1	-2.136 + (15.249 * Forearm length)	0.53	0.73
ROM flexion (°)	2	-5.509 + (0.180 * ROM flex) - (0.00109 * ROM flex^2)	0.19	0.44
ROM extension (°)	1	2.507 - (0.0108 * ROM ext)	0.03	0.17

Table 3 (continued)

Normalized EL	Order	Equation	r^2	r
Age (years)	2	0.0150 + (0.000168 * Age) - (0.00000232 * Age^2)	0.08	0.28
Sex (1=woman)	1	0.013 + (0.002 * Sex)	0.05	0.22
Forearm circumference (m)	1	0.0126 + (0.0161 * Forearm circumference)	0.01	0.10
ROM flexion (°)	1	0.0136 + (0.0000335 * ROM flex)	0.01	0.10
ROM extension (°)	1	0.0171 - (0.00000721 * ROM ext)	0.00	0.00
Normalized VI	**Order**	**Equation**	**r^2**	**r**
Age (years)	2	0.00132 + (0.0000744 * Age) - (0.000000853 * Age^2)	0.47	0.69
Sex (1=woman)	1	0.002 + (0.0003 * Sex)	0.03	0.17
Forearm circumference (m)	1	-0.000749 + (0.0124 * Forearm circumference)	0.24	0.49
ROM flexion (°)	2	-0.006 + (0.0002 * ROM Fl) - (0.000001 * ROM fl^2)	0.19	0.44
ROM extension (°)	1	0.00324 - (0.0000118 * ROM ext)	0.03	0.17

EL=elastic stiffness – VI=viscous stiffness – ROM= range of motion; Bold= moderation association; Underline=good association; r^2= determination coefficient; r= correlation coefficient.

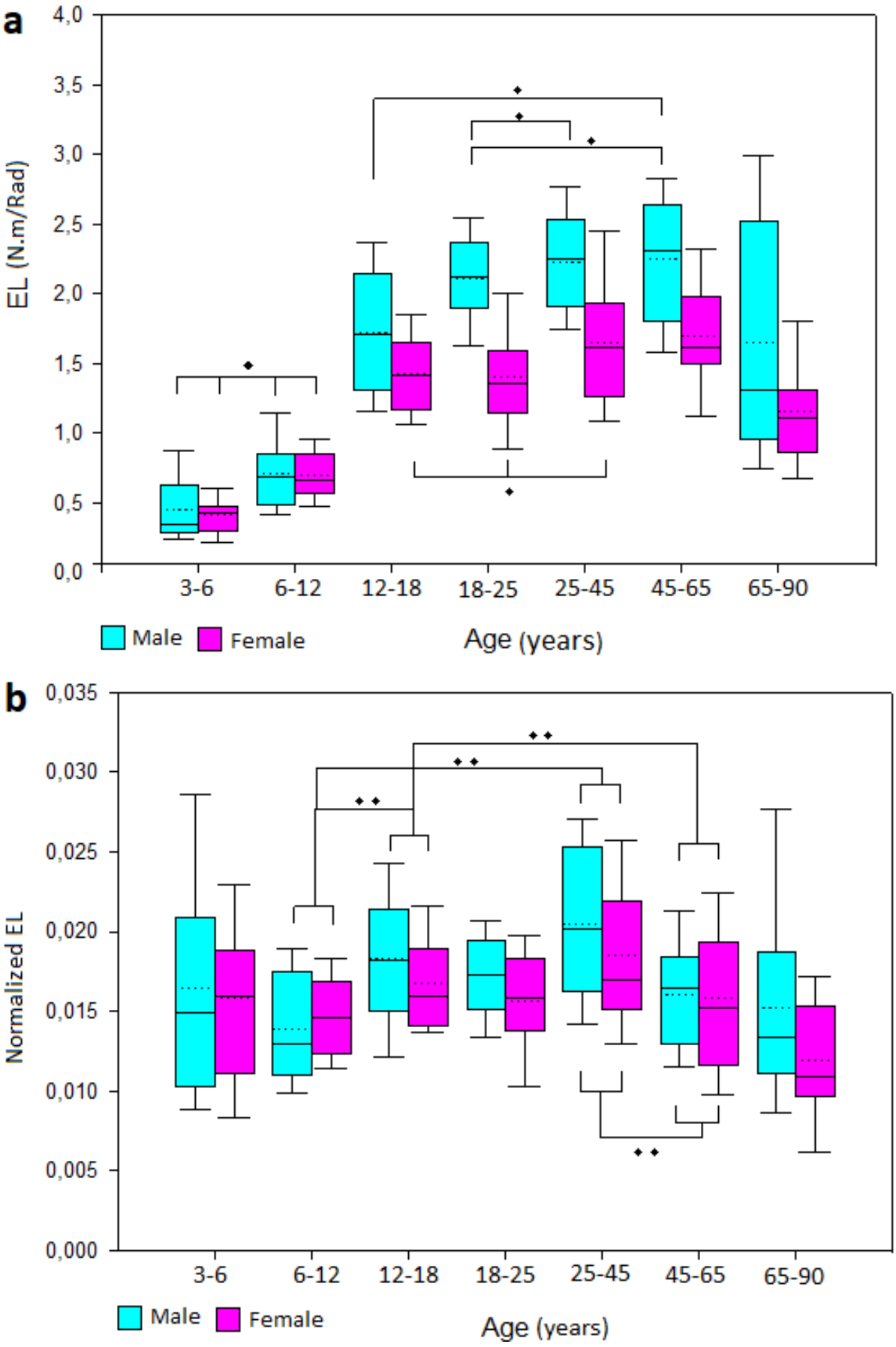
EL is highly associated to height, body mass and forearm circumference. It is moderately correlated to age, forearm length and ROM flexion. VI is highly associated to height, body mass forearm circumference and forearm length. It is moderately associated to age and ROM flexion. After normalization, EL is no longer influenced by age nor body size. However, normalized VI stays moderately associated to age, forearm circumference and ROM flexion (Fig. 18).

Discussion

Our results showed normative data of PMAS in 451 healthy participants ranged from 3 to 90-years-old. EOD showed excellent intra- and inter-operator reliability in a healthy population between 5- and 90-years-old. Our results also suggest that body size characteristics such as height, body mass and forearm circumference are the most influencing factors for EL and VI components. After eliminating the effect of body size, we observed that aging seems to impact VI rather than EL.

Reliability

Our results showed excellent intra- and inter-operator reliability of the EOD in healthy participants of 5 to 90 years old. Lobet et al showed comparable results in their study on the ankle, with good-to-excellent intra- and inter-day reliability in PMAS of the ankle in children, adolescents and adults with hemophilia using the same assessment protocol. [64] Reliability study showed that EOD might not be as accurate with the standard error of the mean (SEM) around 10% and minimal detectable change (MDC) of 25%. This might be related to excessive forearm movement into the cast during the acquisitions at high frequencies or non-linear soft tissue behavior at high frequencies. This should continue to be investigated.


a
4,0
3,5
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EL (N.m/Rad)
3-6
6-12
12-18
18-25
25-45
45-65
65-90
Male
Female
Age (years)
b
0,035
0,030
0,025
0,020
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Normalized EL
3-6
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12-18
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25-45
45-65
65-90
Male
Female
Age (years)

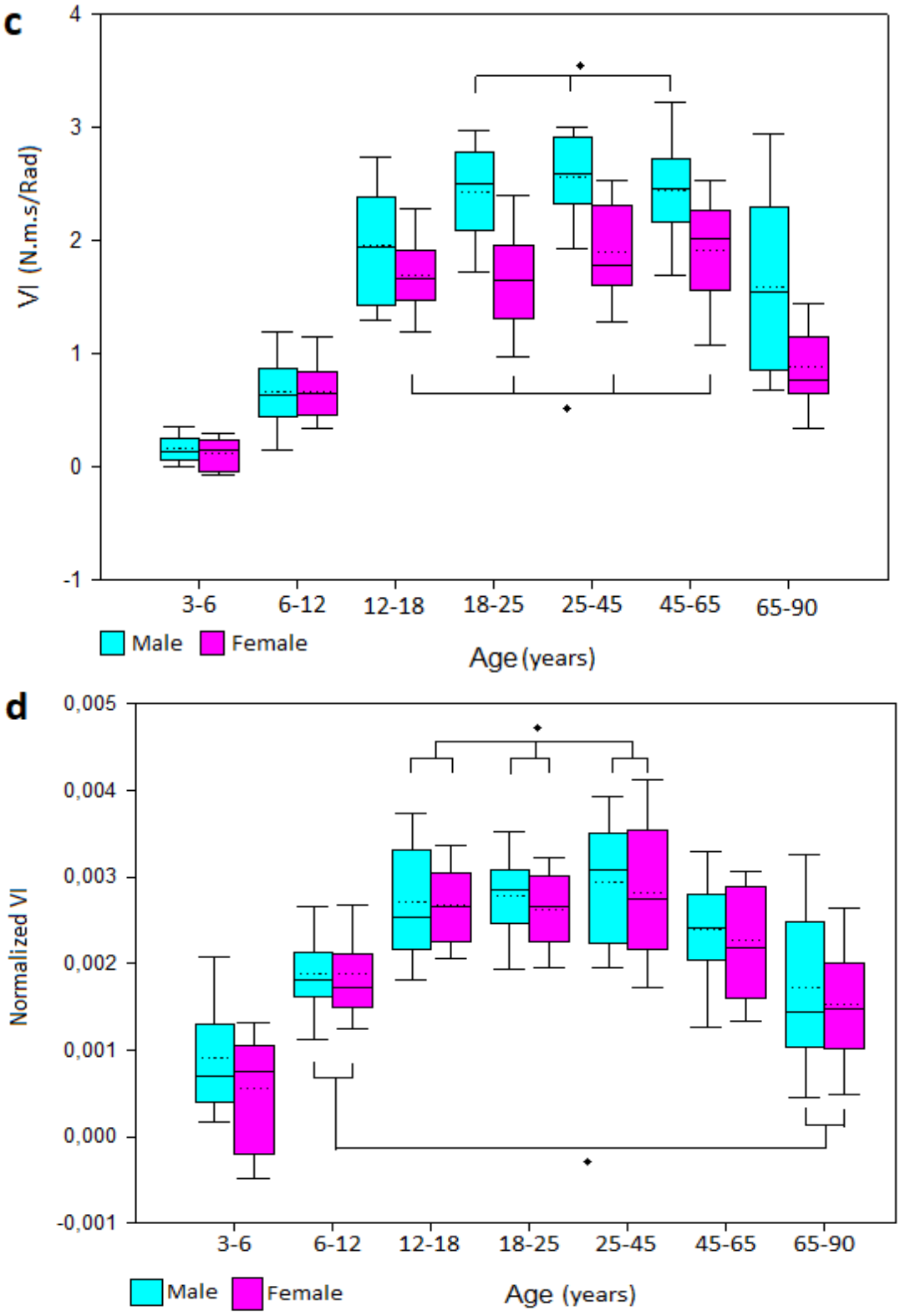

Figure 19. *Boxplot of Male and Female a) Elastic stiffness (EL) and c) Viscous stiffness (VI) values and Normalised b) EL and d) VI values by group of age. * : no significant differences between groups, **: significant differences between group.*

PMAS and dominance

Our study reported no difference in PMAS between the dominant and non-dominant wrist. Several publications showed higher passive stiffness in the dominant upper limb. [80–82] Durand et al. recently described the influence of handedness in passive wrist stiffness suggesting that differences between studies were probably caused by sample size and age of participants. However, EOD only assessed the flexion/extension axis while Durand et al. assessed flexion/extension and ulnar/radial axis. [82] Moreover, the velocity of our assessment was greater than in previously cited study. Dominant and non-dominant limb could display different properties following their daily activities. Some authors also described the relationship between the development of intramuscular fibers, i.e., Type I or type II, and the intensity of physical activities. They explained that repetitive movement in one limb induced an oxidative adaptation of type I fibers without modifying their volume while loaded activities induced the hypertrophy of both the type I and II. [83] Therefore, sports or professional activities could have an influence on the morphological properties of the participants. More accurate assessment of specific asymmetric populations -i.e., tennis players, musicians, carpenters, laborers, etc.- may show different results. No significant PMAS difference was observed for the dominant and non-dominant wrist. But adults might demonstrate asymmetrical PMAS between wrists with dominant PMAS higher than non-dominant wrist.

PMAS and sex

First, children demonstrated significantly weaker PMAS than adolescents or adults. Furthermore, men showed a significantly higher PMAS than women. These results are in concordance with the literature. [58,68,84] Bodysize characteristic does not differ between boys and girls before puberty. Sexual dimorphisms are induced by sex steroid hormones during puberty. [85] In accordance with this theory, sex differences were apparent towards the 12-18-year-old group in current study. Previous research demonstrated that male participants displayed more stiffness than female participants and it could be due to a greater height and body mass. [68,72,82] Another study pointed out that muscular fiber's diameter did not differ in boys and girls before the age of 15. [86] We showed that height, body mass or forearm circumference were mostly associated to PMAS components. Similar results were found by Decostre et al. concerning the wrist strength. Indeed, height and forearm circumference

influenced the wrist strength in children while forearm circumference, age and sex were predictable factors in adult population. [87] Concomitantly, another study confirmed with a hand-held dynamometer that sex, body mass and age might be predictive factors for wrist extensors strength. [88]

PMAS and aging

PMAS is the result of muscle rheology (viscoelasticity) but also articular structures such as the joint capsule, ligaments, skin and fat tissues. It seems thus impossible to isolate the impact of age from the many other factors that vary through the life of a subject. [89] While height and body mass are associated to both viscoelastic elements, after eliminating the size effect, age and forearm circumference only tend to explain VI not EL. Indeed, forearm circumference's impact in PMAS may be explained by the number of intramuscular crossbridge. [58,73,90] It is difficult to analyze the isolated impact of one structure in the joint. Diminution of fiber dimensions, intramuscular fat deposit or insulin resistance are several factors that must be considered in aging and PMAS. Such degenerations could affect the structural organization of the surrounding tissues around the wrist. [91] Skin aging can also play a role in PMAS as intrinsic and extrinsic factors induce skin modifications. Several hypotheses, such as oxidative stress, DNA damage, telomere shortening or inflamaging, could influence skin aging process. [92] Distribution of collagen, elastin or proteoglycans in the extracellular matrix can affect the tensile strength, elasticity and hydration of the aged skin. [93] Wilke et al. described an age-related modification in fascia thickness as older participants showed higher thickness in the lumbar spine than younger participants. [94] Lastly, Intramuscular or biological properties of subcutaneous structures may also influence viscoelasticity characteristics of older participants.

There are limitations in our study that should be considered.

(1) We did not record muscle activity during acquisitions. EMG could have been used to make sure that the muscle was inactive. [95] However, active muscular activity can be seen by the examiner during data analysis although it is not sure that every patient stood in the standardized position and that children were as compliant as our adult participant.

Participants that provided outliers patterns due to muscular activity were excluded (n=7).

(2) Children were classified by chronological age and not by maturation rate. As it is known, growth rate differs from boys and girls. This could have biased our results, "childhood" must therefore be defined with cautions.

(3) PMAS were assessed in both flexion/extension muscles groups, providing a general non-specific PMAS for the wrist. However, PMAS of the wrist is mostly provided by fingers and thumb muscles. [96]

(4) Analysis of the torque and displacement signals were based on the Hill's three element muscle model. [6] As the movement provided was of low amplitude, we assume that the signals have linear behaviour and therefore applied a linear regression model. However, behaviour of the wrist complex is not perfectly linear (see fig. 6) and current results may be overestimated.

(5) Sports activities were not recorded (lack of answers, evasive answers) and professional activities were omitted. Such activities could have impacted the development of PMAS and should have been considered.

Perspectives

Developing further clinical trials might provide better guidelines concerning neurological disorders as spasticity. PMAS could also play a major role in assessment of many musculoskeletal dysfunctions. [97] Quantifying diminished PMAS could be relevant as PMAS may influence proprioceptive acuity. [98] EOD methodology was already established in the lower limb. Detrembleur et al. demonstrated a significant decrease in PMAS in spastic patient before and after baclofen injections. [8]

In surgical practice, EOD might assist the clinician for example, in tendon lengthening or to measure tremor reduction by means of a deep brain stimulation in Parkinson disease. [99–101] Moreover, EOD could be an interesting post-treatment tool to objectify the success or failure of the latter. [102] The viscoelastic models based on stiffness data extracted from the EOD might be considered in modeling soft tissues for virtual and robotic surgery. [103]

In conclusion, EOD permits quantifying PMAS reliably and reproducibly. EL and VI stiffness are mostly explained by body size characteristics, like height, body mass and forearm circumference. Age variable is moderately associate to stiffness. After body size normalization, EL was no longer influenced by the variables measured. On the other hand, VI remained moderately influenced by age and body size. EOD provides a normative data in a large sample of healthy participants that can be used as a reference for pathological disorders in neurological or musculoskeletal fields, or in surgery.

3.3. Does EOD technology truly measure passive stiffness?

Libert, Thibaut, Christine Detrembleur, Francois Melebeck, and **Anh Phong Nguyen**. Validating the measurement of passive Musculo-articular wrist stiffness without intentional or reactive contraction using axillary plexus block. *Clinical Biomechanics (2024): 106190.*

WHAT IS ALREADY KNOWN ON THIS TOPIC:

- Electromechanical Oscillation methods have been demonstrated to be reliable in assessing the wrist in healthy subjects.
- EOD reported passive stiffness, but its in vivo validation has not been established.

WHAT THIS STUDY ADDS:

- EOD reported valid measurements in both the tested and control conditions.
- EOD can be utilized in clinical practice settings.
- EOD accurately reported Passive Stiffness values, as confirmed by the axillary block.

HOW THIS STUDY MIGHT AFFECT RESEARCH, PRACTICE OR POLICY:

- The clinical application of EOD could improve guidelines for treatments and rehabilitation in neurological or orthopedic contexts.
- This was the first instance of measuring both elastic and viscous stiffness in pathological conditions.
- PMAS is both valid and reliable for use in healthy and pathological individuals.

3.3.1. Validating the measurement of passive Musculo-articular wrist stiffness without intentional or reactive contraction using axillary plexus block.

ABSTRACT

BACKGROUND: The term passive musculo-articular stiffness describes how easily a joint may move passively. To accurately measure wrist stiffness, an electro-oscillation device was developed. The objectives were to 1) ensuring that the measurement are free from intentional or reflex contraction, 2) analyzing how forearm anatomy affects the passive stiffness of the wrist and 3) determining the clinical practical relevance of the device.

METHODS: In this prospective study, the device generated low amplitude sinusoidal motions in flexion and extension on the wrist to quantify elastic and viscous passive stiffness in voluntary orthopaedic patients. The first series of measurements was carried out in the state of voluntary relaxation, the second series of measurements was carried out after an axillary plexus anesthetic block. A matched group of healthy subjects were use for control.

FINDINGS: The EOD methods effectively enable the measurement of passive joint stiffness, since the stiffness values obtained show no statistically significant difference pre-post the anesthesia. The stiffness values are comparable to those of healthy subjects. The effect of forearm passive structure, estimated by the perimeter of the forearm, influences the passive stiffness of the wrist, mainly the viscous component.

INTERPRETATION: The use of sinusoidal oscillation was well accepted by the participants, demonstrating its usefulness and applicability in a clinical setting. This work serves as a foundation for future investigations of orthopaedic and/or neurological pathological conditions characterized by abnormal passive joint stiffness of the wrist. It paves the way for its use as a diagnostic, prognostic, and monitoring tool in these pathologies.

Keywords: Biomechanics, wrist, stiffness, clinical study.

INTRODUCTION

The wrist can be described as "the most complex and least understood joint in the body" [104]. This joint plays a major role in our interactions with the world, in our daily, creative and sporting activities. Among its characteristics, stiffness is an essential component: it is essential for postural maintenance and potentially penalizing in the event of disturbed intensity. [105] Passive Musculo-articular stiffness (PMAS) refers to the joint's resistance to passive movement [106]. This stiffness is associated with the rheological properties of all intra- and periarticular tissues. [105] It becomes apparent during small-amplitude displacements from the neutral resting position, primarily attributed to the movement of muscle tissue. [107] The evaluation of joint stiffness is crucial for the diagnosis, prognosis and monitoring the treatment of pathological conditions marked by changes in stiffness. [105] Stiffness assessment methods can be categorized into two groups: The first group consists of qualitative tests, such as the Ashworth scale, the Tardieu test, [108] and the pendulum test. [109]These procedures provide results that are deficient in both objectivity and reproducibility [82,105,110]. The second category comprises quantitative assessments, such as the examination of stretch reflexes combined with electromyographic measurements [111]. However, this technique exhibits limited performance reproducibility and high inter-subject variability [8,105]. Other quantitative assessment methods involve ultrasonography or shear-wave elastography. However, these systems are not only costly but also demand an extensive learning curve [13,14]. The absence of reliable, reproducible methods for quantitative assessment of passive stiffness gave rise to the pioneering work of Rack, subsequently continued by Lehmann. Their method involves calculating joint stiffness by analysing the joint's response to sinusoidal motion. [13,14] The utilization of low-amplitude sinusoidal movements at various displacement speeds has the advantage distinguishing between the elastic and viscous components of total stiffness. Taking this into account, Nguyen et al. focused on passive wrist stiffness in healthy subjects aged from 3 to 90 years. These authors evaluated the reproducibility of stiffness measures using a novel device developed by the Institute of Mechanics, Materials and Civil Engineering at UCLouvain (BELGIUM), namely an electro-oscillation device (EOD). [110] This investigation into passive stiffness through sinusoidal motion has prompted several questions. Is the measured stiffness genuinely passive, or is it influenced by involuntary or reflex muscular contractions? How can we assess and control the presence or absence of "parasitic" muscle contractions? Past work has addressed this problem by monitoring relaxation via electromyographic recording. [76] Another limitation of previous studies is that they were conducted in a laboratory setting, which could pose

challenges in implementing these measurement methods with patients in routine clinical settings [82,110]. It is relevant, therefore, to assess the applicability of the EOD measurement methodology in routine clinical practice to gauge its suitability within these clinical conditions.

The primary objective was to validate the measurement of passive musculo-articular stiffness of the wrist without deliberate or reactive contraction. Subsequently, we aimed to assess the suitable of EOD for routine clinical use and examine the impact of body size and anthropometric factors on passive wrist stiffness.

METHODS

This prospective interventional study was conducted in September 2019 at the *Centre Hospitalier Régional de Namur* (Belgium) following approval by the local ethics committee in early September 2019.

Population

The study participants were recruited from the elective operative program of four hand surgeons over a 2-week period. Volunteers were required to be of adult age of any gender and to sign an informed consent form prior to study participation. The selected surgeries included: finger cures, carpal tunnel release, arthro-synovial cyst exeresis and minor finger surgery. These procedures were chosen under the assumption that they would have a minimal impact on stiffness due to their low impact on muscle tissue.

The study excluded individuals with following conditions: previous hand surgery up to the elbow, conditions affecting the upper limb studied (other than those described above), prior wrist sprains, and previous rheumatic diseases. A control group of healthy subjects was selected from the study by Nguyen et al. [110]. This cohort included healthy subjects with similar characteristics in terms of age, gender, body mass and height - factors known to influence the wrist PMAS. [110]

Twenty-eight eligible patients comprising 12 men and 16 women met the inclusion criteria and were all enrolled in the study (Age: 52.6 ± 15.4 years, Height: 1.66 [1.6-1.69] m, Weight: 74 [65-89] kg). A control group of forty-six healthy subjects (Age: 57.1 ± 11.81 years, Height: 1.66 [1.62-1.70] m, Weight: 70 [62.3-85.5] kg) were drawn from the study conducted by Nguyen et al, (2020). (which included 451 subjects). This control group was selected to match the gender, age, height and weight characteristics of the 28 participants in the study.

Material

Electromechanical oscillation device (EOD)

The device translated the rotary movement generated by the motor into an oscillating movement applied to the wrist. (Fig.20) This motion has an amplitude of 20° of flexion-

extension (+/- 10° from the neutral position) at frequencies of 3 to 12 Hz (in increments of 1 Hz) in a randomised order. The forearm was securely immobilized in a splint, and the hand was firmly fixed to a handle using a Velpeau bandage. Angular displacement was measured using an optical encoder (Sick Absolute 4096 encoder), and the opposition of the wrist to the movement were recorded using a torquemeter (Sensy 62200). Recordings are taken at a frequency of 1024 Hz. The effects of system inertia and friction were subtracted. The data were analysed by Fourier transformation, using software developed in-house to isolate viscous and elastic components from the total stiffness.

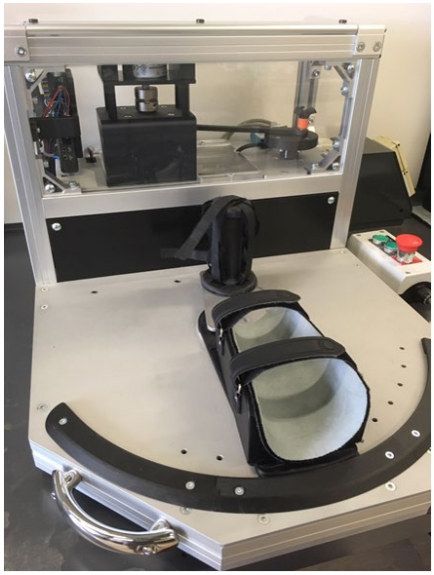

Figure 20. *Electromechanical Oscillation Device for the wrist.*

Outcomes

For each participant, the following data were collected: age, gender, dominant hand (for writing), height, weight, occupation, and sports played. The PMAS was assessed at rest both before and after anesthetic plexus of limb using the EOD. Additionally, the following measurements were recorded:

- forearm circumference (at widest point) in cm

- forearm length from head to styloid process of radius (tape measure) in cm

- maximum wrist flexion-extension mobility (goniometer) in degrees

- Skin fold (radial edge, opposite extensors at widest part of forearm) via Fat Track II® (AccuFitness - Denver, USA) in cm

- Maximum grip strength (3 repetitions) using Jamar® hydraulic hand dynamometer (Sammons Preston Rolyan - Bolingbrook - USA) in N

- Beighton score (generalized joint hyperlaxity evaluated on a scale of 9 points)

Procedure

Axillary plexus anaesthetic block

The plexus anaesthesia was performed by two anaesthetists using the same methodology, ultrasound-guided axillary plexus anaesthesia (SonoSite SII Ultrasound System). In practice, the subject's shoulder was carried in abduction and external rotation. The needle insertion site was locally anaesthetised with 1-2 mL Linisol® 2% (lidocaine hydrochloride 20mg/mL). Under ultrasound control, a Stimuplex Ultra 360 needle was inserted and a total of 40 mL Scandicaïne® 1% (mepivacaine hydrochloride 10mg/mL) mixed with 0.05 mL Sterop® Adrenaline (adrenaline tartrate 1 mg/mL) was injected. The injections were performed in the following order: 5 mL in the musculocutaneous nerve, 10 mL in the radial nerve, 10 mL in the ulnar nerve, 10 mL in the median nerve. The remaining 5 mL were injected subcutaneously to target the medial cutaneous nerves of the upper arm and forearm. This anaesthetic technique allows all myotomas of the forearm and hand to be targeted.

PMAS measurement

The same operator conducted all the measurements, under consistent set-up. The participant was positioned in a semi-seated position (i.e. with the chest at approximately 45° to the horizontal) on a height-adjustable stretcher, with the upper limb externally rotated by 70° and the elbow flexed by 45° to place the forearm in an adjustable groove. Pro-supination was neutral, and the ulnar styloid was positioned at the centre of rotation of the handle. The hand was firmly secured. The participant was instructed not to grasp the handle and to maintain a state of maximum relaxation during the procedure [76]. Initially, the subject underwent to 2 trials of 3 and 12 Hz (minimum and maximum values) to assess the comfort of the set-up and prevent any surprise that might lead to involuntary reflex or anticipatory muscle contraction. This was followed by a series of 10 measurements at random frequencies from 3 to 12 Hz. Subsequently,

the subject underwent axillary plexus anaesthesia. A minimum time of 20 minutes was observed with confirmation of anesthesia verified by the complete disappearance of the sensation of cold (ether) and touch, and the complete paralysis of the muscle groups dependent on the median, ulnar, and radial nerves, comparable to the peak action of lidocaine (Scandicaïne®). [112] Finally, a second series of 10 measurements was taken under the same set-up conditions.

Pilot test for clinical implementation

The pilot test was intended to study the feasibility of clinical implementation of viscosity-elasticity measurement and associated assessments. The pilot test can be seen as a trial period for a new implementation to highlight the potential incentives and disincentives for wider use. The following data were collected:

- Acceptability by the participant

- Tolerance by the participant

- Time constraints, in terms of delays in the operating schedule

- Logistical constraints, in terms of the use, handling and clutter of the EOD.

Statistical analysis

A Shapiro-Wilk test was used to assess the normality of the data. If normality was confirmed, a paired t-test was used to assess the differences between the pre-post values. If normality failed, a Wilcoxon test was used to compare the median of the two groups (before vs after surgery). To compare the group of patients with the group of healthy subjects, we used independent samples. For normal data, t-student test was performed. If the data were not normal, a Mann-Whitney test was used.

Backward stepwise multiple linear regression was performed to identify the impact of the various independent factors on viscous and elastic stiffness. The coefficient of determination R^2 shows how well the regression model explains the distribution of points: it increases when a factor improves the model.

The data were presented as the mean (standard deviation) or as the median [1st - 3rd quartile]. A p-value < 0.05 was considered statistically significant and all analysis were performed using Medcalc software (version. 22.013).

RESULTS

The patient group (n=28) and the control group (n=46) were compared. The groups did not show statistically differences in outcomes, including forearm length, forearm circumference, elastic stiffness, and viscous stiffness, as indicated by p-values > 0.05 (Table 4).

Table 4. baseline values of participants compared to matched control

	Control (N=46)	**Patient (N=28)**	**p-value**
Genre (Female/Male) (%)	69.6 / 30.4	57.2/42.8	0.40
Age (years)	57.1 ± 11.81	52.6 ± 15.43	0.16
Height (cm)	1.66 [1.62-1.70]	1.66 [1.6-1.69]	0.51
Weight (kg)	70 [62.3-85.5]	74 [65-89]	0.23
Forearm length (cm)	0.25 [0.243-0.27]	0.253 [0.245-0.265]	0.59
Forearm circ. (cm)	0.264 ± 0.022	0.274 ± 0.027	0.07
EL (N.m.rad^{-1})	1.76 [1.14-2.32]	1.83 [1.47-2.29]	0.21
VI (N.m.s. rad^{-1})	1.86 [1.04-2.42]	1.68 [1.31-2.59]	0.49

Shapiro-Wilk test was used to assessed normative values. If the normality was confirmed, a paired t-test was performed to compare the mean of patient and matched control. Normative data were presented as mean ± standard deviation. If the normality was rejected, a Wilcoxon test was performed to compare the median of the two groups. Data were then presented as median [1st – 3rd quartile]. *Abbreviations: SD= standard deviation, circ.= circumduction, ROM= range of motion, EL= elastic stiffness, VI= viscous stiffness.*

Changes in passive stiffness values before and after intervention

The paired t-test showed no significant difference (Table 5) between pre- and post-intervention for either elastic or viscous stiffness values (Fig 9 – general introduction).

Table 5. Paired t-test on pre- and post-intervention values

	Pre-intervention	Post-intervention	Difference	95%CI	p-value
Elastic Stiffness (N.m.rad^{-1})	1.96 ± 0,61	2.01 ± 0,62	0.04	[-0,14 to 0,22]	0.6515
Viscous Stiffness (N.m.s. rad^{-1})	1.90 ± 0.68	1.99 ± 0.75	0.085	[-0,05 to 0,22]	0.2091

Both elastic and viscous stiffness values demonstrated normative values assessed by Shapiro-wilk test. The data are presented with mean ± standard deviation.

Backward stepwise regression of stiffness measurements

Backward stepwise regression analysis revealed that the passive elastic stiffness values were partially explained ($R^2 = 0.639$) by age, height, forearm circumference, and extension range of motion (Table 6). Other parameters, including weight (p-value=0.360), Beighton score (p-value=0.239), grip strength (p-value=0.546), forearm length 210 (p-value=0.641), adapted forearm circumference (p-value=0.670), skinfold thickness (p-value=0.967) and gender (p-value=0.456) were not included in the model.

Table 6. Backward stepwise regression

	Coefficient	Standard error	p-value	Coefficient of determination (R^2)	Adjusted R^2
Elastic stiffness				0,639	0,577
Constant	-5,406	-1.616			
Age (years)	0.0183	0,005	0,002		
Height (m)	0,0272	0,0125	0,009		
Forearm circumference (cm)	0,099	0.009	0,004		
Extension ROM (°)	-0,0162	0,007	0,033		
Viscous stiffness				0,812	0,789
Constant	-5,447	0.874			
Forearm length (cm)	0,200	0,0377	<0.001		
Forearm circumference (cm)	0.102	0.0269	<0.001		
Extension ROM (°)	-0.0114	0.0054	0.047		

In terms of passive viscous stiffness values, the results of stepwise backward regression indicated that forearm length and circumference as well as extension mobility were the determining factors influencing viscous stiffness ($R^2 = 0.812$) (Table 6). None of the following variables - age (p-value=0.280), height (p-value=0.338), weight (p-value=0.920), Beighton score (p-value=0.617), grip strength (p-value=0.484), adapted forearm circumference (p-value=0.627), skinfold thickness (p-value=0.910) nor gender (p-value=0.359) – were statistically significant, were not considered in the regression model.

Clinical results of the pilot test

After introducing the study to participants, all 28 willingly joined and reported any pain or discomfort during measurements, both before and after the administration of anesthesia. The entire process, including installation, safety explanations, and the measurement series, took approximately 5 minutes, totalling around 10 minutes per participant for both pre-intervention and post-intervention measurements. The measurements were conveniently conducted in the local anesthetic room, streamlining the procedure.

In terms of transportability, the EOD can be effortlessly handled by a single person and occupies a space comparable to a standard chair (70 cm long, 50 cm wide, 40 cm high, and weighing approximately 16 kg). It is important to note that the EOD does not provide immediate viscous and elastic stiffness values. Instead, raw data is collected on a USB key and processed using in-house software, that take approximately 2 minutes.

DISCUSSION

The primary goal of this experiment was to demonstrate that the PMAS measurement in a state of maximal relaxation was unaffected by voluntary or reflex contraction. The results indicate that there was no significant difference in passive wrist stiffness measurements with or without anesthetic motor block. This suggests that, given the modest size of our population, administering anesthesia or measuring relaxation (by EMG, for example) is not required in a healthy participant. Simply instructing the patient to relax yields comparable assessment of PMAS. Our study found no significant difference in results for viscous and elastic stiffness of the wrist between our 28 participants and a sample of 46 healthy subjects, all of whom shared the same age, gender, height, and weight. To the best of our knowledge, this is the first study to examine the effect of a full anesthetic block on the viscoelastic characteristics of the wrist. The findings suggest that passive stiffness, both elastic and viscous, is closely related to the passive structure of the forearm, including muscles, tendons, and joint capsule. The pilot test indicates that implementing EOD in ordinary clinical practice is feasible.

EMG has traditionally served as a mean to verify the absence of voluntary or reflex disruption of PMAS measurement [76,107] . In case where EMG activity is limited or nonexistent, muscle lengthening is regarded as passive. However, it's important to note that low levels of muscular activity, often undetectable through this method, can introduce errors in PMAS measurement [113]. The primary objective of the current study was to mitigate extrinsic voluntary or "reflex" contraction by completely blocking nerve impulses. Locoregional anesthesia (LRA) achieves reversible suppression of nerve transmission by restricting sodium flux through direct interaction with voltage-dependent sodium channels [114]. Theoretically, an anesthetic block products effect akin to denervation [102]. When considering viscoelastic wrist stiffness, no statistically significant difference was observed between voluntary relaxation and motor anesthetic block. Lakie et al. (1984) reported similar wrist stiffness between subjects in state of voluntary relaxation and those under general anesthesia with medications blocking the neuromuscular connection. [115] Another potential cause of involuntary contractions is stretch reflexes. In healthy patients, sinusoidal joint mobility in area like the elbow [116] and fingers [117] can trigger those reflexes. These reflex contractions are additionally heightened in individuals with spasticity, as demonstrated by Bollens et al. (2013) in the case of the ankle. [118]Such reflex contractions manifest a reduction in elastic stiffness at the frequency of motion corresponding to tissue resonance. Given that the reflex movement is in phase with the motor movement, both

forces act in the same direction. Consequently, the motor movement is reinforced by muscular contraction [116], leading to a measured stiffness that appears lower. The occurrence of reflex contraction increases elastic stiffness at frequencies beyond the resonance point. Plexus anaesthesia is expected to inhibit this reflex mechanism in healthy subjects, our patients, and in individuals with spasticity post-stroke [102]. Furthermore, voluntary contraction by the participant could potentially yield inaccurate PMAS measurement. Such contractions can be addressed during the analysis, and the LRA should serve to mitigate this possibility.

The PMAS is closely associated with muscle size and strength. Stiffness tends to rise with the size of muscle tissue [71,113]. Specially, the cross-sectional area of the wrist flexors is approximately twice that of the wrist extensors. The anisotropic nature of wrist stiffness results in higher resistance from flexors compared to extensors, attributed to their larger volume [119]. Forearm circumference stands out as a significant characteristic linked to PMAS (Nguyen et al., 2020a). The adapted forearm circumference calculation aimed to meticulously isolate the perimeter attributed to muscle tissue, excluding the influence of skin and subcutaneous tissue. Unfortunately, we did not observe a more robust relationship between this adapted forearm perimeter and viscous stiffness compared to the measured perimeter.

There was no observed correlation between stiffness and activity levels (professional or sports-related) or grip strength (Jamar®). The absence of precision in these activity statistics, coupled with variability in the population and the relatively small sample size, may account for outcome. It would be intriguing to conduct a study that compares groups with diverse levels of objective physical exercise. Asymmetrical activities, emphasizing the use of one limb over the other, have been linked to changes in muscle development and biomechanical characteristics. Functional demand play a pivotal role in the development of muscle tissue [120].Additionally, muscle fiber type adapts to the nature of use, with an increasing proportion of slow type I fibers associated with complex tasks and prolonged posture maintenance. The type of muscle fibers may impact the rheological properties of the tissue and, consequently, the PMAS [83]. Muscle strength tends to decrease with age, likely attributed to a reduction in the number and size of motor function units. Histological remodelling occurs with age, characterized by an increase in fat and intramuscular connective tissue, leading to a decrease in passive stiffness. [113] While men generally exhibit greater PMAS than women, this difference disappears after adjustment for body size [68]

This study has certain limits. The minimum detectable change (MDC) in stiffness was found to be 25% [110], indicating limited accuracy in detecting subtle changes in results. These minor variations could be attributed to the non-linear behavior of soft tissues in response to mobilization frequencies or to low-intensity parasitic contractions. One suggested approach to enhance the MDC, as indicated by Nguyen et al, is to improve hand fixation [110]. This study implemented the recommended approach by using Velpeau bandage fixation for hand fixation, providing more comprehensive hand coverage compared to the previous fixation method. Due to the diversity within the population and the limited sample size (28 individuals), it was not possible to detect variations based on specific medical conditions or analyze the newly anthropometric parameters (grip strength, perimeter of the adapted forearm, Beighton score). To enhance the validity of our finding, we included a control group of 46 healthy subjects. The clinical effectiveness of anaesthetic administration was evaluated. Despite implementing several preventive measures to minimize potential bias, including monitoring the time taken to achieve neurologic blockage, there remained a risk of subclinical administration of the anaesthetic, which could potentially interfere with the measurements. [121] Additionally, ultrasound-guided locoregional anaesthesia has been reported to be faster, more effective, and requiring lower doses of local anaesthetic compared to non-ultrasound-guided locoregional anaesthesia. [122]

The present research has highlighted the practical applicability of EOD in a clinical setting. Nguyen et al. (2020) previously established the reliability and reproducibility of the EOD for assessing passive stiffness. Our findings confirm that PMAS measured with EOD is genuinely passive. Future research could explore the potential utilization of the EOD in clinical practice. Investigating the biomechanical adaptations induced by diseases or therapies holds various implications. Improved estimates of the effectiveness of orthopedic surgical and rehabilitation interventions, where stiffness plays a significant role [105], can be anticipated. Additionally, a deeper understanding of spasticity in various contexts such as stroke, head trauma, perinatal accidents, and neurodegenerative pathologies [123,124] is expected. This hold particular promise in the context of spasticity, providing insights to guide future therapeutic indications. Simulation of the effects of interventions, including botulinum toxin intramuscular injection, neurotomy or neurolysis [125], tenotomies [123], or intrathecal baclofen injection [102] could be especially beneficial. LRA would also enable the differentiation of muscular hyperactivity, associated with extrinsic tone, from passive constitutional elements (linked to musculo-articular remodelling associated with chronicity of pathology) [126].According to Kwon et al.

(2014), in the context of deep brain stimulation procedures, quantitative evaluation of joint stiffness can enhance intraoperative positioning of electrodes and refinement of stimulation parameters. [100] Finally, this research contributes to our understanding of viscoelastic stiffness, as well as the impacts of musculature and LRA on it. These factors may prove crucial in the development of virtual surgical models. [103]

CONCLUSION

According to the result of this study, the EOD methodology proved effective in quantifying the passive stiffness of non-contractile tissue in the wrist. The integration of EOD in clinical routine was found to be safe and logistically feasible. This research established a foundation for future explorations of orthopaedic and neurological pathological conditions characterized by abnormal passive joint stiffness of the wrist. It also lays the groundwork for the utilization of EOD as a diagnostic, prognostic, and monitoring tool in these pathologies.

4. The need for an on-field stiffness measurement device

What about the myotonometry?

4.1. Preamble

Electromechanical oscillation (sinusoidal methods) technology has proven to be valid and reliable. [8] It is also one of the few devices capable of reporting viscous stiffness. While user-friendly and portable, as evidence in the previous chapter, it has limitations in use during specific activity, particularly in sports. For activities such as running, it is valuable to monitor PMAS before and after the race, but understanding changes in PMAS during the race could be also crucial. Myotonometry could potentially be the appropriate methods to bridge this gap.

Myotonometry is currently being utilized in various conditions in human. The MyotonPro Digital palpation device, developed by MyotonSAS (Estonia), is the primary used device in research for myotonometry. The MyotonPro is able to assess superficial structures, including muscles, tendons, ligaments, and skin. The measurement method involves recording the damped natural oscillation of soft biological tissue in the form of an acceleration signal, followed by the simultaneous computation of the parameters related to the state of tension, biomechanical, and viscoelastic properties.

MyotonPro appears to be a reliable device for assessing superficial myofascial structures and distinguish the level of stiffness between different intensity of muscle contraction. (Zinder 2011) Research spanning several decades has shown that myotonometry's reliability is predominantly excellent. Zinder's study in 2011 is one example that supports the reliability of the MyotonPro for evaluating muscle and soft tissue properties, demonstrating its utility in both clinical and research settings (Table 7).

Table 7. reported reliability of myoton technology

Intraclass correlation coefficient	Myoton 2	Myoton 3	MyotonPro	
			intra-operator	inter-operator
Fair			Ko et al. (2018) [127] Aird et al. (2012) [128]	
Excellent	Bizzini et al. (2003) [129] Viir et al. (2006) [130]	Chuang et al. (2012) [131] Zinder et al. (2011) [132] Ditroilo et al. (2011) [133]	Liu et al. (2017) [134] Ortega-Cebrian et al. (2016) [135] Liu et al. (2018) [134] Sohirad et al. (2017) [136]	Kelly et al. (2018) [137] Aird et al. (2012) [128] Chang et al. (2020) [138]

However, only few studies have reported on the validity of Myoton technology in comparison to other methods of PMAS measurement. In 2016, Pruyn et al. conducted a study comparing three different methods of stiffness assessment: myotonometry, the vertical hop test, and the free oscillation technique. This research focused on the gastrocnemii muscles and Achilles tendons of female netball players. They found a strong correlation between the vertical hop test and myotonometry for stiffness (r = 0.99) and logarithmic decrement (r = 0.98). This high correlation suggests that myotonometry can be a reliable method for measuring stiffness, comparable to the vertical hop test, which is a more traditional and established method. Additionally, Kelly et al. (2018) conducted a comparison of measurements between Shear Wave Elastography (SWE) and the MyotonPro device in healthy individuals. [137] They reported an excellent intraclass coefficient correlation for both devices. Furthermore, myotonometry appears to be more sensitive to different level of muscle conditions compared to SWE. This increased sensitivity of myotonometry could be attributed to a potential ceiling effect associated with the SWE methodology. To the best of our knowledge, the only study that has reported on the validity of the MyotonPro device using the piezoelectric accelerometry method was conducted by Sohirad et al. (2017). [136] In this study, they used self-made ballistic gel as a medium for validation. They found that the ballistic gel exhibited high consistency between the two methods, particularly concerning stiffness and logarithmic decrement. These initial

studies present promising results. However, myotonometry, specifically the MyotonPro device, has yet to be conclusively validated for detecting viscoelastic properties or the viscous component of tissues. Consequently, the availability and understanding of EOD technology present a valuable opportunity to rigorously test these hypotheses.

4.2. How does the MyotonPro device's measurements of viscoelastic properties compare to those obtained by the EOD in terms of accuracy and reliability?

Nguyen, A. P., Detrembleur, C., Fisette, P., Selves, C., & Mahaudens, P. (**2022**). Myotonpro is a valid device for assessing wrist biomechanical stiffness in healthy young adults. *Frontiers in Sports and Active Living*, *4*, 797975.

WHAT IS ALREADY KNOWN ON THIS TOPIC

- Myotonometry has demonstrated potential validity in its ability to assess superficial myofascial structures, quantifying potential viscoelastic properties in humans.
- MyotonPro device is considered safe, but the specific structures being tested remain unclear.

WHAT THIS STUDY ADDED

- The elastic stiffness measured by the MyotonPro device is the only variable that has been validated.
- The MyotonPro device demonstrated sensitivity in measuring elastic stiffness.
- Myotonometry provided adequate PMAS measurements when compared to EOD results.

HOW THIS STUDY MIGHT AFFECT RESEARCH, PRACTICE OR POLICY

- The MyotonPro device showed significant potential for quantifying elastic stiffness in humans.
- MyotonPro stiffness measurements demonstrated sufficient sensitivity to detect changes in elastic stiffness.
- Researchers and clinicians should exercise caution when considering the viscoelastic measurement capabilities of MyotonPro.

4.2.1. Myotonpro is a valid device for assessing wrist biomechanical stiffness in healthy young adults

ABSTRACT

BACKGROUND: The MyotonPRO is a portable device for measuring biomechanical and viscoelastic properties in superficial soft tissues. The aims of this study are firstly to validate the MyotonPRO compared to a reliable gold-standard frame and secondly to observe the influence of MyotonPRO measurement on the total wrist viscoelasticity.

METHODS: Three silicone polymers with different elastic properties were assessed with the MyotonPRO and with a reference rheometer (Universal Tribometer Mod). Then, a free oscillations method was used to measure the passive elastic and viscous stiffness of the wrist and compared to MyotonPRO forearm measurements.

RESULTS: A one-way ANOVA demonstrated the validity of the MyotonPRO's stiffness ($p=0.001$), decrement ($p<0.001$) and relaxation ($p=0.008$) parameters for measuring the elastic stiffness (K) of the three polymers. Proximal and distal anterior myofascial measurements of the MyotonPRO were moderately correlated to the elastic stiffness ($p=0.0027–0.0275$, absolute r=from 0.270 to 0.375) of the wrist while the postero-distal myofascial tissues of the forearm demonstrated a moderate correlation with the viscous stiffness of the wrist ($p=0.0096–0.0433$, absolute r=from 0.257 to 0.326).

DISCUSSION: The MyotonPRO is a valid device for measuring elastic stiffness as well as a portable, affordable and easy-to-use tool for quantifying the biomechanical properties and viscoelasticity of myofascial tissue in healthy subjects.

Keywords: myotonometry, viscoelasticity, elasticity, viscosity, myofascial stiffness,passive musculoskeletal stiffness

INTRODUCTION

The understanding of the biomechanical and viscoelastic stiffness of the musculoskeletal system is of great interest in numerous fields such as health, sports and physical activities. These properties are important components of joint stability and movement control. [139] In mechanical engineering, this stiffness, or K, can be defined as the relationship between the applied stress and the induced strain in a system/structure. In biomechanics, we use the term 'elastic stiffness' to define the 'k'. [59] This term differs from 'elasticity', which is used sometimes and refers to the damping effect of a tissue, e.g. skin elasticity. In health, the quantification of stiffness is relevant in neurological disorders, e.g. spasticity [8] and Parkinson's disease [140] or musculoskeletal injuries. It allows a better understanding of the pathophysiology of these disorders, as well as better follow-up and treatment decision-making. In sports, the assessment of stiffness allows a better understanding of physical performance and the prevention of health problems. [4,141,142]

The MyotonPRO (Myoton AS, Tallinn, Estonia) is a non-invasive hand-held, affordable and easy-to-use myotonometer device aimed at recording the biomechanical and viscoelastic stiffness of myofascial tissues. [5,128,129,132] The data obtained with the MyotonPRO consist of five parameters. Three of them correspond to biomechanical stiffness: frequency, representing the tonus or state of active tension, dynamic stiffness and decrement (or elasticity). The last two represent viscoelastic stiffness: relaxation time and creep. [5] Many studies have assessed its feasibility and reliability in a healthy population, both young and older individuals, [143,144] in spastic patients [145,146] and in musculoskeletal disorders. [147] The MyotonPRO elicits a good to excellent reliability in the quadriceps femoris, [148] the biceps brachii [149], the upper trapezius [150], the paraspinal muscles [151], the gastrocnemius and Achilles tendons [152], and the pelvic floor muscles [153]. Previous studies did not usually assess the five parameters together and focused only on one or two of interest, with stiffness as the main component. To the best of our knowledge, no study has determined the validity of the five parameters with respect to an accurate and valid gold-standard measurement tool for elastic stiffness or a device that differentiates elastic from viscous stiffness.

The free oscillations technique is one of the valid methodologies used to quantify viscoelasticity in the extremities of the joints of limbs.[8,110] The electromechanical oscillation device (EOD) is a proven, reliable tool for assessing and quantifying the passive elastic and viscous stiffness of the ankle [64] or wrist joint.[110] Elastic stiffness and viscous stiffness

represent respectively the spring-like and damper-like properties of tissues. However, while the EOD can measure the total stiffness of a joint, it cannot identify the involvement of the capsule joint, ligaments, muscle or tendon tissues in the total measured stiffness.

The aims of the present study are: i) to assess the validity of the five parameters measured by the MyotonPRO compared to an accurate and reliable testing machine tool, a tribometer, which assesses biomechanical properties using silicone polymers; and ii) to establish the relationship between forearm myofascial tissue, as measured by the MyotonPRO, and its correlation to elastic and viscous stiffness of the wrist in healthy young adults, as measured by the EOD.

Methods

This observational study was performed by using two standard reference systems, a tribometer and the EOD. The accuracy of the five MyotonPRO parameters in assessing elastic stiffness was evaluated on three different polymer surfaces and was compared to the tribometer assessment. The relationship between the passive stiffness of the wrist as assessed by the EOD and the measured stiffness of posterior and anterior, distal and proximal myofascial tissues of the forearm as assessed by the MyotonPRO, was evaluated on healthy young volunteers. This study received the approval of the local ethics committee (Comité d'Ethique Hospitalo-Facultaire) from the Université Catholique de Louvain (UCLouvain, Belgium – N°: B403201942384) and was performed in accordance with the principles of the Declaration of Helsinki. All 52 participants provided written informed consent.

Materials

MyotonPRO

The MyotonPRO produces a series of damped oscillations induced by a single external mechanical impulse. The probe of the MyotonPRO was placed perpendicularly to the assessed structure and, with a pre-force of 0.18 N, a quick mechanical impulse of 0.4 N was released for 15 ms. The total force applied to the structure was 0.58 N. Following the single external mechanical impulse, free oscillations were recorded for 400 ms. Then, the device needed 10 ms for processing raw signals and 5 more to compute all the parameters. The sequence was looped five times for a total duration of 2.15 seconds. The mean value of the five sequences corresponds to one assessment allowing the calculation of five parameters: frequency, stiffness, decrement, relaxation time and creep. Frequency (F) corresponds to the oscillation frequency and represents either the tonus of the muscle or the state of tension if the muscle is contracted: $F\ [Hz] = Fmax$ (Eq.1). Stiffness (S) should represent the k of the structure: $S\ \left[\frac{N}{m}\right] = \frac{a_{max}\ .m}{\Delta l}$ (Eq. 2), where a = acceleration of the probe, m = the mass of the probe and Δl = displacement. The decrement (D) is described as the logarithmic decrement of acceleration between the first and second cycle. It represents the loss of energy and is described as the elasticity of the system: $D = \ln\ (\frac{a_1}{a_3})$ (Eq.3), where a_1 and a_3 are the maximal accelerations of the first and second cycles, respectively. A decrement value of 0 would represent a perfect elasticity with no loss of elastic energy (Figure 21). Relaxation time is

the delay between the maximal value and the end of the first cycle. It represents the time that the structure needs to recover its initial position. Creep is a representation of the Deborah number. It is a dimensionless number that characterizes the fluidity of a material under specific conditions. [154] As regards the MyotonPRO methodology, it is defined as the gradual elongation of tissue over time when placed under a constant tensile stress. [5] The more solid-like the material is, the higher the creep (or Deborah number) will be.

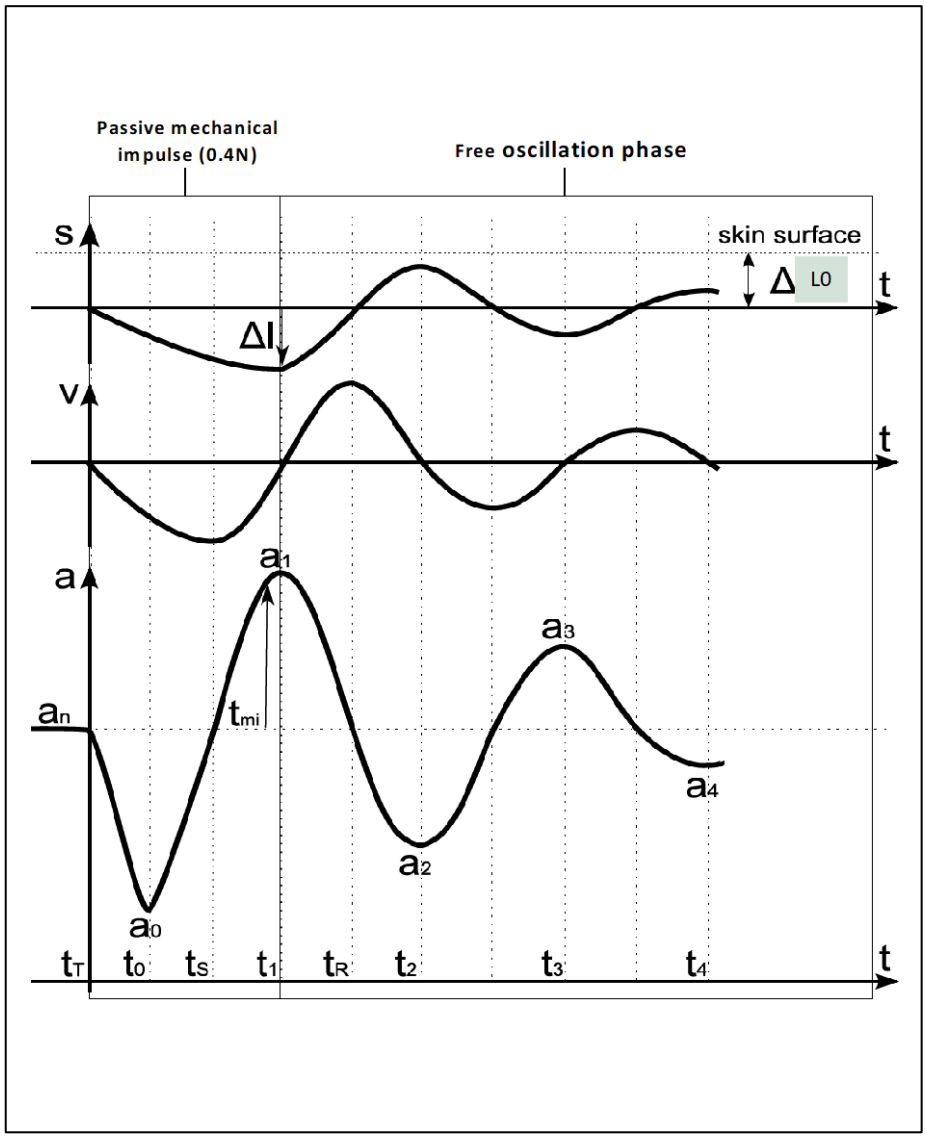

Figure 21. *Description of the induced mechanical impulse and the free oscillation phase and the relationship with displacement (l), velocity (V) and acceleration (a). Adapted from the MyotonPRO user manual (MyotonPRO AS, London, United Kingdom) and previously described by Schneider et al. in 2015.* [5]

Tribometer and polymers

The elastic stiffness of the polymers was assessed using a universal tribometer mod (UMT-3MT) equipped with a testing block, s/n T45185 (Bruker-Nano Surfaces Division, Campbell, CA, USA). Polymers were created using 'Poly-addition' silicone SICASIL® materials (Capron Podologie, Ecuisses, France). Different dosages of catalysis and base layers (1:1, 1:2, 2:1) were adjusted to obtain three polymers of different density and stiffness. The tribometer device executed two tests at three different velocities (0.5 mm s^{-1}, 1 mm s^{-1} and 2 mm s^{-1}) to verify the k, which was independent of speed, for each polymer. The stiffness, k, of each polymer was calculated as the slope of the vertical force (Fz recorded during each tribometer test at 1024 Hz) over the vertical displacement (simultaneously recorded). All measurements were taken in the middle of the top of the polymers, which was marked with an indelible dot in order to standardize all measurements at the same testing spot. MyotonPRO measurements were assessed twice on each polymer and compared to the k of each polymer.

Electromechanical oscillation device (EOD) in healthy subjects

The EOD generated sinusoidal rotatory movements to assess the viscoelasticity of the wrist. This methodology was developed by Rack in 1966,[13] adapted by Lehman in 1989 [14] and followed by Detrembleur and Plaghki in 2000. [8] The EOD induces a passive flexion/extension movement (±10°) of the wrist at several frequencies from 3 Hz to 12 Hz. A potentiometer and a torquemeter record the angular displacement and torque. By computing the data and assuming that the subject does not exert any voluntary contraction, the EOD extracts the Elastic and viscous stiffness values of the wrist. Elastic stiffness (EL) is defined as the restoring force that acts like a spring mechanism and is directly proportional to the deformation of the structure and independent of the velocity of the deformation, as opposed to viscous stiffness (VI), which acts as a damper and which is correlated to the velocity of the deformation. EOD technology has been validated as a safe, reliable device for measuring passive viscoelastic stiffness in healthy subjects. [110] The participants sat in front of the EOD, with their elbow flexed at about 120° and their forearm relaxed. The wrist was firmly attached to the gripping handle to keep the fingers passively closed around the handle. The fingers are held in this way thanks to a strap in the form of a glove enclosing

the fingers around the handle with a velcro system to prevent the fingers from opening and releasing the handle. For this evaluation, only the dominant wrist was tested. The participant either began with the EOD or the MyotonPRO measurement, following a simple randomization (with sealed letter).

The MyotonPRO was applied at four different points of the dominant forearm. Two lines were drawn from both medial and lateral epicondyles of the humerus to the radial styloid to delimit the four measurement sites. The intersection of these lines at the most prominent part of the forearm represented the first site, on the palmar side, the proximal flexor (PF), and on the dorsal side, the second site, the proximal extensor (PE). The middle of the line between the PF and the radial styloid and the PE and the radial styloid corresponded to the third and fourth sites, respectively the distal flexor (DF) and distal extensor (DE). All marked sites are shown in Figure 22.

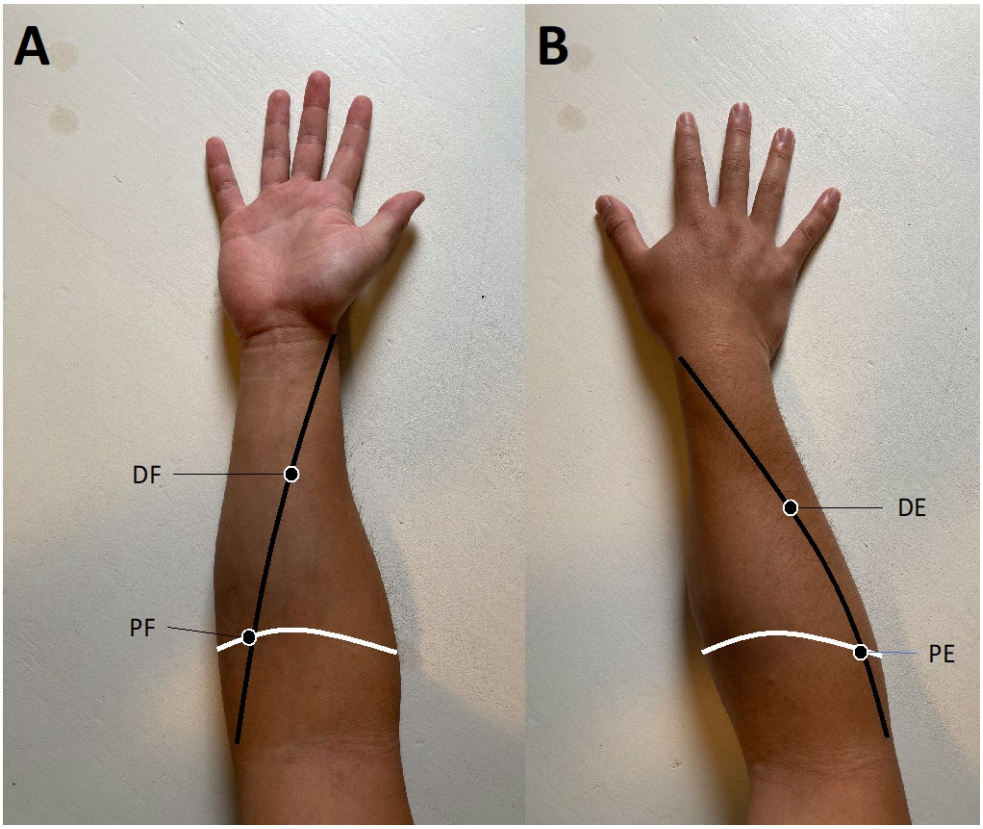

***Figure 22**. A represents the anterior face of the forearm while B represents the posterior part of the forearm. The white line determines the most prominent part of the forearm. The black line represents the line from the medial epicondyle (A) and the lateral epicondyle (B) to the radial styloid. These lines determine the four measurement sites: PF: the proximal flexor; DF: the distal flexor; PE: the proximal extensor; and DE: the distal extensor.*

Participants

Wrist stiffness was assessed using both the MyotonPRO and the electromechanical oscillation device (EOD) in 52 healthy participants. Participants were included if they were between 18 and 30 years old, presented no musculoskeletal disorders of the upper limb and had not engaged in any physical activity within the 24 h prior to the experience. Participants were recruited from the university and surroundings areas (UCLouvain, Belgium) and received no financial compensation.

Statistical analysis

Statistical analysis was performed using Sigmaplot version 14.0 (Systat Software, San Jose, CA, USA) and MedCalc Statistical Software version 16.4.3 (MedCalc Software, Ostend, Belgium).

Tribometer vs MyotonPRO: polymer surfaces

A One-Way ANOVA was performed in order to test the effect of three different polymer densities on k parameters (tribometer) and MyotonPRO's parameters (F, S, D, R and C). Due to the small sample of values, a post hoc analysis was performed using the Holm-Sidak method. Measurements between the three polymers were also expressed in per cent.

EOD vs MyotonPRO: healthy participants

In order to compare EOD and MyotonPRO measurements in healthy subjects, we normalized all parameters as the units of those parameters were different and could not be compared without standardization. More precisely, we processed all parameters following this equation: $normalized\ (x) = \frac{x - \min\ (x)}{\max(x) - \min\ (x)}$, where x represents the parameter of interest. We performed a Passing & Bablok linear regression [155] on each normalized value for MyotonPRO (X variables), i.e. F, S, D, R and C, compared to EOD (Y variable), i.e. EL or VI. The intercept (A) and slope (B) of the linear regression were calculated with a 95% interval confidence. The intercept A is a measure of the systematic difference between the two methods. This hypothesis is accepted if the confidence interval for A contains the value 0. The slope B is a measure of the proportional difference between the two methods. This hypothesis is accepted if the confidence interval for B contains the value 1. As the Intercept (A) represents the systematic difference, Bland-Altman plots was not performed. Finally, a Spearman rank correlation coefficient was calculated for all significant interactions. The level of significance for all statistical analysis was $p < 0.05$.

RESULTS

Tribometer vs MyotonPRO: polymers

Tribometer measurements showed significant differences between the three polymers ($p<0.001$). MyotonPRO statistical results showed significant differences between polymers for stiffness ($p=0.001$), decrement ($p<0.001$) and relaxation time ($p=0.008$). Post hoc analysis showed that all parameters were different between all polymers (Table 8).

Table 8. Tribometer (k) and MyotonPRO parameters for polymers A, B and C

	Polymer B		**Polymer C**		**Polymer A**		**p-value**
Tribometer	**mean**	**SD**	**mean**	**SD**	**mean**	**SD**	
k (N m^{-1})	**1.436**	0.024	**2.420**	0.036	**2.732**	0.021	**<0.001**
MyotonPRO							
Frequency Hz)	60.8	6.22	56.9	0.77	65.2	5.44	0.360
Stiffness (N)	**1341.0**	25.45	**1494.0**	5.65	**1610.5**	9.19	**0.001**
Decrement	**0.35**	0.01	**0.41**	0.001	**0.61**	0.007	**<0.001**
Relaxation (s)	**6.05**	0.07	**5.8**	0.00	**5.55**	0.07	**0.008**
Creep	0.63	0.01	0.62	0.00	0.61	0.01	0.354

SD = Standard deviation. P-values are the results of the post hoc analysis (Holm-Sidak). The significant value and p-value are indicated in bold.

The change assessed by the tribometer and expressed in per cent showed a 47% difference between Polymers A and B, 11% between Polymers A and C and 36% between Polymers B and C. Similar changes were observed with the MyotonPRO for the stiffness parameters, respectively 42, 17 and 32 %. The changes observed between polymers for D and R were not comparable (17, 7 and 11 % for S; 9, 5 and 4 % for R) (Figure 23).

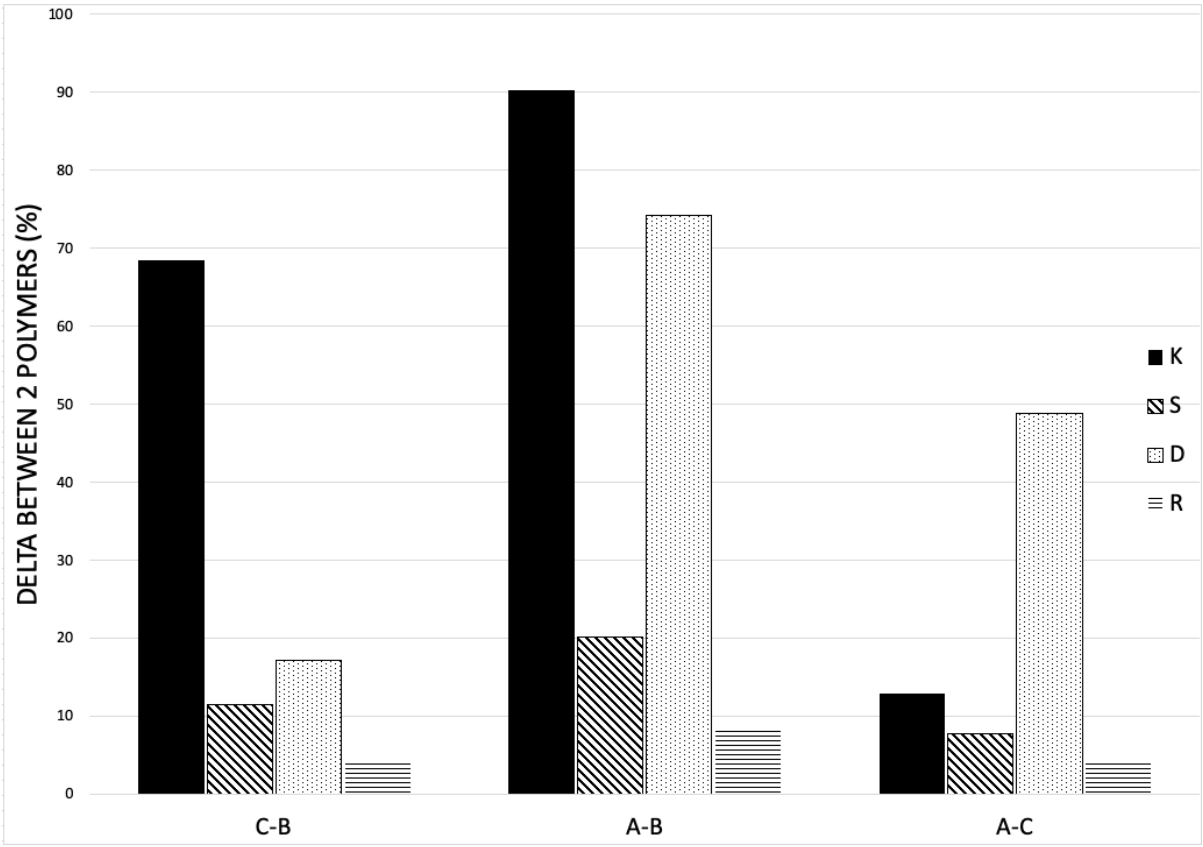

***Figure 23**. Delta between polymers. A=Polymer A, B=polymer B, C=polymer C. k=Stiffness or elastic stiffness, measured with the tribometer. S=(dynamic) stiffness, D=Decrement, R=Relaxation time. S, D and R were measured with the MyotonPRO.*

EOD vs MyotonPRO: healthy participants

Of the 52 participants, all were included in the study. There were 31 men and 21 women (22.8 ± 2.37 years, 173.8 ± 9.42 cm, 69.9 ± 11.68 kg). With the exception of the decrement of the DF, PE and DE, no systematic differences were found between the MyotonPRO's parameters and EL values (Table 9). Likewise, no systematic differences were found

between MyotonPRO and VI values, except for the decrement of the PE. Due to the normalization of values, no proportional differences were identified. Spearman rank correlation coefficients can be seen in Table 9. The MyotonPRO's parameters such as frequency, stiffness, relaxation and creep in the flexor sites were significantly ($p<0.05$) correlated to EL. As for VI, only the stiffness and relaxation in the PF were significantly correlated ($p=0.03$ and $p=0.04$). Finally, the frequency, stiffness, relaxation and creep of the DE were significantly ($p<0.05$) correlated to VI.

Table 9. Results of Passing & Bablok linear regression

EL	Systematic differences		Spearman rank correlation	
	Intercept A	95% CI	p-value	r
PF-Frequency	0.04144	-0.1120 to 0.1479	**0.0279**	**0.279**
PF-Stiffness	-0.09245	-0.2438 to 0.03488	**0.0102**	**0.324**
PF-Decrement	0.1732	-0.08000 to 0.3308	0.1821	
PF-Relaxation	0.04439	-0.4201 to 0.2746	**0.0077**	**-0.336**
PF-Creep	0.02448	-0.4306 to 0.2095	**0.0149**	**-0.308**
DF-Frequency	0.146	-0.01771 to 0.2205	**0.0063**	**0.344**
DF-Stiffness	0.06226	-0.08075 to 0.1996	**0.0092**	**0.328**
DF-Decrement	0.2014	0.003487 to 0.3358	0.7672	
DF-Relaxation	0.1798	-0.1835 to 0.3580	**0.0027**	**-0.375**
DF-Creep	0.1625	-0.2249 to 0.3466	**0.003**	**-0.371**
PE-Frequency	-0.01575	-0.3068 to 0.1398	0.6302	
PE-Stiffness	-0.2106	-0.7430 to 0.09951	0.3372	
PE-Decrement	0.4113	0.2125 to 0.5311	0.0126	-0.315
PE-Relaxation	-0.06436	-0.3771 to 0.05531	0.3671	
PE-Creep	-0.03836	-0.3277 to 0.08602	0.3376	
DE-Frequency	0.1276	-0.1456 to 0.2620	0.3618	-
DE-Stiffness	0.1441	-0.01739 to 0.2392	0.3933	-
DE-Decrement	0.2688	0.1519 to 0.3655	0.8229	-
DE-Relaxation	0.1777	-0.01753 to 0.2694	0.1834	-
DE-Creep	0.1678	-0.03159 to 0.2813	0.1559	-

Table 9 (continued)

VI	Systematic differences		Spearman rank correlation	
	Intercept A	**95% CI**	**p-value**	**r**
PF-Frequency	0.002473	-0.2905 to 0.1918	0.1408	
PF-Stiffness	-0.1217	-0.3473 to 0.04591	**0.0354**	**0.268**
PF-Decrement	0.1585	-0.1376 to 0.2980	0.3276	
PF-Relaxation	0.09816	-0.4260 to 0.2636	**0.0433**	**-0.257**
PF-Creep	0.047	-0.5253 to 0.2239	0.0689	
DF-Frequency	0.1122	-0.06703 to 0.2658	0.4079	-
DF-Stiffness	-0.009202	-0.2599 to 0.2354	0.5557	-
DF-Decrement	0.1914	-0.03304 to 0.3133	0.6392	-
DF-Relaxation	0.06471	-0.2544 to 0.2210	0.3085	-
DF-Creep	0.04279	-0.2409 to 0.2084	0.336	-
PE-Frequency	-0.1727	-0.7130 to 0.06490	0.2951	-
PE-Stiffness	-0.4413	-1.9636 to 0.00462	0.0651	-
PE-Decrement	0.401	0.08415 to 0.4984	0.1304	-
PE-Relaxation	-0.1517	-0.4068 to 0.02574	0.0807	-
PE-Creep	-0.1258	-0.3590 to 0.05175	0.0748	-
DE-Frequency	0.04843	-0.3547 to 0.2464	**0.0254**	**-0.284**
DE-Stiffness	0.1513	-0.1095 to 0.3045	**0.0141**	**-0.31**
DE-Decrement	0.1673	-0.04759 to 0.3092	0.8714	-
DE-Relaxation	0.1433	-0.02971 to 0.2612	**0.0097**	**0.326**
DE-Creep	0.1487	-0.05152 to 0.2684	**0.0096**	**0.326**

No systematic difference was established if the 95% CI (confidence interval) was 0. Significant Spearman correlation p-values are indicated in bold. r = coefficient correlation (in bold). PF = proximal flexor; DF= distal flexor; PE= proximal extensor; DE= distal extensor. MyotonPRO parameters correlated to the EOD measurements are indicated in bold and underlined.

DISCUSSION

The first aim of this study was to assess the validity of the five MyotonPRO parameters in regard to the elastic stiffness. For this purpose, we used silicone polymers with different stiffness and compared the gold standard, i.e. the tribometer, with the MyotonPRO measurements. We found that k (the elastic stiffness) could be validly assessed by the (dynamic) stiffness, decrement and relaxation time of the MyotonPRO device. Secondly, we assessed the influence of the biomechanical and viscoelastic properties of the forearm myofascial tissues measured by the MyotonPRO on the passive stiffness of the wrist joint tested with the EOD. We observed a small correlation between the anterior forearm tissue for F, S, R and C and the elastic stiffness of the wrist. Also, a small correlation between the distal posterior tissue and the proximal and anterior tissue of the forearm F, S, R and C was found in regard to the viscous stiffness.

Tribometer and polymers

To the best of our knowledge, the present study is the first to compare the five MyotonPRO parameters to an accurate and valid device to measure polymer elasticity. Sohirad et al. previously used gelatin polymers and compared piezoelectric accelerometer measurements to MyotonPRO measurements. [136] Similarly to our results, they found that stiffness and decrement were consistent with the modification of gelatin concentration (and therefore the stiffness, k). Another study found that stiffness measured with the MyotonPRO was highly correlated (r^2= 0.96) with Young's modulus of a gel-based polymer, calculated with a compression test of 100 N at a velocity of 20 mm/min. [156] Based on our results, we confirmed that stiffness and decrement parameters are related to elastic stiffness (k) measurements. The stiffer a structure is, the more acceleration it will induce on the MyotonPRO's probe, increasing the stiffness parameter. Likewise, the less stiff it is, the more energy it will release between two oscillation peaks, decreasing the decrement value, and thus increasing the elasticity. We also observed that relaxation time was related to elastic stiffness. Conversely, we could not confirm that frequency and creep parameters were relevant in assessing passive elastic stiffness. This might be due to a lack of validity among those parameters.

Furthermore, an *in vivo* study compared the MyotonPRO parameters to ultrasound shear wave elastography in myofascial tissue.[137] Lohr et al.[157] compared MyotonPRO parameters

with tensiomyography measurements. [158] Both studies demonstrated that MyotonPRO parameters were more sensitive to modifications of joint position [158,159] or intensity of muscle contraction.[137] We confirmed that the MyotonPRO demonstrated sufficient accuracy in detecting small changes in elastic-related stiffness.

EOD and healthy participants

To the best of our knowledge, this is the first study to investigate the local biomechanical and viscoelastic properties of the forearm myofascial tissue and its relationship with the passive stiffness of the wrist. It is well known that the passive stiffness of the wrist is multifactorial. [70] The wrist's passive stiffness could be induced by several biological structures, such as the muscle-tendon unit, the skin, the subcutaneous tissue, the fascia or ligament, the cartilage or the joint capsule.[70] Also, external factors could influence the passive stiffness, such as ambient or body temperature, blood flow, alcohol consumption, time of the assessment or physical activities.[149] A device such as the EOD can determine the elastic and viscous stiffness of the wrist joint but cannot differentiate the influence of the biological structures surrounding the wrist region. [110]

Zonnino et al. demonstrated the influence of the fingers and thumb muscles in the passive stiffness of the wrist with a computational framework based on a musculoskeletal model in OpenSim. [96] We partially confirmed this hypothesis with our results. The anterior myofascial tissues of the forearm showed a small but significant correlation with the elastic stiffness of the wrist measured by the EOD in healthy subjects. Furthermore, our study was the first to compare MyotonPRO parameters with viscous stiffness as assessed by the EOD. In the wrist, our results showed that the DE and stiffness and relaxation time of the PF were related to the viscosity of the wrist. In our opinion, the MyotonPRO cannot measure all the surrounding tissues involved in wrist stiffness as several structures cannot be palpated. The MyotonPRO is described as relevant if the tissue of interest is palpable and located at least 2 cm below the skin. In contrast, the decrement value, representing elasticity or the energy loss following the mechanical impulse, did not demonstrate any significant correlation with either elastic or viscous stiffness. In the literature and in our study, the stiffness parameters seemed to be the most relevant parameters measured with the MyotonPRO compared to F, D, R or C. In addition, Pruyn et al. compared the MyotonPRO's stiffness parameters to

vertical stiffness of the lower limb, measured with a unilateral hopping test, and free oscillation of the calf. [160] Their results showed that the MyotonPRO parameters correlated to many performance assessments, such as the eccentric isometric ratio of the quadriceps and hamstring, squat jump or counter-lateral hop test. They found convergent results between the leg's vertical stiffness and the MyotonPRO's stiffness parameters.

Based on our results, the MyotonPRO revealed itself to be a suitable device for performing *in vivo* measurements of myofascial tissue's viscoelastic properties. Unlike other, more accurate devices such as magnetic resonance elastography,[65] ultrasound shear wave elastography [66161] or an EOD, the MyotonPRO is by far more portable, affordable and easy to use, even for a novice operator.[128,143,162] The stiffness measurement collected in passive conditions with the MyotonPRO appears to be related to dynamic muscle performances [163] as well as passive elastic and viscous joint stiffness in the wrist.

Clinical implications

Our study confirmed some underlying questions that were not completely answered in regard to MyotonPRO assessment. The stiffness parameter is mostly related to passive elastic stiffness. In the neurological field, several authors have studied the usefulness of the MyotonPRO in the quantification of spasticity in stroke patients,[164] paratonia patients [146165] or patients with spinal cord injury disorders. [127] In musculoskeletal pathologies, the MyotonPRO technology had already been used in myofascial disorders,[147] anterior cruciate ligament rehabilitation [135] and chronic low back pain follow-up.[151] Based on our results, future studies should focus their attention on the stiffness parameters measured by the MyotonPRO.

Quantifying and following passive musculoarticular stiffness in myofascial tissues and joints is a growing field of interest among biomechanics specialists, coaches, researchers and clinicians with a view to better understanding movement and its intrinsic properties. This could enhance the validation of therapeutic approaches or sport performance issues.[166–168]

Limitations

This study presented several limitations. Firstly, the stiffness MyotonPRO parameters could be confirm in a silicon polymer but a perfect comparison with myofascial tissues should be taken with caution. Secondly, despite showing excellent relative and absolute reliability in the literature, the MyotonPRO showed some variability depending on the muscle tested.[143,169,170] The present study did not assess the reliability of forearm tissues with the MyotonPRO. Furthermore, the sample population was relatively young and could influence the results [110] Therefore, we cannot generalize the results to the general population. Thirdly, we used a method of standardization with bone anatomical references rather than manual palpation. We thus tried to increase the reproducibility of measurements; however, we lost the ability to identify anatomical divergences in the tested tissue. Therefore, we could not specify the nature of the tissue targeted by the MyotonPro's probe. Finally, we had to normalize the EOD and MyotonPRO value in order to perform the Passing & Bablok analysis.

Conclusion

The stiffness and decrement parameters measured with the MyotonPRO technology are valid and demonstrated accuracy in identifying materials with different elastic stiffness. Furthermore, the MyotonPRO frequency, stiffness relaxation time and creep of the forearm showed small but significant correlations with passive elastic and viscous stiffness of the wrist. More particularly, the proximal and distal anterior myofascial tissues of the forearm showed a small relationship with the elastic stiffness of the wrist while the distal tissue of the posterior side of the forearm demonstrated a relationship with the viscous stiffness of the wrist.

4.3. What are the optimal sites on a runner's body for stiffness measurements using the MyotonPro device to obtain the most relevant and reliable data?

AnhPhong Nguyen, Julien Verdebout, Antoine Fontaine, Philippe Mahaudens, Christine Detrembleur. Recommendation for stiffness measurement on gastrocnemii and Achilles tendon at rest and in loaded conditions. *Accepted in European Rehabilitation Journal*

WHAT IS ALREADY KNOWN ON THIS TOPIC

- Myotonometry provides reliable measurements in resting tissues.
- The optimal localization for measuring PMAS on the lower limb, particularly in the Achilles tendon, remains unclear.
- The reproducibility of measurements remains uncertain.

WHAT THIS STUDY ADDED

- We provide a Mapping of myotonometry points for measurement on the gastrocnemii and Achilles tendon.
- Myotonometry was conducted in both resting and standing conditions.
- Reproducibility measures were provided for each location and condition.

HOW THIS STUDY MIGHT AFFECT RESEARCH, PRACTICE OR POLICY

- Myotonometry for the lower limb should be performed away from tendon insertion to ensure reliable measurement.
- Normative values for resting and standing conditions could assist researchers and clinicians in providing comparable results.
- Myotonometry could be used for reference or monitoring purposes in pathological tissue conditions.

4.3.1. Recommendation for stiffness measurement on gastrocnemii and Achilles tendon at rest and in loaded conditions

ABSTRACT

Background: It is challenging to offer both clinically accessible and reproductible musculotendinous stiffness assessment in patient. This challenge is particularly pronounced in the ankle plantar flexor muscles, including the gastrocnemii and the Achilles tendon, given their crucial role in both walking and running. Design: cross-sectional studies

Objectives: to establish the reliability and clinically relevant stiffness points of measurement for gastrocnemii and Achilles tendon.

Methods: Stiffness was evaluated using MyotonPro® device on both legs in twenty-two healthy individuals. The Achille's tendon stiffness was measured at four points, and the gastrocnemii at six. Two conditions were assessed: at rest, in lying position and under load while standing. All participants underwent measurements in two sessions with one week apart to evaluate intra- and inter- reliability.

Results: The result demonstrated moderate to excellent intraclass correlation coefficients in the lying condition, with coefficients ranging from 0.735 to 0.933 depending on the location. The standard error of measurement varied from 3.7% to 19.8% depending on the location, and the minimal detectable change ranged from 10.3% in the best condition to 54.5% in the worst. In the standing condition, the results showed poor to good measurement reliability for the Achilles tendon (ranging between 0.000 and 0.636) and excellent reliability for the gastrocnemii (ranging between 0.902 and 0.986).

Conclusion: Researchers and clinicians should exercise caution when assessing Achilles tendon stiffness. The Achilles tendon should be assessed at least 6 cm above the bone insertion point, as this provides reliability and is the only point that can be reassessed between days.

Keywords: stiffness, reproducibility, reliability, recommendation, clinical monitoring

INTRODUCTION

The human body is supported by the musculoskeletal system and is moved by the muscle contractions. The force emitted by the muscles is transmitted to the bones through the tendons. [171] Tendons can also store and release potential energy, thereby enhancing the efficiency of human locomotion. Bipedal locomotion is a characteristic feature of the H*omo specie* and is made possible through a variety of anatomical feature that make the human body remarkable. [172] These features include the orientation of the pelvic bone, the voluminous size of the gluteus maximus, the Achilles tendon, and the arch of the foot, all of which are examples of the incredible evolution of the human species.[21]

To walk or run, humans are highly dependent on the ankle plantar flexor muscles. Sutherland et al. [173]reported that these muscles are essential for transferring weight forward and avoiding a shortened stance phase during walking, which could lead to limping. Moreover, the ankle plantar flexors play a crucial role in determining the anterior, posterior, and vertical ground reaction forces during the propulsive phase of walking. [174] Additionally, this force has been reported as a significant determinant of the transition from walking to running. [175]

In biomechanics, assessing muscle and tendon stiffness can be of interest for monitoring pathological conditions or for athletes seeking performance optimization. Muscle and tendon stiffness can be defined as the relationship between the load applied to the tissue and the amount of deformation that occur within it. [20] Specifically, musculotendinous tissue could reveal excessive or insufficient level of stiffness, increasing the risk of injuries. [176] Several devices can quantify musculotendinous stiffness, such as shear wave elastography (SWE) which is used in both healthy and pathological conditions and has demonstrated excellent reliability in assessing Achilles tendon stiffness. [177] Additionally, several studies have reported that SWE can measure stiffness in muscle such as infraspinatus, erector spinae, and gastrocnemii.[137,178] However, the cost and the level of expertise required to use the SWE are hindering its adoption in clinical setting. [177] For clinicians, trainers or athletes, the quantification of the musculotendinous stiffness must be accessible, user-friendly and quick to operate. [179]

In this regard, the MyotonPro® device is a handheld device that can record stiffness in human tissues, including muscles and tendons. [20] The MyotonPro® has been used in various conditions, both in the upper and lower limbs [127,152]. It has been proven to be valid for measuring stiffness [20] and has shown similar results compared to shear wave elastography

device. [180]Regarding the reproducibility of the MyotonPro® for the plantar flexor muscle, several studies have demonstrated good to excellent reliability and agreement in healthy subjects for both the medial and lateral gastrocnemii muscles [177,181] as well as for the Achilles tendon.[127,152,182] It has also shown good reliability in neurological patients.

However, there are few to no studies that have reported reproducibility measurements on the plantar flexor muscles complex in functional conditions, such as when standing. The assessment of stiffness holds potential relevance in various clinical conditions. For instance, Morgan et al. (2016) found that individuals with tendinopathy exhibited reduced stiffness compared to healthy individuals in a standing position. Furthermore, neurological clinicians and researchers have emphasized the significance of reporting quantitative assessments rather than qualitative measurement to enhance the monitoring and treatment of spasticity. [8,132,183]

Therefore, the first aim of this study was to report the reliability and agreement of measurements for the *triceps surae*, specifically the medialis gastrocnemius (MG), lateralis gastrocnemius (LG), and Achilles tendon (AT), in two standardized conditions: at rest and standing, involving 22 healthy participants. Secondarily, this study reported the reliability and clinically relevant measurement site for the medialis gastrocnemius, lateralis gastrocnemius and Achilles tendon. Our hypothesis is that measurements taken in standing conditions may be less reliable than those taken at rest.

METHODS

This cross-sectional reproducibility study was conducted at the facilities of the University catholic of Louvain (UCLouvain - Belgium) and received approval from the local ethical committee (CEHF N°: B403201942384). The present study adhered to the ethical principles for medical research involving humans as outlined in the Declaration of Helsinki. To prevent gender bias, an equal number of women and men were selected. Each participant voluntary enrolled and provided a written, signed consent. Participants did not receive any financial compensation.

To be included in the study, the participants were required to self-describe as healthy, meaning that had not experienced any disease or musculoskeletal disorders in the last six months. Additionally, each participant had to be above 18 years-old and free from any neurological pathologies. Participants under 18-years-old, those with musculoskeletal disorders or diseases, or those had reported any lower limb pain within the last six months were systematically excluded from the study.

Variables and material

The primary variable measured in this study was the passive stiffness of the triceps surae muscle, assessed using myotonometry. For this purpose, the MyotonPro® device was employed with standard settings. Specifically, the probe applied pre-force of 0.18 N and released a single impulse of 0.4 N over 15 ms duration. Subsequently, the maximal acceleration (a_{max}) of the resulting free oscillations was recorded for 400 ms. This process was repeated five times, with a total estimated duration of 2.15 s. The stiffness (*S*), expressed in $N.m^{-1}$, was determined as the mean value of five acceleration measurements calculated as follow: $S\ (N.m^{-1}) = a_{max}\ .\ \mathrm{m}\ .\ l^{-1}$, where a_{max} is the maximal acceleration of the free oscillation of the tested tissue, m is the mass of the probe and l is the displacement. [20]

Procedure

After enrolling in this study, participants provided general information including age, height, medical history, dominance, and physical activities. Subsequently, one investigator measured

their weight, length of lower limb, and ankle dorsiflexion range of motion (ROM). Ankle dorsiflexion ROM was assessed using the weight bearing lunge test [184] , and both legs were systematically tested.

Ten marks were placed on each leg. Four of them were positioned at 2, 4, 6, and 8 cm above the distal AT insertion on the calcaneus. Additionally, three markers were placed on each gastrocnemius. The markers on the MG were positioned at 15, 30, and 45% of the distance between the medial epicondyle and the lateral malleolus, measured using a flexible ruler. Similarly, the markers of the LG were placed at 15, 30, and 45% of the distance between the fibula's head and the center of the posterior face of the calcaneus. All marker placements were consistently performed by the same investigator (B) (Figure 24).



Figure 24. *location of measurement with the myotonpro device in lying and standing conditions.*

Two measurement positions were recorded. The first position involved assessing the musculoskeletal system at rest. In this case, the participant lay on their belly with a cushion placed beneath their ankles, and their feet were positioned in a resting position without specific

instructions. The second position involved evaluating the musculoskeletal system under mechanical stress. To apply external stress, the standing position was used. To standardize this position, four weight scales were placed under the participant's feet - two under the right foot and two under the left foot. Under each foot, the weight scales were positioned equally along a longitudinal plane. At any given time, each scale should have indicated one-quarter of the total weight of the participant.

The measurements were taken at rest and under stress by two novice investigators, JL and AF, during a single session. Each measurement session followed this order: The first rater (Rater A), the second rater (Rater B) and then, the first rater (Rater A). Participant were asked to leave their markers in place for the second measurement session, which was scheduled one week later. The second session was scheduled during a similar period as the first, either early or late morning or early or late afternoon. For the second session, Rater A took a single measurement.

Statistical analysis

Statistical analysis was performed using the Medcalc software. Reliability was assessed using the intraclass coefficient (ICC), which included within-session intra-operator (ICC 2.1), inter-operator reliability (ICC 3.1), and between-session inter-operator reliability (2.1). Following the recommendations of koo et al. [185] an ICC below 0.5 was considered as poor, between 0.5 and 0.75 was considered as moderate, between 0.75 and 0.9 was considered good, and above 0.9 was considered as excellent. Agreement was estimated by evaluating the standard error of measurement (SEM) and the minimal detectable change (MDC). The SEM calculated as followed: $SEM = SD.\sqrt{1 - ICC}$ with the SD representing the pooled standard deviation and the ICC of the corresponding analysis. The MDC calculated as followed: $MDC = 1.96 .\sqrt{2 .SEM}$. The significance level was set at alpha=0.05.

RESULT

Participants

A total of twenty-two participants (11 females, 11 males) were included in the study. The mean age of the sample was 21.6 years [standard deviation: 1.78], with an average height of 1.74 [0.11] m and an average weight of 70.1 [11.9] kg. Participants engaged in various physical activities, including running, cycling, swimming, general strengthening, dancing, track and field, basketball, and football, with an average of 2.52 [1.17] sessions per week. Each session lasted approximately one hour.

Within day reproducibility

Measurement points for the Achilles tendon demonstrated good to excellent inter-rater reliability at rest but poor to moderate inter-rater reliability in a standing position. In contrast, measurements for gastrocnemii demonstrated moderate to good inter-rater reliability at rest but excellent inter-rater reliability in the standing position (Figure 25). The agreement measurements for the SEM and the MDC at rest ranged from 6.5% to 14.3% and from 17.1% up to 39.7% respectively. In the standing position, the SEM and MDC ranged from 5.3% to 19.5% and from 14.7% to 54% (Table 10).

Table 10. Within day Inter-rater reproducibility

		OP A (N/m)	OP B (N/m)	ICC	95%CI	SEM (N/m)	SEM (%)	MDC (N/m)	MDC (%)
at rest	AT2	878.9	861.1	0.866	0.793 to 0.913	85.1	9.8	235.9	27.1
	AT4	817.2	799.4	0.859	0.781 to 0.908	79.8	9.9	221.1	27.4
	AT6	753.8	752.4	0.933	0.896 to 0.956	53.4	7.1	148.1	19.7
	AT8	677.5	674.2	0.927	0.887 to 0.952	50.7	7.5	140.5	20.8
	GM-15	254.2	243.8	0.886	0.787 to 0.938	16.5	6.6	45.8	18.4
	GM 30	280.5	265.1	0.735	0.506 to 0.857	17.8	6.5	49.3	18.1
	GM +15	343.9	330.1	0.861	0.741 to 0.925	24.8	7.4	68.9	20.4
	GL -15	288.3	273.9	0.661	0.364 to 0.819	40.2	14.3	111.5	39.7
	GL 30	284.3	275.2	0.820	0.664 to 0.903	17.3	6.2	47.9	17.1
	GL +15	315.3	314.3	0.789	0.606 to 0.886	24.4	7.7	67.5	21.4

Table 10 (continued)

Standing	AT2	1121.1	1115.5	0.000	-0.231 to 0.641	217.7	19.5	603.4	54.0
	AT4	1133.1	1089.0	0.000	-0.423 to 0.595	189.7	17.1	525.8	47.3
	AT6	993.8	997.3	0.570	0.200 to 0.768	111.3	11.2	308.4	31.0
	AT8	887.9	883.6	0.636	0.323 to 0.804	91.5	10.3	253.7	28.6
	GM-15	389.0	373.8	0.947	0.900 to 0.971	33.9	8.9	94.1	24.7
	GM 30	440.4	432.1	0.920	0.851 to 0.957	24.8	5.7	68.9	15.8
	GM +15	525.3	523.6	0.946	0.899 to 0.970	34.8	6.6	96.5	18.4
	GL -15	438.5	422.6	0.956	0.918 to 0.976	30.7	7.1	85.0	19.7
	GL 30	447.2	448.8	0.971	0.946 to 0.984	23.8	5.3	65.9	14.7
	GL +15	498.2	507.0	0.959	0.924 to 0.978	30.3	6.0	84.0	16.7

The table provides a summary of the results, with column indicating the mean of measurement by the first operator (OP A) and the second operator (OP B). Standard error of measurement (SEM) is presented in both absolute (N/m) and relative (%) values. The intraclass coefficient (ICC) and 95% confidence interval (95% CI) are also included. Minimal detectable change (MDC) is expressed in both absolute (N/m) and relative (%) values. The rows represent the different measurement locations including AT (Achilles tendon) measurements at 2-4-6-8 cm above the distal insertion of the Achilles tendon. Gastrocnemius medialis (GM) and Gastrocnemius lateralis (GL) measurements at three points: 15% above, 30% (centered), and 15% below the distance between the knee and the ankle.

Intra-rater reliability was reported to be good to excellent reliability for the Achilles tendon and the gastrocnemii at rest and in standing position, except for T4, which demonstrated poor test-retest reliability in standing (Table 11). Our results indicated that the SEM for the Achilles tendon ranged from 6.5% to 9.7% at rest and from 6.9% to 17.1% while participants were in a standing position. Regarding the measurement of the gastrocnemii in both resting and standing positions , the SEM ranged from 5.1% to 8.6% and from 3.7% to 7.7%, respectively. The MDC for both the Achilles tendon and gastrocnemii ranged from 14.9% to 47.3%.

Table 11. Within day test-retest reproducibility

		OP A - 1rst (N/m)	OP A - 2nd (N/m)	ICC	95%CI	SEM (N/m)	SEM (%)	MDC (N/m)	MDC (%)
at rest	AT2	878.9	846.7	0.867	0.793 to 0.913	83.5	9.7	231.5	26.8
	AT4	817.2	803.1	0.899	0.844 to 0.934	67.7	8.4	187.6	23.2
	AT6	753.8	742.7	0.907	0.857 to 0.939	58.6	7.8	162.4	21.7
	AT8	677.5	679.5	0.943	0.911 to 0.962	44.0	6.5	122.0	18.0
	GM-15	254.2	245.8	0.869	0.7552 to 0.9294	17.1	6.8	47.3	18.9
	GM 30	280.5	274.3	0.804	0.6376 to 0.8942	14.3	5.1	39.6	14.3
	GM +15	343.9	339.0	0.916	0.8438 to 0.9546	18.4	5.4	51.0	14.9
	GL -15	288.3	285.1	0.824	0.6719 to 0.9052	19.7	6.9	54.7	19.1
	GL 30	284.3	279.4	0.897	0.8094 to 0.9445	14.3	5.1	39.6	14.0
	GL +15	315.3	324.2	0.756	0.5498 to 0.8686	27.5	8.6	76.2	23.8

Table 11 (continued)

Standing	AT2	1121.1	1060.6	0.776	0.5835 to 0.8797	107.1	9.8	296.9	27.2
	AT4	1133.1	1101.8	0.000	-0.1970 to 0.6541	190.8	17.1	528.8	47.3
	AT6	993.8	989.2	0.751	0.5380 to 0.8665	83.7	8.4	232.1	23.4
	AT8	887.9	880.4	0.818	0.6610 to 0.9021	61.3	6.9	169.9	19.2
	GM-15	389.0	382.1	0.961	0.9277 to 0.9790	29.6	7.7	82.0	21.3
	GM 30	440.4	424.5	0.902	0.8157 to 0.9476	29.1	6.7	80.7	18.7
	GM +15	525.3	516.5	0.954	0.9140 to 0.9750	32.0	6.1	88.7	17.0
	GL -15	438.5	423.5	0.961	0.9272 to 0.9792	29.0	6.7	80.4	18.6
	GL 30	447.2	437.4	0.961	0.9279 to 0.9790	28.0	6.3	77.7	17.6
	GL +15	498.2	492.2	0.986	0.9734 to 0.9923	18.3	3.7	50.8	10.3

The table provides a summary of the results, with column indicating the mean of measurement by the first operator (OPA) and the second operator (OPB). Standard error of measurement (SEM) is presented in both absolute (N/m) and relative (%) values. The intraclass coefficient (ICC) and 95% confidence interval (95% CI) are also included. Minimal detectable change (MDC) is expressed in both absolute (N/m) and relative (%) values. The rows represent the different measurement locations including AT (Achile tendon) measurements at 2-4-6-8 cm above the distal insertion of the Achilles tendon. Gastrocnemius medialis (GM) and Gastrocnemius lateralis (GL) measurements at three points: 15% above, 30% (centered), and 15% below the distance between the knee and the ankle.

Table 12. Between day intra-rater reproducibility

		OP A - 1rst (N/m)	OP A - 3rd (N/m)	ICC	95%CI	SEM (N/m)	SEM (%)	MDC (N/m)	MDC (%)
at rest	AT2	842.7	908.3	0.744	0.628 to 0.822	105.5	12.0	292.4	33.4
	AT4	811.3	849.3	0.818	0.738 to 0.873	85.6	10.3	237.2	28.6
	AT6	755.6	767.0	0.850	0.784 to 0.895	71.9	9.4	199.4	26.2
	AT8	683.5	689.8	0.907	0.866 to 0.935	53.6	7.8	148.5	21.6
	GM-15	250.5	249.5	0.790	0.483 to 0.915	22.9	9.2	63.5	25.4
	GM 30	280.1	274.4	0.689	0.244 to 0.874	19.1	6.9	53.0	19.1
	GM +15	344.6	337.9	0.376	-0.565 to 0.752	51.3	15.0	142.1	41.6
	GL -15	285.6	287.9	0.690	0.212 to 0.877	22.2	7.7	61.5	21.5
	GL 30	280.0	283.8	0.696	0.213 to 0.880	23.7	8.4	65.7	23.3
	GL +15	311.5	328.9	0.707	0.288 to 0.882	29.1	9.1	80.6	25.2

Table 12 (continued)

Standing	AT2	1173.9	1011.6	0.000	-1.091 to 0.268	215.0	19.7	596.0	54.5
	AT4	1126.4	1120.8	0.343	-0.255 to 0.655	156.5	13.9	433.7	38.6
	AT6	1027.2	964.4	0.465	0.0271 to 0.711	123.8	12.4	343.0	34.4
	AT8	896.1	869.8	0.630	0.311 to 0.803	86.3	9.8	239.1	27.1
	GM-15	373.4	399.0	0.774	0.441 to 0.909	76.5	19.8	212.2	54.9
	GM 30	425.4	440.3	0.509	-0.220 to 0.804	67.3	15.6	186.6	43.1
	GM +15	520.2	521.7	0.874	0.678 to 0.950	54.7	10.5	151.6	29.1
	GL -15	415.6	448.0	0.847	0.620 to 0.939	59.6	13.8	165.2	38.3
	GL 30	430.9	454.8	0.896	0.737 to 0.958	46.7	10.6	129.5	29.3
	GL +15	476.9	515.4	0.932	0.789 to 0.975	39.5	8.0	109.4	22.0

The table provides a summary of the results, with column indicating the mean of measurement by the first operator (OPA) and the second operator (OPB). Standard error of measurement (SEM) is presented in both absolute (N/m) and relative (%) values. The intraclass coefficient (ICC) and 95% confidence interval (95% CI) are also included. Minimal detectable change (MDC) is expressed in both absolute (N/m) and relative (%) values. The rows represent the different measurement locations including AT (Achile tendon) measurements at 2-4-6-8 cm above the distal insertion of the Achilles tendon. Gastrocnemius medialis (GM) and Gastrocnemius lateralis (GL) measurements at three points: 15% above, 30% (centered), and 15% below the distance between the knee and the ankle.

Between day reproducibility

Between-day reliability ranged from poor to excellent for both at rest and in standing (Table 12), In the resting position, the Achilles tendon showed moderate reliability at T2 (ICC= 0.744) but excellent reliability at T8 (ICC=0.907), suggesting better reliability when measuring stiffness further away from the calcaneus. Conversely, the Achilles tendon showed poor reliability in the standing position, with only the T8 location reporting moderate reliability (ICC=0.630).

The gastrocnemii demonstrated poor to moderate reliability in resting conditions. However, in standing conditions, they showed moderate to excellent reliability, with ICC values ranging from 0.509 to 0.932 (Table 12). Agreement measurements ranged from 6.9% up to 54.9% depending on the location and conditions assessed (Figure 26).

Discussion

This study represents the first attempt to assess the reproducibility of myotonometry measurements for the gastrocnemii and the Achilles tendon across multiple measurement points and in two standardized conditions: at rest and a standing position. Based on our result, it is advisable to assess Achilles tendon stiffness either at six or at eight centimeters above the distal insertion point using the MyotonPro® device, as it demonstrated the highest reliability measurements in both resting and standing positions. The Achilles tendon exhibited greater reliability at rest, whereas the gastrocnemii were more reliable in a standing condition.

It is interesting to note that the stiffness of the Achilles tendon should be assessed away from the distal insertion point. Indeed, the proximity of the bone for the T2 and T4 sites could lead to increased stiffness values, which could negatively impact reliability. The proximity to the insertion point could introduce noise in the probe's measurement, such as rebound, resulting in higher recorded acceleration and, consequently, artificially increased stiffness values and imprecision. The lack of between-day reproducibility suggests that using MyotonPro® near the bone insertion point for monitoring tendon stiffness in a standing position is not recommended. Chang et al. [180]reported similar results with an MDC of approximately 177 N/m (18%) at the Achilles tendon bone insertion and 59 N/m (5%) at 6 cm above the insertion point. In contrast, Tas et al. [152] reported good ICC of 0.83 and 0.9 for Achilles tendon stiffness (T2) at 0° and 10° of dorsiflexion, respectively, in a population of young healthy males. While these findings are consistent with our results, the loads applied on the Achilles tendon in their study may not be considered clinically relevant. This is because passive stretching of the plantar flexor muscles primarily induces strain on the contractile tissue rather than on the tendon itself. [186] Moreover, there does not appear to be a correlation between ankle dorsiflexion and tendon stiffness. [187] To induce sufficient load on a tendon, the involvement of the contractile tissue is essential.

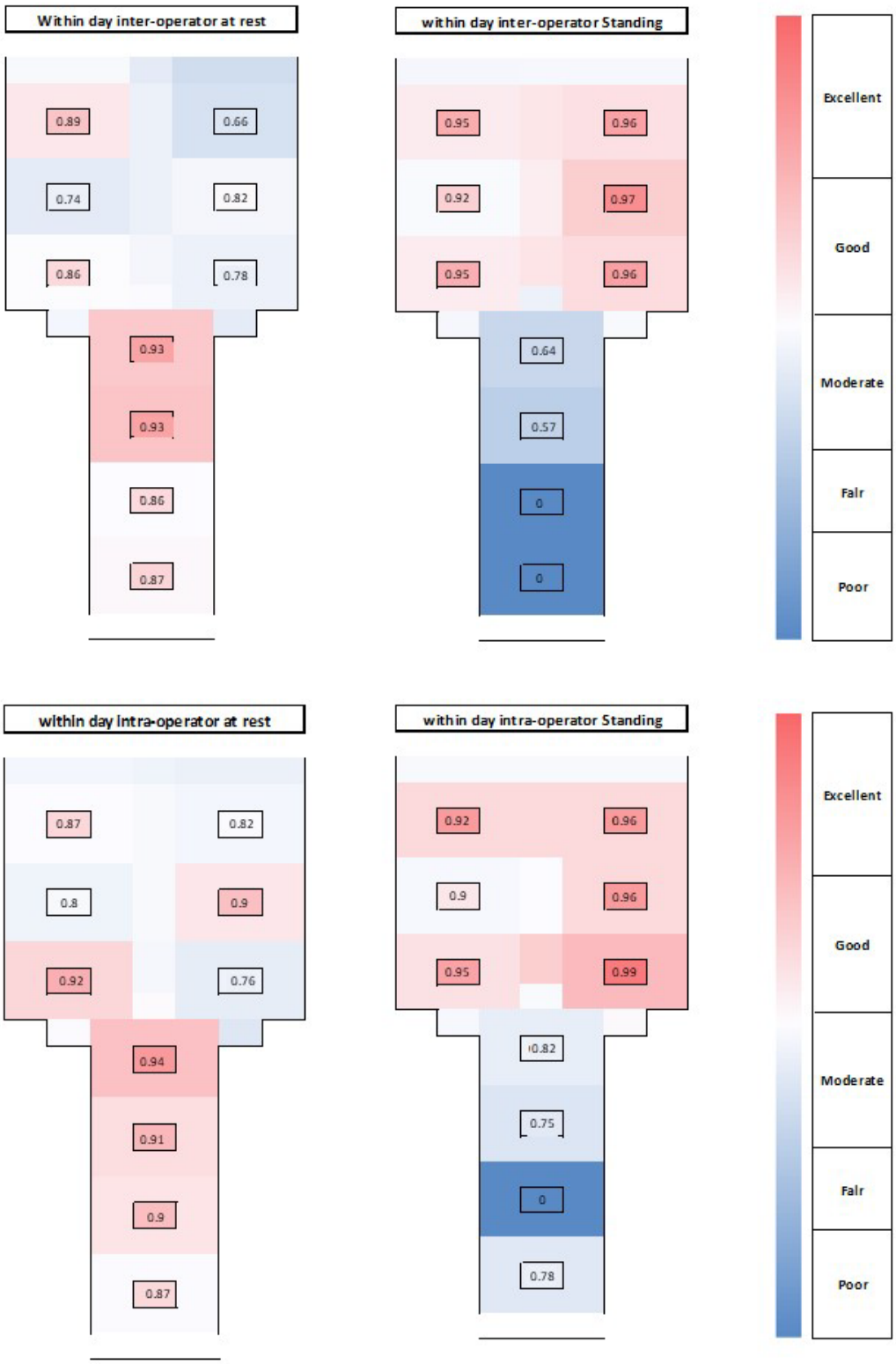

Figure 25. *heat map of within day reliability measurement*

On the other hand, contractile tissue such as the gastrocnemii demonstrated better reliability in a standing condition compared to a resting one. This is consistent with the findings of many studies that have reported good to excellent reliability for the gastrocnemii. [137,177,181] The contraction of the muscle provides more reliable measurements than assessing the loaded tendon in standing position. Kelly et al [137] demonstrated that thit was also true for different loading conditions, with an ICC of 0.97 at 40% of maximal voluntary contraction and 0.98 at 80%. Similarly, Albin et al. [181] demonstrated excellent reliability of the medialis gastrocnemius in both resting and contracted conditions. While their contraction did not mimic weight-bearing condition, it is reasonable to assume that contractile tissue reliability is maintained under loaded conditions.

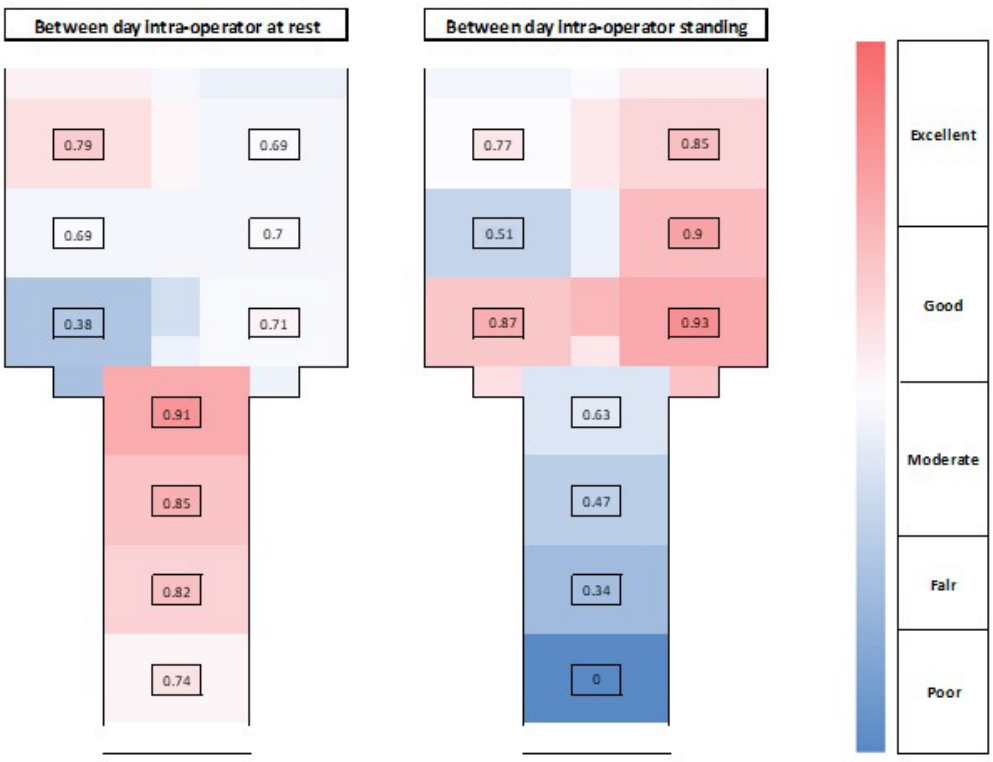

***Figure 26**. heat map of between day reliability measurement*

Based on our results, the assessment of plantar flexor muscle stiffness can be performed on healthy subjects. Finnamore et al.[188] demonstrated that patients with Achilles tendinopathy

could also be evaluated with the MyotonPro®. In their study, they selected the most tender region and the greatest tendon thickness, which corresponds to our T8 site. They reported an ICC of 0.8, a SEM of 43 N/m, and a MDC of 122 N/m, which is equivalent to a 17% MDC based on the findings of our current study. Although Finnamore's result did not exceed the MDC, it was above the SEM. This is crucial in clinical practice as the T8 site has been shown to be the most reproducible site of measurement for the Achilles tendon. In case of Achilles tendinopathy, the tendon may exhibit a smaller difference in stiffness between rest and standing conditions [189] For example, Morgan et al. (2019) reported a less stiff tendon measured with myotonometry in their Achilles tendinopathy group compared to healthy controls.[54] However, these studies did not rely on reliability standards to estimate differences between patients and healthy controls. The present study fills the gap in the literature, making comparisons and rehabilitation recommendations potentially accessible.

The MyotonPro® could provide reliable quantitative measurements for monitoring Achilles tendon rehabilitation and offer clinicians improved possibilities for therapeutic trials. In the context of prevention, the MyotonPro® could offer valuable information about the Achilles tendon for sport stakeholders and provide recommendations for athletes who may be risk of issues. It is worth nothing that the Achilles tendon stiffness is more reliable at rest, while the gastrocnemii are more reliable in standing condition.

The studied population consists of young individuals. It is well-known that myofascial tissue quality can change with age. [110] Therefore, caution should be exercised when interpreting agreement measurement values, as they may not be applicable to older individuals or those with pathological conditions. Additionally, the measurements were provided using bone references. While this approach can enhance the reproducibility of the protocol, this may also yield non-relevant points of measurement in the gastrocnemii depending on the subject's anatomy. While our results align with the available literature, these factors could explain some discrepancies.

Finally, one could argue that the standardization of the standing position could be approximative. To our knowledge, only one other study assessed Achilles tendon stiffness in a standing position. Schneebeli et al. [190] described the reproducibility of the MyotonPro® in standing using a Wii balance board, which allowed for control over the center of mass position. Their reported an intra-rater reliability of 0.87, an inter-rater reliability of 0.56 and an inter-session reliability of 0.75. These results are remarkably similar to ours. While the Wii balance

board can provide real-time information about the center of mass, the techniques used in the present study are easier to employ and record.

In summary, the ankle plantar flexor muscles exhibited good to excellent reliability in stiffness measurement. The Achilles tendon should be clinically monitored at a location at least six centimeters above the distal insertion to the bone, both in resting and standing positions. The gastrocnemii provided excellent stiffness measurements in both resting and standing positions, with the latter being the more reliable methodology.

5. Passive Stiffness in Running

5.1. Preamble

Dumke et al demonstrated a correlation between the stiffness of triceps surae and running economy. [41] It their study, they used a free oscillatory technique with the subject sitting, knees and ankles flexed at 90°. External loading was induced by placing weight on the knee, and the resulting damped force oscillation pattern along the dampened spring model equation were recorded. While this technique offers reliable estimation of PMAS [191], it does not differentiate between elastic and viscous stiffness or between contractile and tendinous tissues.

Building on the findings from the previous chapter, myotonometry could facilitate in vivo and per-protocol measurement in runners. This approach hold promise for improving the observation of rheological characteristics exhibited by runners during their activities, as well as offering a means to monitor the rheological properties of musculoskeletal tissues during recovery or rehabilitation process.

This chapter explores the role of ankle plantar flexor myofascial tissue in running performance, with a particular focus on running economy. It will examine the rheological properties of the gastrocnemii muscles and the Achilles tendon during both an incremental running protocol and a constant speed running protocol. Through this investigation, we aim to understand how these myofascial tissues contribute to the efficiency and cost of running, potentially offering insights into optimizing running practice and performance.

5.2. Is ankle plantar flexor stiffness related to running performance?

Nguyen AP, Bosquet L, Belaoued R, P Mahaudens, C Detrembleur. Impact of ankle plantar flexor stiffness in the metabolic cost of running and running biomechanics. *In review process, Sport Biomechanics*

WHAT IS ALREADY KNOWN ON THIS TOPIC

o Running performance is affected by biomechanical factors including kinematics and kinetic parameters, as well as dynamic stiffness.

o According to the spring-mass model, dynamic stiffness is associated with running performance.

WHAT THIS STUDY ADDED

o Passive stiffness of the Achilles tendon is associated with running performance, particularly the metabolic cost of running.

o The net metabolic cost of running is multifactorial and should be interpreted with caution.

o Passive stiffness of the Achilles tendon at rest is associated with running biomechanics.

HOW THIS STUDY MIGHT AFFECT RESEARCH, PRACTICE OR POLICY

o Running stakeholders should consider the impact of Achilles tendon passive stiffness in training or rehabilitation.

o Implementing exercises that modified rheological properties, such as strength training or plyometric exercises, is relevant for runners.

5.2.1. Impact of ankle plantar flexor stiffness in the cost of running and running biomechanics

ABSTRACT

Objectives: This study aimed to report 1) the correlation between running performance, i.e., net metabolic cost of running (NMC), and passive stiffness, and 2) explore the relationship between passive stiffness and running biomechanics.

Methods: Twenty-two male recreational runners participated in this study. The runners performed a 5-minute run at 50% of their Maximal Aerobic Speed (MAS) as a warm-up on an instrumented treadmill, followed by a 6-minute run at 65% of their MAS, where NMC was recorded. Passive stiffness was measured using a myotonometry device both before and after the run.

Results: Significant results were observed only for measurements taken on the Achilles tendon. There was a negative linear relationship between NMC and pre-run stiffness. Specifically, NMC and Achilles tendon stiffness exhibited a significant negative correlation with leg stiffness and a significant positive correlation with braking peak force and step length. Moreover, we observed an increase in stiffness between pre and post run measurements at rest for the Achilles tendon and both gastrocnemius muscles.

Conclusion: Greater stiffness at rest is correlated with improved NMC.

Keywords: stiffness, running biomechanics, running economy, performance, myotometry

INTRODUCTION

Running is often likened to a spring-mass model, where the body's mass represents the mass and the behavior of the legs represents the spring. [33] This model has catalysed biomechanical research on stiffness and viscoelastic properties, capturing the interest of running stakeholders such as runners, coaches, and healthcare practitioners. [2,33] "Stiffness" encompasses passive stiffness, derived from the mechanical properties of musculoskeletal tissues, and active stiffness, induced by muscle contraction.

In the context of running, active stiffness emerges from a blend of active and passive components, yielding variables such as vertical stiffness (Kvert), joint stiffness (Kjoint), and leg stiffness (Kleg). [18,192] Passive stiffness primarily stems from the mechanical characteristics of skin, joint capsules, ligaments, and other passive myofascial components. [110] Enhanced lower limb stiffness can improve the storage and release of elastic energy, thus boosting power output during the push-off phase, and, consequently, the energy efficiency of runners—i.e., running economy and metabolic cost of running. [193,194]

Running economy, defined as the oxygen consumption (ml kg^{-1} min^{-1}) at a steady running pace, gauges efficiency of movement and is a pivotal determinant of performance in middle- and long-distance events, alongside maximal oxygen uptake ($\dot{V}O_2max$) and its fractional utilization during exercise. [195] It's notably crucial for predicting performance among individuals with comparable $\dot{V}O_2max$ [196] and has been identified as a key factor in the global dominance of East African runners in international competitions. [193] Similarly, the net metabolic cost (NMC) of running, denoting overall energy expenditure (J $kg^{-1}min^{-1}$), serves as a measure of running performance at a specific pace. Factors influencing running economy (RE) or NMC range from environment and physiology to biomechanics, anthropometry, and training. [194]

The linkage between RE and performance is well-established. Initial studies comparing elite American distance runners highlighted that those with superior RE also demonstrated enhanced performance levels. [194] For instance, a 5% increase in RE could potentially elevate distance running performance by 3.8%. [197] Moreover, improvements in RE have been shown to reduce perceived effort and enhance performance in top athletes. [196] Comparative analyses between elite Caucasian and Kenyan athletes have also underscored the significance of RE in equating performance outcomes despite differences in $\dot{V}O_2max$. [198]

Given the importance of NMC in endurance performance and the myriad factors that can influence it, it becomes imperative to elucidate the specific role of passive stiffness, particularly in the ankle plantar flexors and the triceps surae, to refine training, enhance performance, and mitigate injury risks. Consequently, this study aims to 1) investigate the association between plantar flexor passive stiffness and NMC, and 2) explore the relationship between plantar flexor passive stiffness and running biomechanics.

METHODS

This study formed part of a broader research initiative focusing on measuring passive stiffness in runners. Conducted as a cross-sectional study, it received approval from the Ethical Committee of the Catholic University of Louvain, Belgium (CEHF - N°: B403201523492).

Participants

Participants were recruited on a voluntary basis, without any financial incentives. Before partaking in the study, each individual was thoroughly informed about the research procedures and provided written consent. To be eligible for inclusion, participants had to meet the following criteria: 1) be at least 18 years old, 2) engage in a minimum of two running sessions per week for the past six months, and 3) achieve a maximal aerobic speed (MAS) of at least 10 km/h. Individuals with any cardiovascular, respiratory, or musculoskeletal issues in the six months preceding the study were excluded.

Outcomes and Materials

Multistage Incremental Test

During their initial visit, participants underwent a continuous multistage incremental test conducted on a 200m indoor track. The test commenced at a speed of 8 km h^{-1}, with increments of 1 km h^{-1} every 2 minutes until the participant reached exhaustion. To ensure maximal effort, participants received standardized encouragement throughout the test. The speed of the last fully completed stage, along with the time sustained during the final incomplete stage, were used to calculate the MAS (expressed in km h^{-1})

Submaximal intensity running test

On their second visit, participants underwent a submaximal intensity running test on a motorized instrumented treadmill (HP Cosmos, ARSALIS, Louvain-La-Neuve, Belgium) in a laboratory room set to a constant temperature of 20.2°C and humidity of 37.7%. The test commenced with a 5-minute period of standing to calculate the standing metabolic state, followed by a 5-minute warm-up phase at 50% of the MAS, and a 6-minute phase at 65% of MAS to assess running metabolic demand. Oxygen uptake ($\dot{V}O2$) was continuously monitored using a metabolic card (CPX, Medisoft, Sorinnes, Belgium), calibrated before each session according to the manufacturer's guidelines. The mean $\dot{V}O2$ for the last two minutes of the

standing phase and the last two minutes at 65% of MAS were used to determine standing and running metabolic demands, respectively. The standing metabolic demand was subtracted from the running metabolic demand to calculate the net running metabolic demand, which was then divided by the running speed to calculate running economy (expressed in ml kg^{-1} m^{-1}) and converted to J $kg^{-1}m^{-1}$ to determine the Net Metabolic Cost (NMC) of running. These parameters were recorded using the instrumented treadmill for 20 seconds at 65% of MAS, with a sampling frequency of 1024Hz.

Additionally, running biomechanics, particularly spatiotemporal variables, were measured. These included step length (cm), stride frequency (strides per minute), aerial duration (ms), contact duration (ms), duty factor, stride duration (ms), time to braking-propulsion transition (ms), time to active peak (ms), and time to braking peak (ms). Kinematic variables such as the center of mass (COM) path (cm) and vertical displacement (cm) were also recorded. Several kinetic variables, including leg stiffness (%BW m^{-1}), impact peak force (%BW), loading rate (%BW ms^{-1}), braking peak force (%BW), propulsive peak force (%BW), propulsive impulse (%BW s), and external mechanical work in horizontal, vertical, and both directions (J kg^{-1} m^{-1}), were documented.

Passive stiffness measurement

Passive stiffness of the dominant limb was evaluated in both lying and standing positions before and after a submaximal intensity running test. For measurements in the lying position, participants were placed prone on an examination table with their ankle and foot extending off the table to expose the gastrocnemii and Achilles tendon while being instructed to relax. The standing position was standardized using weight scales to ensure the body mass was equally distributed between both legs as well as between the rear and front of the foot. Passive stiffness of the gastrocnemii and Achilles tendon was measured using a MyotonPro device (Myoton, AS, Tallinn, Estonia). The MyotonPro, a digital handheld device, calculates biomechanical and viscoelastic properties of tissues by generating a single mechanical impulse and recording the tissue's oscillation response. This method has been validated against a rheometer for its accuracy. [20] Measurements were taken at three distinct sites as shown in Figure 27. The first measurement was conducted on the Achilles tendon (AT), 8cm above its distal bone insertion. The second, i.e., the lateral gastrocnemius (LG) was performed at 30% on the proximal end of a segment extending from the superior border of the fibula to the inferior edge of the calcaneus. The third measurement site was on the medial gastrocnemius (MG), located at 30% on the

proximal end of a segment extending from the muscle proximal insertion in the popliteal fossa to the inferior margin of the lateral malleolus.

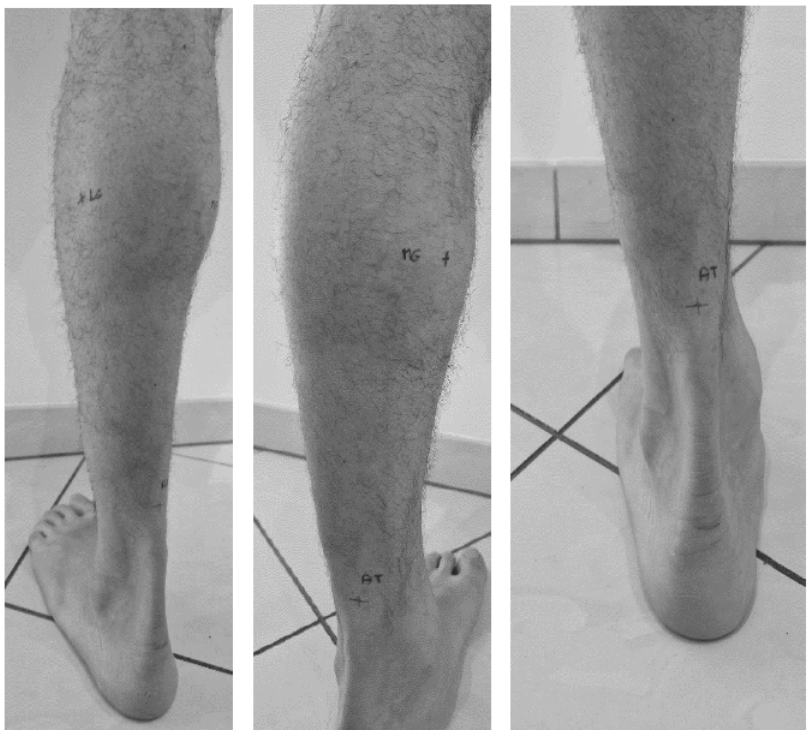

Figure 27. *Pictures of the mapping of the 3 pre marked points on the lateralis gastrocnemius (LG) right, on the medialis gastrocnemius (MG) middle and Achilles tendon (AT) left.*

Statistical analysis

Standard statistical techniques were employed to compute the mean and standard deviation. The Shapiro-Wilk test was utilized to confirm the normal Gaussian distribution, while the modified Levene test assessed homoscedasticity. To evaluate the hypothesis that there was no difference in passive stiffness before and after running, a paired t-test was applied, or a Wilcoxon matched pairs test when necessary. The effect size of any difference was calculated using Cohen's d. The Pearson product-moment correlation was utilized to examine the hypothesis that there is no relationship between passive stiffness or biomechanical parameters and the net metabolic cost of running NMC. A linear regression analysis was performed to assess the strength of the relationship between variables. A p-value of less than 0.05 was considered statistically significant for all tests. These analyses were conducted using R Studio (version 4.2.1).

RESULT

The study recruited twenty-two healthy male volunteers. Descriptive data about the participants is presented in Table 13.

Table 13. Descriptive data of healthy males runners

	mean [sd]	**min - max**
Age (years)	31.3 [12.67]	18 - 60
Body height (cm)	176.7 [5.74]	169 - 192
Body mass (kg)	70.9 [8.86]	57 - 94
Sessions per week (n)	4.45 [2.01]	2 - 10
Maximal aerobic speed (km h^{-1})	17.7 [2.25]	14 - 22.5

Abbreviations: sd: standard deviation, min: minimal value, max: maximal value

Pre- and post-run stiffness values

The paired t-test indicated a small increase in stiffness for the MG and LG, and a moderate increase for the AT between pre- and post-run measurements at rest. Specifically, in the lying position, we observed a small to moderate difference for the AT with an increase of +44.86 N m^{-1}, 95% confidence interval (CI) [8.65 to 81.07], p=0.017, effect size (ES)=0.54; for the MG with an increase of +18 N m^{-1}, 95% CI [14.07 to 31.92], p=0.013, ES=0.32; and for the LG with an increase of +20.72 N m^{-1}, 95% CI [4.47 to 36.97], p=0.014, ES=0.27. No significant differences were found in the standing position for any of the measurement points (i.e., AT, MG, and LG; as detailed in Table 14).

Table 14. Paired t-test between pre- and post-stiffness values for the triceps surae muscle. Data are reported as mean [sd].

	pre-run	post-run	Difference	[95%CI]	ES
	(N m^{-1})	(N m^{-1})	(N m^{-1})		
Lying position					
AT	637.1 [83.5]	682.0 [81.7]	44.86*	[8.6 : 81.0]	0.54
MG	319.4 [47.3]	337.4 [58.1]	18*	[4.0 : 31.9]	0.32
LG	364.6 [72.6]	385.4 [78.7]	20.72*	[4.4 : 36.9]	0.27
Standing position					
AT	1041.8 [160.9]	1011.6 [78.7]	-30.22	[-81.1 : 20.7]	0.18
MG	501.7 [117.8]	503.7 [113.1]	2.05	[-21.7 : 25.8]	0.01
LG	646.3 [122.3]	654.6 [119.9]	8.36	[-7.9 : 24.6]	0.06

*AT: achilles tendon; MG: medial gastrocnemius; LG: lateral gastrocnemius. Alpha significance level was set at 0.05. * : pvalue <0.05, **: pvalue <0.01, ***: pvalue < 0.001. Difference reprensented the difference of the mean, i.e., post minus pre values. Effect size (ES) were estimated using Cohen d with the adjustment methods for sample below 50 participants. A cohen d of 0.2 is considered a small effect size, 0.5 a medium effect size and 0.8 a large effect size.*

Net Metabolic Cost of running and stiffness

The linear regression analysis demonstrated a significant negative correlation between NMC of running and pre-run AT stiffness at rest. Specifically, for every 1 N m^{-1} increase in AT stiffness, there was a 0.0035 J min^{-1} kg^{-1} decrease in the metabolic cost of running ($r = 0.76$, $p < 0.001$). This relationship indicates that the variability in pre-run AT stiffness accounted for 58% of the variability in NMC of running. No significant associations were observed between the metabolic cost of running and the stiffness of the MG or LG, either before or after the exercise. (Figure 28)

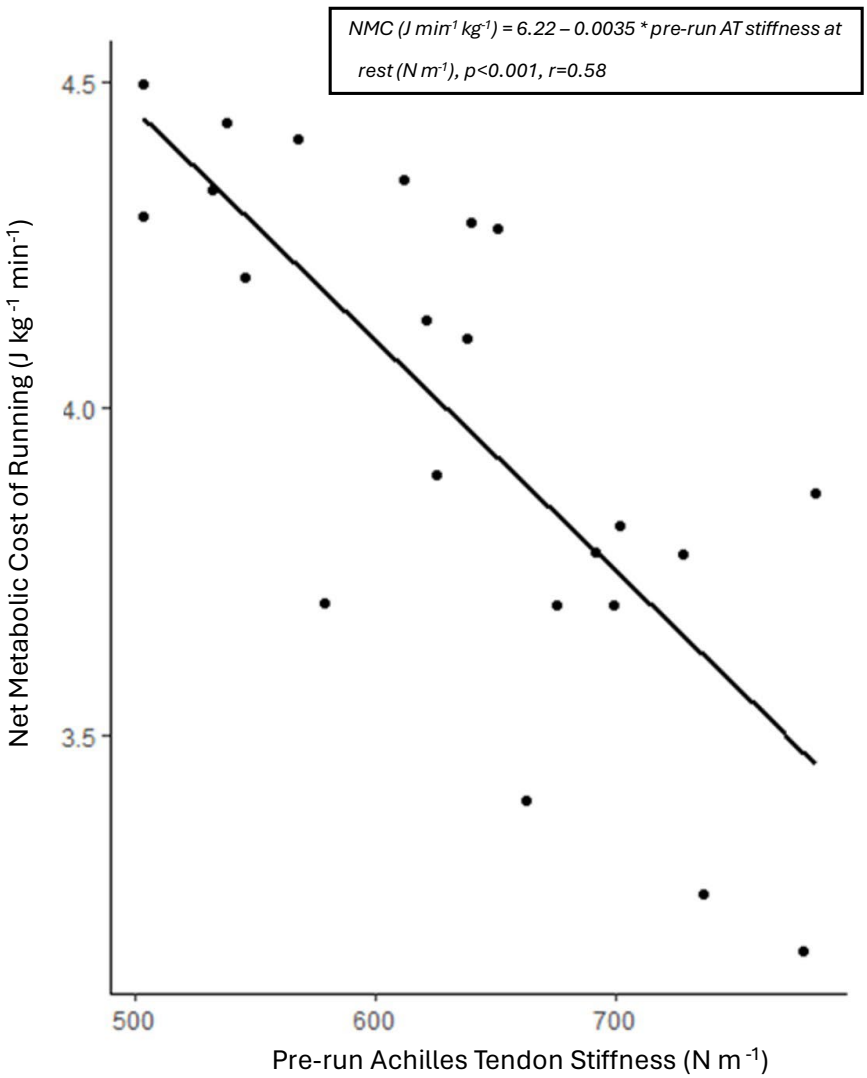

Figure 28. *Linear regression model between the Net Metabolic Cost (NMC) of running and pre-run Achilles tendon (AT) at rest.*

The relationship between pre-run AT passive stiffness, NMC of running, the variation in AT stiffness before and after running, and various running biomechanical parameters is detailed in Table 15. It shows that pre-run AT stiffness is correlated with running biomechanics. Globally, NMC and pre-run AT stiffness are correlated with most of the spatiotemporal and many other kinematic and kinetic variables. This suggests that while AT stiffness before running may have an impact on running biomechanics, the overall efficiency of running, as indicated by NMC, is likely determined by a broader range of biomechanical factors.

Table 15. Pearson correlation coefficient between Net Metabolic Cost of running, Achilles tendon stiffness and running biomechanical parameters

	Net Metabolic Cost	pre-run Achilles tendon stiffness	Difference pre-post
Spatiotemporal variables			
Step length (cm)	0.48*	-0.45*	0.43*
Stride frequency (spm)	-0.61*	0.41	-0.03
Aerial duration (s)	0.65***	-0.64**	0.59*
Contact duration (s)	-0.65***	0.64**	-0.59**
Duty factor	-0.65***	0.64**	-0.59**
Stride duration	0.59**	-0.40	0.01
Time to B-P transition (s)	-0.68***	0.66***	-0.68***
Time to active peak (s)	-0.61**	0.63**	-0.52*
Time to breaking peak (s)	-0.69***	0.72***	-0.58**
Kinematics variables			
COM path (cm)	0.71***	-0.59**	0.28
COM vert displacement (cm)	0.71***	-0.59**	0.27
Kinetics variables			
Leg stiffness (%BW m^{-1})	-0.54**	0.43*	0.01
Impact peak force (%BW)	0.31	-0.25	0.49*
Loading rate (%BW/ms)	0.22	-0.25	0.15
Braking peak force (%BW)	0.47*	0.68**	0.31
Propulsive peak force (%BW)	0.42	-0.44*	0.42*
Propulsive impulse (%BWs)	0.52*	-0.46*	0.26
Mechanical W horizontal (J kg^{-1} m^{-1})	0.21	-0.23	0.21
Mechanical W total (J kg^{-1} m^{-1})	0.56**	-0.46*	0.05
Mechanical W vertical (J kg^{-1} m^{-1})	0.40	-0.28	-0.12

*Significance indication: * = pvalue <0.05, ** = = pvalue <0.01, ***= = pvalue <0.001. Abbreviation: spm: step per minute, B-P= braking to propulsion, COM= Center of Mass, W= work, BW= bodyweight*

DISCUSSION

The primary aim of this study was to explore the relationship between the passive stiffness of the ankle plantar flexor muscles and the NMC of running. Our results indicate a significant negative linear correlation between pre-run Achilles tendon stiffness in a lying position and NMC, accounting for 58% of the NMC variance. However, no linear relationships were found between MG or LG stiffness, biomechanical variables, and NMC.

One potential explanation for the observed correlation is the role of tendon stiffness in energy conservation during running. Tendons act as elastic components, storing potential energy during the stance phase and releasing it during the push-off phase. This stretch-shortening cycle allows the musculotendinous unit (MTU) to undergo lengthening and shortening, returning to its original shape. [199] Variations in the center of mass during running, involving deceleration and acceleration phases, lead to changes in potential energy. The contractile and elastic properties of the MTU, which facilitate energy production and storage, contribute to interindividual differences in running efficiency. [18,40]

Interestingly, a stiffer tendon might store more energy, thus enhancing running efficiency. [46] Athletes have been observed to possess higher Achilles tendon stiffness compared to non-athletes, potentially due to the repetitive mechanical loading from running. [40] Indeed, studies have shown that forefoot striker runners can improve the viscoelastic properties of their Achilles tendons, enhancing energy storage and release capabilities. [92] Tendons are known to adapt to mechanical loading, [200] with plyometric, isometric, and dynamic heavy resistance training shown to increase strength and stiffness[35,47] Such training methods can also improve running performance and NMC. [201–203]

The study also found several biomechanical running variables to be associated with Achilles tendon stiffness. Mechanisms for the observed decrease in oxygen consumption include reduced leg swing inertia, decreased dorsiflexion, and enhanced support from ankle plantar flexors. [204–206] Stiffer shoes have been shown to move the contact point forward during the propulsion phase, increasing the ground reaction force moment while maintaining the Achilles tendon moment. [207] Some studies suggest that certain ground reaction force variables correlate positively with NMC. [208]

Recent interest in running shoes, particularly "supershoe" designs with carbon insoles, has shown that these shoes can improve running economy and contribute to world record improvements across various distances. The potential biomechanical benefits of enhancing Achilles tendon energy yield underline its significance in running performance and injury prevention. However, NMC's multifactorial and complex nature means it cannot be fully explained by a single factor. While Achilles tendon stiffness has a significant impact, other variables such as lower limb force production rate and total active muscle volume should not be overlooked. [209–211]

This study's limitations include a relatively small and young sample size, cautioning against overgeneralizing the findings. Additionally, some participants had not undergone a MAS test within six months prior to the study, potentially affecting NMC measurements' accuracy.

Practical applications

Enhancing the rheological properties of lower limb structures, notably the Achilles tendon, is crucial for runners. As individuals age, passive stiffness properties can decline, with myofascial tissue quality changes affecting the viscosity and viscoelasticity of aging human muscles. [110] Consequently, strengthening exercises are strongly advocated for both health and performance benefits. Furthermore, all types of runners could gain from strengthening training. Plyometric exercises, in particular, have been shown to improve tendon stiffness. [47] Therefore, running stakeholders, including runners, coaches, and healthcare practitioners, are encouraged to incorporate strategies for strengthening or plyometric training into their training programs to adjust tendon stiffness effectively.

Conclusion

The negative correlation between Achilles tendon stiffness and NMC of running, along with the association of several biomechanical parameters with both NMC and Achilles tendon stiffness, highlights the importance of measuring Achilles tendon stiffness in the management of runner performance.

5.3. What is the adaptation of the ankle plantar flexor muscles and tendons before, during and after an incremental running protocol?

Nguyen AP, Bosquet L, Cugnet L, C Detrembleur, P Mahaudens. Observation of the Achilles tendon and Gastrocnemii muscles passive stiffness modification during an incremental running protocol. *In review process, International Journal of Sports Physiology and Performance*

WHAT IS ALREADY KNOWN ON THIS TOPIC

- Previous studies have examined the passive stiffness of the lower limb before and after running trials.
- A stiffer Achilles tendon and triceps surae is associated with improved running performance.

WHAT THIS STUDY ADDED

- This study is the first to observe passive stiffness variations in per running.
- The Achilles tendon and gastrocnemii muscles respond differently to running.
- Passive stiffness of the ankle plantar flexor appears to increase at the beginning of the run and decrease as exhaustion or higher velocities are approached.

HOW THIS STUDY MIGHT AFFECT RESEARCH, PRACTICE OR POLICY

- Variations in passive stiffness in pathological tendons could provide insight for more effective rehabilitation recommendations.
- Future studies should assess the impact of speed, fatigue, or running on potential variations in passive stiffness.

5.3.1. Observation of the Achilles tendon and Gastrocnemii muscles passive stiffness modification during an incremental running protocol

ASTRACT

Objective: Stiffness, a biomechanical parameter, exerts influence on running economy and thus plays a pivotal role in performance. This study aimed to quantify passive stiffness of the Achilles tendon and gastrocnemius muscle using myotonometry throughout an incremental running protocol.

Method: Twenty-one male participants underwent a multistage incremental test on a treadmill until exhaustion. Passive stiffness was measured on the gastrocnemius muscles and Achilles tendon using MyotonPro device. Measurements were taken before, during, and after the incremental test.

Results: Our findings indicate that passive stiffness of the medial gastrocnemius tends to decrease at rest between pre- and post-run assessments (-20.12 Nm^{-1}, p= 0.012). Furthermore, during the test, stiffness significantly increased at low intensity (at 50% of maximal aerobic speed: +104.8 Nm^{-1}, p= 0.042), returning to baseline values as intensity increased.

Conclusion: Throughout an incremental running protocol, both Achilles tendon and gastrocnemius muscle stiffness initially increased and then decreased with escalating percentages of maximal aerobic speed. These observations offer novel insights into the behavior of the triceps surae under loading conditions, encompassing pre-, per-, and post-run phases.

Keywords: stiffness, tendon, muscle, running, myotonometry

INTRODUCTION

Improving running performance is a shared objective among long-distance runners and coaches. Past research has demonstrated that running economy, a critical element in performance, is affected by a range of biomechanical parameters including spatiotemporal parameters, kinematics, kinetics, and neuromuscular factors. [194,212,213] Given that running can be understood, in part, through a spring-mass model [33], stiffness emerges as a fundamental consideration for athletes and coaches. [214]

In mechanical engineering, stiffness (k) is defined as the ratio between the stress applied to a structure and the deformation induced (according to Hooke's Law). In the context of running, stiffness is commonly referred to as leg stiffness (Kleg) or vertical stiffness (Kvert) when considering the entire limb. Both Kleg and Kvert are crucial for understanding the store-recoil mechanisms involved in running [46] and jumping. [215]

Furthermore, as noted by Feng et al. (2018), stiffness is also a characteristic observed in smaller structures such as muscles. [216] Musculoskeletal researchers often utilize the Hill model to elucidate musculotendinous stiffness. [6] This model incorporates a contractile component, represented as a damper, along with two elastic components modelled as springs placed in series and parallel to the contractile component. The contractile component generates active force through actin and myosin cross-bridges, while the series-elastic component and parallel elastic component are linked to the connective tissues and tendon, respectively. The implication of these components without any voluntary contraction represents passive stiffness.

The passive stiffness of the triceps surae (TS), particularly the Achilles tendon (AT), has been extensively examined due to its significant role in propulsion during running and its crucial contribution to the stretch-shortening cycle. [217] Previous research has established a positive correlation between TS stiffness and Kleg in middle-distance runners. [218] Additionally, a stiffer TS muscle has been linked to improved running economy in multiple studies. [41,219]

Numerous investigations have explored the impact of running on AT stiffness to better comprehend muscle adaptation. [40–42,220] One study indicated that neither a steady-speed run nor 60-minute run altered AT properties. [44] However, another study found that following a 90-minute run at lactate threshold, AT stiffness decreased. [45] Various methodologies or instruments, such as ultrasonography[42] , shear wave elastography[221], and the free oscillation

technique [41], can be employed to assess passive stiffness. These methods enable measurements of passive stiffness both before and after effort.

Currently, the exclusive method to observe and measure stiffness during running is through global measures such as Kleg and Kvert, typically recorded on a treadmill with force platforms. Although dynamic stiffness is more accessible to measure during running, the passive stiffness of the gastrocnemius muscle and the AT remains unexplored. Myotonometry presents as a clinical and user-friendly device for measuring passive stiffness, with demonstrated reliability and validity [20]. Recent studies have even shown a correlation between stiffness measured on the belly of the medial gastrocnemius muscle and the AT with the Young's modulus measured by shear wave elastography. [216]

Therefore, the objective of this study was to characterize the passive stiffness of the triceps surae (TS), utilizing myotonometry, during a multistage incremental test performed on a treadmill. Based on existing literature, we hypothesized that plantar flexor stiffness should increase during the test.

METHODS

This study was carried out as an observational cross-sectional study. It was approved by the ethic committee of the Catholic University of Louvain in Belgium. (CEHF - N°: B403201523492).

Participants

The participants were recruited on a voluntary basis and did not receive any financial compensation for their involvement. Prior to the study, each participant received comprehensive information about the experiment and provided written consent.

Inclusion criteria comprised individuals who: 1) engaged in at least two running sessions per week, and 2) had been practicing running for a minimum of one year. Exclusion criteria included individuals who: 1) had sustained a lower limb injury within the past six months, 2) had cardiovascular or neurological conditions, and 3) were classified as elite athletes. Elite athletes were defined as individuals either on the national team or actively participating in international competitions.

Tests and measures

Multistage incremental test

Participants underwent a multistage incremental test on a motorized treadmill (HP Cosmos) within a laboratory room maintained at a constant temperature of 20.2°C and humidity level of 37.7%. The protocol comprised 3-minute stages interspersed with 1-minute recovery periods [229]. The test commenced at a running speed of 8 km h^{-1} and increased by 2 km h^{-1} increments until reaching 12 km h^{-1}, after which the speed incremented by 1 km h^{-1} until exhaustion. Participants were encouraged to exert maximum effort, following standardized instructions. The velocity attained during the last completed stage and the duration maintained during the final incomplete stage, if applicable, were utilized to calculate MAS (in km h^{-1}).

Myotonometry tests

The mechanical properties of the Achilles tendon (AT), medial gastrocnemius (MG), and lateral gastrocnemius (LG) were evaluated using the MyotonPro device (Myoton AS, Tallinn, Estonia). This portable device, when applied perpendicular to the skin, generates a sequence of damped impulses that induce soft tissue deformation. The probe of the device was positioned on the selected sites. Initially, a force of 0.18 N was applied, followed by 5 rapid mechanical impulses of 0.4 N each (15 ms duration per impulse). Subsequently, the device recorded the resulting oscillations in the form of an acceleration signal. The mean values obtained provided quantification of five biomechanical parameters. Among these parameters was passive stiffness (N m^{-1}), estimated using the formula: $S = a_{max}\ m\ l^{-1}$, where "a" represents the acceleration of the probe, "m" denotes the mass of the probe, and "l" signifies the displacement of the tissue.



Figure 29. *Location of the 6 sites used for stiffness measurements. MG: medial gastrocnemius, LG: lateral gastrocnemius, 2-4-6-8 respectively represent 2-4-6-8 cm above the tendon insertion.*

This measurement technique has been validated through comparison with rheometer. [20] Stiffness was assessed in both resting and standing positions by the same examiner. Six specific sites were marked on the dominant leg: four on the Achilles tendon (at 2, 4, 6, and 8 cm from the tendon insertion) and one on each gastrocnemius muscle (located at 30% of the length from the head of the fibula to the heel for the LG, and at 30% of the leg length from the popliteal fossa to the lateral malleolus for the MG (Figure 29). It's worth noting that Chang et al. (2020) demonstrated that stiffness measurements were not affected by whether they were taken on the dominant or non-dominant leg. [222]

Measurements were conducted as follows: before the running protocol at all six sites, during each one-minute break at three sites (AT 8 cm, and both gastrocnemius marks), and immediately after the test at the initial six sites.

In the resting position, participants lay prone on a treatment table with knees extended and feet hanging off the table. For the standing position, participants stood on two scales with their

dominant leg to distribute the weight evenly on the limb, while the non-dominant leg rested on a plinth at the same height as the scales. Participants kept their shoes on due to time constraints, and they were asked to place their forefoot on the first scale and their rearfoot on the second scale, applying a pressure equivalent to a quarter of their body weight to each scale while stiffness measurements were taken using the MyotonPro device (Figure 30). One operator performed stiffness measurements while the other controlled the weight on the scales. During the one-minute break, participants were instructed to transition as swiftly as possible from the treadmill to the treatment table and then to the scales.

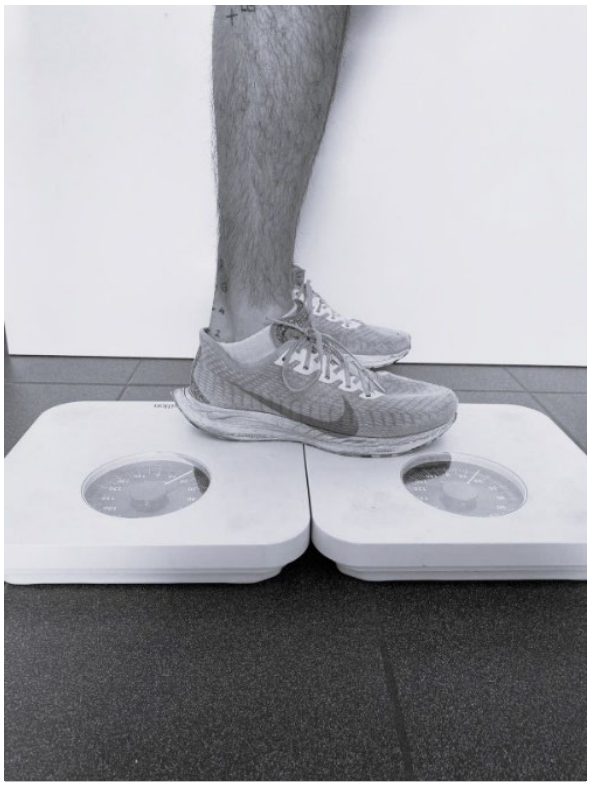

Figure 30. *Setting for the standing position.*

Statistical analysis

The descriptive statistics were presented as mean ± standard deviation. The normality of the distribution was verified using the Shapiro-Wilk test, while homoscedasticity was checked with the modified Levene's test. Differences in pre- and post-run stiffness were assessed using a paired t-test to test the null hypothesis that there is no difference between these two states. A one-way ANOVA with repeated measures, followed by Holm-Sidak post-hoc tests, was

employed to evaluate the null hypothesis of no differences in stiffness across different stages of the incremental test. The effect size was calculated using Cohen's d.

A p-value of less than 0.05 was considered statistically significant for all analyses, which were conducted using MedCalc® Statistical Software version 20.104 (MedCalc Software Ltd, Ostend, Belgium).

RESULTS

Twenty-one healthy male runners participated in the study. Table 16 outlines their demographics and other pertinent details. Each participant successfully completed the multistage incremental test without experiencing any negative effects. The average length of the test was 24.4 ± 5.91 minutes, which equated to a mean MAS of 17.8 ± 1.33 km h^{-1}.

Table 16. Descriptive data of sample (n=21)

Age (years)	27.9 ± 6.47
Height (cm)	179 ± 7.4
Weight (kg)	79.9 ± 7.25
Sport's session per week (n)	3 ± 1
Maximal aerobic speed (km h^{-1})	17.8 ± 1.33

Data are present as mean ± standard deviation

Stiffness values pre- and post- running

No significant differences were found in the pre- and post-run stiffness measurements of the AT in lying position (Table 17). However, a small decrease in stiffness was observed in the medial gastrocnemius (-20.1 N m^{-1}, 95% Confidence Interval [-35.3 to -4.9], p= 0.012, Effect Size=0.42). In the standing condition, there was a moderate decrease in stiffness observed post-run in T6 (-127.5 N m^{-1}, 95% Confidence Interval [-182.7 to -72.3], p<0.001, Effect Size=0.54) and in the LG (-57.3 N m^{-1}, 95% Confidence Interval [-83.9 to -30.7], p<0.001, Effect Size=0.65). The difference between the rest (lying position) and standing conditions was altered in T4, T6, and LG (Table 17).

During the running protocol, stiffness in the standing position remained unchanged for both the Achilles tendon (T8) and the gastrocnemii (medialis or lateralis). Post-hoc Holm-Sidak

analysis revealed a significant increase in stiffness in the lying position at 50% of MAS for the Achilles tendon (T8) compared to the baseline (+104.8 N m^{-1}, p= 0.042). The MG showed increased stiffness in lying position at both 50% (p= 0.049) and 65% (p= 0.32) of MAS compared to the baseline, with increases of 65.6 N m^{-1} and 71.3 N m^{-1}, respectively. Finally, the stiffness in lying position of the LG showed increased stiffness at 50% (+61.3 N m^{-1}, p= 0.042), 65% (+72.2 N m^{-1}, p= 0.012), and 70% (+75.5 N m^{-1}, p= 0.006) of MAS, compared to the baseline value (Figure 31).

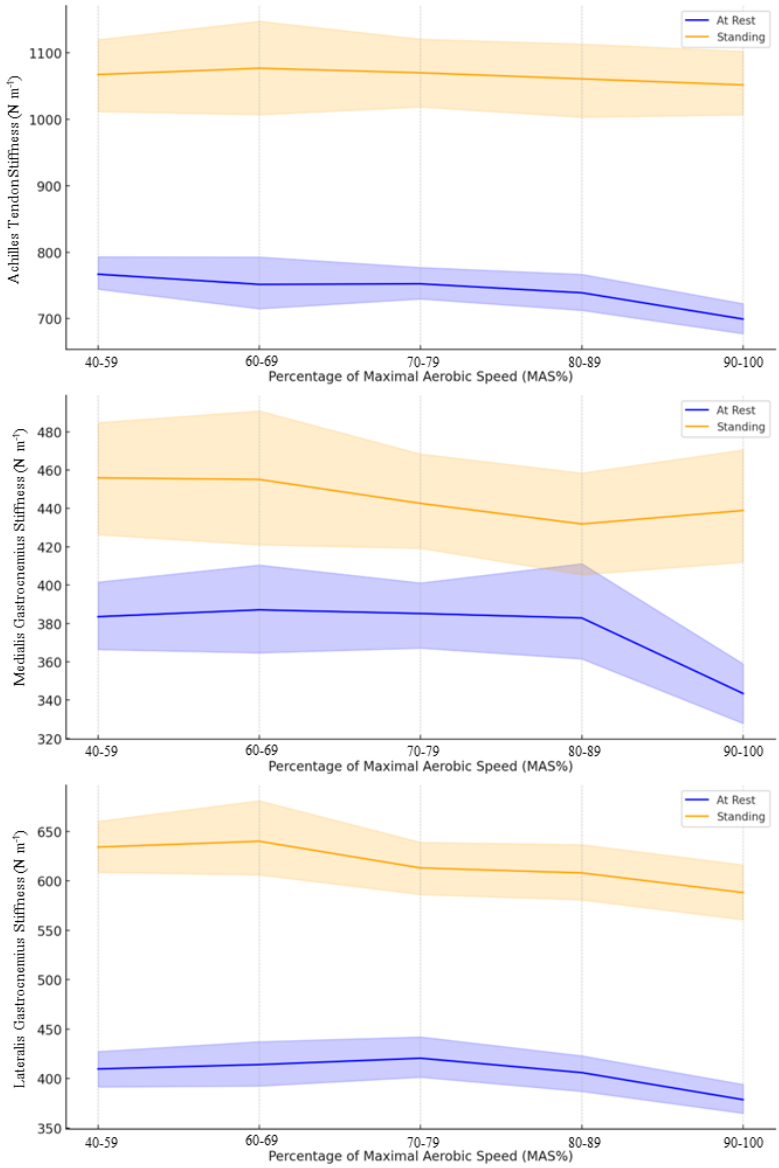

Figure 31. *Evolution of stiffness of the Achilles Tendon (T8), and the belly of the gastrocnemius medialis and lateralis at rest and standing. X-axis represented the percentage of maximal aerobic speed and y-axis represented the stiffness of the point of interest expressed in N* m^{-1}.

Table 17. Pre- and post-test stiffness values for the triceps surae muscle

		pre-run ($N\ m^{-1}$)		post-run ($N\ m^{-1}$)		Difference ($N\ m^{-1}$)		magnitude of difference (cohen d)
		mean	sd	mean	sd	mean	95%CI	
At rest	**T4**	775.4	84.45	797.5	60.14	22.1	-14.87 to 59.08	0.28
	T6	735.8	68.78	733.5	75.28	-2.35	-34.53 to 29.83	0.03
	T8	684.7	70.54	666.5	67.55	-18.25	-45.97 to 9.46	0.24
	MG	327.9	42.03	310.4	36.35	-17.55*	-31.38 to -3.71	0.41
	LG	365.1	45.06	353.7	41.81	-11.4	-30.04 to 7.21	0.24
Standing	**T4**	1281.6	193.3	1240.4	197.5	-41.2	-103.4 to 21.09	0.19
	T6	1228.9	208.7	1101.4	221.7	-127.5***	-182.7 to -72.25	0.54
	T8	1084.7	193.9	1017.8	172.5	-66.85	-134.1 to 0.41	0.33
	MG	477.1	102.4	448.8	119.9	-28.3	-71.62 to 15.02	0.23
	LG	632.2	78.63	574.9	83.42	-57.3***	-83.9 to -30.68	0.65

Table 17 (continued)

Delta	**T4**	506.2	218.7	442.9	199.4	-63.3*	-126.5 to -0.01	0.28
	T6	493.1	216.5	367.9	174.8	-125.1***	-182.5 to -67.73	0.58
	T8	399.9	200.3	351.3	147.5	-48.6	-116.7 to 19.59	0.26
	MG	179.2	95.21	138.4	111.1	-10.75	-53.35 to 31.85	0.1
	LG	267.2	68.31	221.3	71.4	-45.9***	-68.05 to -23.74	0.6

Alpha significance level was set at 0.05. * Indicated a pvalue < 0.05, ** < 0.01, *** < 0.001. Tx represents the stiffness measurement point at x centimeter from the distal insertion. MG: gastrocnemius medialis, LG = gastrocnemius lateralis. Difference represented the difference of the mean, i.e., pre minus post values. Effect size were estimated using Cohen d with the adjustment methods for sample below 50 participants. A Cohen d of 0.2 is considered a small effect size, 0.5 a moderate effect size and 0.8 a large effect size.

DISCUSSION

The primary aim of this research was to investigate how passive AT and gastrocnemius muscle, i.e., MG and LG, stiffness vary during an incremental running protocol up to the point of exhaustion. This study is possibly the first of its kind to examine changes in passive stiffness during exercise. The results indicate that while the passive stiffness values of AT and LG remained relatively constant from before to after the test at rest, there was a noticeable decrease in medial gastrocnemius (MG) stiffness following the running protocol. Interestingly, stiffness measurements taken before running displayed minimal variation but tended to decrease towards the end of the protocol.

One of the novel insights from this research is the observed pattern in stiffness changes for AT, LG, and MG at rest—increasing at lower speeds and then decreasing as speed increased. This behavior diverges from the traditional spring model often applied to tendon and muscle components, suggesting that their viscoelastic properties might differ from a simple spring mechanism. The initial increase in AT stiffness at lower speeds might be attributed to the acute effects of a warm-up, aligning with findings from Pieters et al. (2022), who noted a significant stiffness increase after plyometric exercises or running at 2.7 m s^{-1} in a study involving 40 participants. [223]

This phenomenon might be explained by changes in the hydration state of the AT. It's been hypothesized that mechanical stress causes fluid movement within tendons, which are composed of collagen and water. [49,189] A systematic review and meta-analysis conducted in 2021 found no significant changes in AT volume following running, suggesting that fluid may redistribute within the tendon rather than being expelled. [224,225] This ability of the tendon to retain water might lead to temporary stiffening in response to mechanical loads. [189]

Moreover, the study discusses "tendon fatigue," defined as changes in tendon length and stiffness at a given force. [226] Although tendon length wasn't directly measured in this study, the observed decrease in AT passive stiffness at rest as the running protocol progressed could be indicative of fatigue resulting from increased intensity. Fatigue has been identified as a factor that influences overall stiffness and spatiotemporal gait characteristics.[227] A study by García-Pinillos et al. (2020) involving twenty-two trained runners who underwent a 60-minute run revealed a decrease in leg stiffness (Kleg) under fatigue conditions (towards the end of the run).[228] However, this result contrasts with findings from other studies where it was reported

that Kleg or AT stiffness did not change after a run at constant velocity.[226,229] Although changes in tendon length were observed, the stiffness remained constant, highlighting the influence of mechanical creep rather than overall stiffness changes.

The current study explored the impact of three specific features: prolonged duration, incremental intensity, and exhaustive effort in a running protocol. The findings indicated that none of these features, even when combined, led to changes in AT stiffness post-run compared to baseline measurements. However, the unique mechanisms underlying each factor separately could help explain the observed results during the test. Most existing research on AT behavior during running typically involves protocols longer than 30 minutes. The shorter but more intense protocol in this study likely induced fatigue at both physiological and biomechanical levels in the participants.

Regarding the influence of different running speeds on stiffness, research findings have been mixed, possibly due to variations in the methodologies for calculating leg spring stiffness.[18] For instance, Morin et al. (2005) conducted 30-second running bouts with eight healthy men, gradually increasing pace from 3.33 m s^{-1} to 6.67 m s^{-1}.[192] They observed an increase in Kvert without any change in Kleg.[192] Conversely, Rabita et al. (2011) found a decrease in leg stiffness following a high-speed exhaustive run. [227] Notably, no previous study has specifically focused on the effect of different running speeds on the stiffness of the AT or gastrocnemii muscles making this research a pioneering effort to understand how passive stiffness may vary with increasing velocity.

The measurement of stiffness involves a variety of instruments and protocols, leading to challenges in comparing results and precisely determining how AT, MG or LG stiffness respond to loading, particularly during running activities. The behavior of passive stiffness under various conditions warrants further investigation, as existing studies on the subject present conflicting outcomes due to the diversity of methodologies employed. Thus, careful consideration is required when comparing findings across different studies. Global stiffness measures, such as Kleg and Kvert, are relatively easier to measure both experimentally and in field conditions. Contrary to our findings, Rogers et al. identified a strong relationship between Kleg and AT stiffness during both maximal and submaximal running, though the proportion of Kleg variance explained by AT stiffness was not precisely determined. [218]This highlights the need for further research to deepen our understanding of the impact of passive and global stiffness on running performance and mechanics.

The limitations of this study include a relatively small sample size, though the estimated power for the one-way ANOVA was 0.7, suggesting a reasonable level of statistical power. Additionally, the intended 1-minute break between 3-minute running intervals, as outlined in the protocol, was exceeded by approximately 1 minute and 20 seconds.

A deeper understanding of the functioning of the tendon is essential for guiding coaches and athletes in their preparation and achievement of goals. Strengthening exercises are known to improve both running performance and tendon stiffness. However, investigating the optimal Achilles tendon stiffness or stiffness variation during running could help identify individuals who respond to specific interventions. For instance, Hebert-Losier and colleagues suggested that runners respond differently to improvements from "super shoes" in terms of performance. Identifying subgroups of runners who may or may not benefit from certain running interventions could enhance both performance optimization and rehabilitation recommendations. [230]

In summary, our study revealed that during an incremental running protocol, both the Achilles tendon and Gastrocnemii muscle stiffness initially increased and then decreased as the percentage of maximal aerobic speed increased, which corresponds to higher speeds. Additionally, post-run passive stiffness measurements of the Achilles tendon and lateral Gastrocnemius at rest did not exhibit changes from baseline values, whereas the medial Gastrocnemius stiffness showed a decrease. These findings contribute new insights into the adaptation of the Achilles tendon and Gastrocnemius muscle under loading conditions. Although further research is essential to explore the underlying mechanisms of these observations, understanding the rheology of musculotendinous structures during running could provide valuable normative data for rehabilitating pathological conditions or enhancing running performance.

5.4. What Changes Occur in the Passive Stiffness of Ankle Plantar Flexor Muscles and Plantar Fascia During a 10k Run?

AP Nguyen, C Detrembleur, P Mahaudens. Lower leg stiffness changes during a 10k run. *In submission process in Journal of Sport Biomechanics*

WHAT IS ALREADY KNOWN ON THIS TOPIC

- High lower limb passive stiffness is associated with improved running performance.
- Traditional methods such as elastography and sinusoidal oscillatory techniques for measuring passive stiffness are often impractical outside laboratory settings.

WHAT THIS STUDY ADDED

- Myotonometry has confirmed the potential for real-time measurement of lower limb stiffness in runners during a 10 km race.
- The stiffness of plantar fascia significantly decreases during a 10 km run, in contrast to the constant stiffness observed in Achilles tendon, patellar tendon, and gastrocnemius medialis.

HOW THIS STUDY MIGHT AFFECT RESEARCH, PRACTICE OR POLICY

- Real-time measurement is feasible during on-field running events.
- Variations in the passive stiffness of the plantar fascia could offer new insight for rehabilitation.

5.4.1. Lower leg stiffness changes during a 10k run.

ASTRACT

Introduction: The primary aim of this study was to assess lower limb stiffness in healthy runners over the course of a 10 km race at a competitive pace, with a secondary focus on investigating the tissue-specific changes in stiffness of lower limb structures during the running event.

Methodology: Twenty participants (3 women and 17 men) were recruited for this study. Passive stiffness measurements were assessed on the Achilles tendon, medial gastrocnemius, patellar tendon, and plantar fascia using the MyotonPro device. Measurements were recorded before, during, and after the running activity.

Results: Our findings reveal a significant decrease in the passive stiffness of the plantar fascia, while stiffness levels remain constant for the Achilles tendon, patellar tendon, and medial gastrocnemius throughout the duration of the race.

Conclusion: Contrary to our initial hypothesis, which predicted a decrease in stiffness during the race, our study findings indicate that the stiffness of the Achilles tendon, patellar tendon, and medial gastrocnemius remained constant. However, we observed a noteworthy decrease in plantar fascia stiffness, potentially attributable to fatigue. Further investigation is warranted to elucidate the mechanisms behind the differential response of the plantar fascia compared to other tissues during prolonged running activity.

Introduction

Running has evolved into an increasingly popular global sport, capturing the attention of the running community as well as clinicians and researchers, all of whom are focused on boosting performance and preventing injuries.

While running offers numerous health benefits, [231] its practice has also been associated with a variety of running-related injuries. [232,233] The biomechanical characteristics of running are closely scrutinized by scientists and clinicians with the aim of preventing or treating these injuries, [234] even though these factors may not always be directly useful in preventing, explaining, or treating injuries in runners. On the other hand, running biomechanical metrics appear to hold considerable interest for enhancing running performance. [212]

Passive stiffness refers to the inherent resistance of a tissue to deformation while at rest. In the context of running, high passive stiffness can confer certain benefits. Indeed, several studies have highlighted the correlation between lower limb passive stiffness and running performance. [18,41,235] Bohm et al., for instance, elucidated that tendons serve not only as energy reservoirs but also facilitate optimal shortening of muscle contractile elements during tendon elongation. [200]

Lower limb passive stiffness plays a pivotal role in two fundamental phenomena: power amplification and power attenuation. The former denotes the limb's capacity to generate force surpassing that produced by the muscle alone, owing to the elastic energy stored within the tendon. Conversely, power attenuation enables the tendon to elongate and accumulate energy during rapid movements, thereby preventing excessive muscle lengthening while still generating an appropriate force output. [236]

To assess the passive stiffness of musculoskeletal structures, various methodologies have been explored, including elastography and the sinusoidal oscillatory method. [110] However, these approaches can be costly or impractical for field use. Alternatively, ultrasound presents a viable option for examining tendon stiffness. Utilizing shear wave elastography, which applies Young's model to evaluate tissue stiffness by measuring wave propagation speed, has shown promise in this regard. [137,216]

Additionally, myotonometry offers a convenient and field-applicable method for assessing stiffness in subcutaneous tissue. This technique has been demonstrated to be reliable in stiffness measurements when compared to gold standard rheometer. [20] With its accessibility and ease of use, myotonometry presents a valuable tool for researchers and clinicians alike in evaluating musculoskeletal stiffness. [153]

With the introduction of these more suitable technologies in the field, researchers can now observe the rheological behaviors of tissues in vivo and during sporting events. This capability presents an opportunity to gain insights into the biomechanical characteristics of runners' tissues, not only for enhancing performance but also for injury prevention.

The objectives of this study were twofold: first, to investigate the change in lower limb stiffness in healthy runners during a 10 km race at a competitive pace, and second, to examine the tissue-specific changes in stiffness of lower limb structures, including the patellar tendon (PT), Achilles tendon (AT), the medialis gastrocnemius (MG), and the plantar fascia (PF), throughout the duration of the run.

Methods

Study design

The study was conducted by the UCLouvain in Louvain-la-Neuve and the Ethics Committee gave a favorable opinion on the protocol (V2 DD 10/09/2019).

Population

Twenty participants were enrolled in the study. The inclusion criteria were had to be aged 18 or above and have a minimum of one year of regular running experience. They were required to demonstrate the ability to run a distance of 15 kilometers per week and be familiar with their maximum aerobic speed (MAS). Exclusion criteria included any history of lower limb musculoskeletal pathologies within the past 12 months or neurological pathologies. Participants failing to meet these criteria were ineligible for inclusion in the experiment.

Procedure

Before the race day, demographic information including age, sex, weight, height, medical history, frequency and volume of sports, running experience, previous records, footwear preference, and MAS were collected. For those participants who were unaware of their MAS, a VAM-EVAL test was performed at least one week prior to the test day. Upon arrival, participants were briefed on the study's objectives and provided written consent. The experiment duration was approximately 90 minutes.,

During the session, participants completed a 10 km run on a 200 m indoor track, during which passive stiffness measurements were taken. Measurements of the AT, MG, and PT were conducted at the start and at each kilometer until the end, i.e., the 10th kilometre with running footwear.

PF Measurements were performed at the baseline, at the 5 km mark, and at the 10 km marks, without running footwear. Participants initiated their run at a pace between 75% and 85% of their MAS after baseline assessments. Lap times were recorded, and runners received at every 500 m interval to maintain a constant running pace.

Material

Stiffness measurements were conducted using the MyotonPro device (Myoton AS, Tallinn, Estonia), a portable and non-invasive tool designed to assess the biomechanical properties of soft tissues. This device utilizes mechanical deformation to evaluate the stiffness (in N m^{-1}) of muscles and tendons in the lower limbs.

During the measurements, the probe of the MyotonPro device was positioned perpendicular to the skin surface overlying the muscle or tendon of interest. A light pressure, generated by a short mechanical pulse (0.4N for 15ms) with a constant pre-compression force of 0.18N, was applied. This mechanical pulse induced damped oscillations in the soft tissue, which were recorded by an accelerometer. Subsequently, the acceleration signal was processed to generate an oscillation curve, from which the stiffness was calculated. Measurements using the MyotonPro device were conducted both before and during the test to assess changes in tissue stiffness over time.

Measurement

All measurement points were anatomically identified. The Achilles tendon stiffness was recorded 8 cm above its insertion at the calcaneus. The stiffness of the MG was measured at the belly of the muscle. For the PT, stiffness was assessed at the midpoint between the patella and the anterior tibial tuberosity. As for the PF, stiffness measurement was performed at its proximal insertion on the calcaneus. During measurements of the AT, MG and PF, the participants were positioned in a prone position on an examination table, with their feet placed outside. Conversely, for measurements of the PT, the participants were placed in a supine position on the examination table, with a cushion positioned under the popliteal fossa (Figure 32).

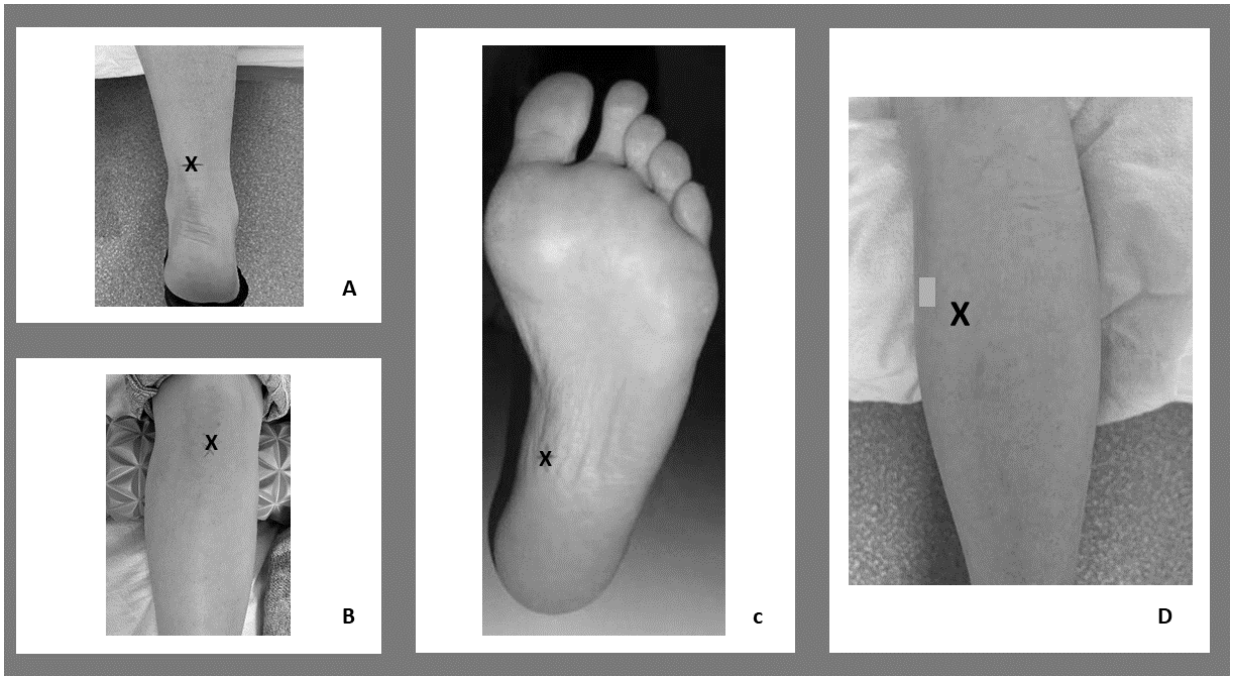

Figure 32. *Location (X) of stiffness measurement in the Achilles tendon (A), the patellar tendon (B), the plantar fascia (C) and on the gastrocnemius medialis (D).*

Statistical analysis

Statistical analyses were conducted using Medcalc software. Normality of the data was assessed using the Shapiro Wilk test. A one-way repeated measured analysis of variance (ANOVA) was carried out to evaluate the time effect at every kilometer interval on each stiffness measurement. Where necessary, Bonferroni post-hoc analysis was applied. The significance level was set at alpha= 0.05

Results

In this study, twenty healthy participants were recruited as subjects. All relevant demographic information for these participants is provided in Table 18.

Table 18. descriptive data of participants.

Total (n)	20
Men/women (n)	18/4
Age (years)	26.5 (8.62)
Height (cm)	178 (8.6)
Weight (kg)	69.9 (12.52)
Km per week	25.4 (8.05)
Session per week	3 (1)
Running experience (years)	7.8 (9.43)
Maximal aerobic speed (km h^{-1})	16.6 (1.62])

data are expressed in mean (standard deviation)

Stiffness modification

The RM-ANOVA analysis conducted on the AT measurements showed no statistically significant differences among the values collected at various kilometers ($p = 0.959$). Regarding the PF, the post-hoc analysis indicated a significant decrease in stiffness between the 5th and 10th kilometers compared to the initial measurement ($p = 0.008$). However, no significant modifications in stiffness were observed throughout the run for the PT and MG measurements (Figure 33).

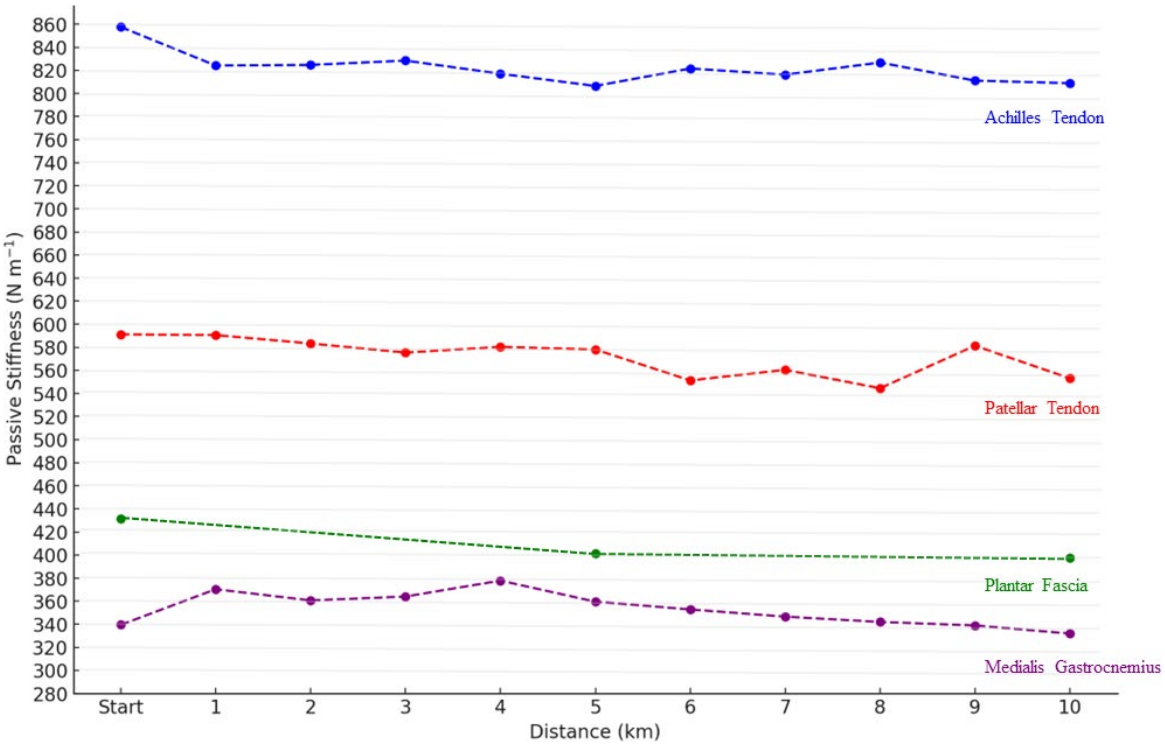

Figure 33. *Evolution of lower limb stiffness measurements (in N m^{-1}) throughout the 10k run, with significant differences indicated by white dots compared to baseline (i.e., plantar fascia measurement at the 5th and 10th kilometer)*

Discussion

The primary objective of this study was to observe lower limb stiffness in healthy runners throughout a 10 km race at a competitive pace. By conducting measurements at each kilometer's interval, we were able to capture the continuous evolution of stiffness throughout the race, providing insights beyond just the initial and final stages. Additionally, a secondary objective was to investigate the change in stiffness of the various structures in lower limb.

Our findings revealed a noteworthy decrease in passive stiffness observed specifically in the plantar fascia measurement. This significant decrease in stiffness highlights the dynamic changes occurring in lower limb biomechanics during prolonged running activity. The study by Houghton et al. (2013) reported that a 60-minute treadmill run did not induce changes in the stiffness of the AT, but rather affected force/strength. Stiffness was calculated from the force-displacement curve obtained using a dynamometer and b-mode ultrasound device. [44] Similarly, Peltonen et al. (2012) found no influence of a marathon on AT stiffness when measured one hour post-race, using the force elongation curve to determine stiffness. [237] These findings suggest that the load experienced by the AT during these activities may not be sufficient to overstress it or predispose it to injuries. These conclusions align with those of Farris, who observed no changes in AT stiffness after a 30-minute steady-state speed running session on a treadmill, contrary to initial hypotheses. [238] The non-modification of AT stiffness in our study may imply that the threshold point required to induce changes in mechanical properties was not surpassed in terms of strain magnitude. [18]

Conversely, other studies have shown a decrease in passive stiffness of the AT following strenuous activities. Ooi et al. (2015) demonstrated a significant reduction in Achilles tendon stiffness after a marathon, employing elastography. [239] Similarly, Fletcher et al. (2010) observed a decrease in AT stiffness after a 90-minute run at 85% of lactate threshold, associated with changes in running economy. [45] This reduction in stiffness was attributed to greater elongation while maximum voluntary contraction (MVC) remained constant.

Our hypothesis regarding the decrease in passive stiffness of the gastrocnemius during the run was not validated. This led us to consider the possibility that peripheral fatigability induced by such running activities could alter myofascial viscoelasticity. While peripheral fatigue has been extensively studied in isolated muscle contractions [240] (eccentric, concentric, and isometric),

the combination of stretch and shortening cycle (SSC) contractions in ankle plantar flexor muscles during running presents a unique scenario. Consequently, the responses of ankle plantar flexor muscles to fatigue may differ from those of other muscle groups. [241]

Intense SSC activities, such as marathon running, have been associated with a decrease in muscle stiffness and a concurrent decrease in the sensitivity of the stretch reflex due to fatigue. [242] Furthermore, intense SSC efforts may induce muscle damage, which can influence the SSC itself. This decreased SSC, coupled with muscle damage, could result from changes in input from muscle spindle III and IV afferents, as well as Golgi tendon organs, all of which impact stiffness. [243] Although the impact of this phenomenon in running remains unclear, the titin protein, a determinant of passive muscle stiffness, is known to be sensitive to eccentric contractions. [244]

Fletcher and MacIntosh (2017) have demonstrated that stiffness influences the rate and extent of muscle fascicle shortening, potentially affecting strength and running economy. [245] However, studies analyzing the behavior of passive stiffness before, during, and after running are scarce. This gap in the literature warrants further investigation to better understand the complex interplay between muscle fatigue, stiffness, and running performance.

Viscoelastic systems are known to be influenced by temperature, and running can lead to an increase in body temperature, consequently affecting muscle temperature as well. Noonan et al. (1993) reported that warming (from 25° to 40°) reduced muscle stiffness in the tibial anterior and extensor digitorum of rabbits during passive stretch. [246] However, the effect of temperature on stiffness can vary depending on the type of contraction and loading rate. A recent in vitro study by Bernabei et al. (2020) on cat muscles showed that increasing temperature (from 26° to 28°) reduced active muscle stiffness during active contraction but increased passive short-range stiffness using shear wave ultrasound elastography. [247] These findings highlight the complexity of the relationship between temperature and stiffness, with inconsistent results observed in the literature, likely due to methodological differences. While it is suggested that temperature variation may indeed affect stiffness, further studies are necessary to elucidate the exact mechanism through which heat influences passive stiffness.

On the contrary, we observed a notable decrease in passive PF stiffness as early as the 5th kilometer during the race. This finding aligns with a study by Shiotani et al., who also reported a significant decrease in passive stiffness during a 10 km outdoor race on asphalt. [248] In their study, they noted a difference in stiffness at the conclusion of the ten kilometers. Using shear wave ultrasound elastography as the measurement apparatus, Shiotani's results were deemed reliable, as supported by the study by Feng et al. [216,248]

To explain this decrease in PF stiffness, we concur with the hypothesis proposed by Shiotani et al. [248] that it is attributable to mechanical fatigue and microscopic damage. As for the consistency observed in the stiffness of other tissues, such as the AT, MG and PT, it is plausible that the effort exerted during the race was not intense enough to induce a significant decrease in their stiffness. This suggests that the mechanical fatigue and microscopic damage experienced by the plantar fascia may have been more pronounced compared to other lower limb tissues during the 10 km race. Indeed, these findings underscore the significance of not only assessing AT stiffness but also considering the stiffness of the MG in the evaluation and management of tendinopathy and plantar fasciitis. Understanding the mechanical aspects of tendon stiffness in tendinopathy is crucial for effective management and primary prevention strategies.

In particular, assessing stiffness during running efforts and monitoring changes in stiffness during recovery could provide valuable insights into the progression and resolution of tendinopathy. By incorporating MG stiffness alongside AT stiffness assessment, clinicians can gain a more comprehensive understanding of the mechanical factors contributing to tendon health and dysfunction. This holistic approach can inform tailored treatment plans aimed at optimizing tendon function and mitigating the risk of injury recurrence.

Regarding our methodology, several aspects require improvement or modification for future studies or repetitions. Firstly, the running pace utilized in our study did not consistently induce a state of exhaustion among participants, contrary to our intention of simulating a competitive pace. Unfortunately, we did not assess the perceived intensity of the race at its conclusion using an effort intensity scale. Integrating such an assessment in future studies would be beneficial. To achieve this, we propose increasing the pace from the current range of 75%-85% of MAS to 85%-90%. However, it is important to acknowledge that increasing the pace carries the risk of participants being unable to maintain it, leading to a potential reduction in pace or cessation of the run. To address this concern, a stepwise approach could be adopted, incorporating a

notion of failure and success to progress to the next step. In this framework, failure to maintain the prescribed pace or stopping altogether would signify the failure of the step and the conclusion of measurements for the participant. This approach offers a balanced compromise between achieving the desired intensity of the run and managing participant fatigue and adherence.

The findings of this study do not align with our initial hypothesis, which anticipated a decrease in the stiffness of various tissues during the race. Contrary to expectations, the stiffness of the AT, PT and MG remained constant throughout the race. However, there was a notable decrease in the stiffness of the PF. It is intriguing to explore why only the stiffness of the plantar fascia decreases while that of other tissues remains constant. Possible explanations could include differences in biomechanical demands and loading patterns during running, variations in tissue composition and structure, or differential responses to fatigue among the various tissues. Further investigation into these factors is warranted to better understand the mechanisms underlying tissue-specific changes in stiffness during running activity. Such insights could provide valuable information for optimizing training regimens, injury prevention strategies, and rehabilitation protocols for runners.

6. General Discussion

6.1. Main achievements of the thesis

The investigation into the implications of lower limb peripheral muscle architecture (PMAS) in running presents a real challenge owing to the intricacies involved in precise measurement and subsequent accurate assessments. [20] This thesis pursues three primary objectives: firstly, to advance our comprehension of PMAS dynamics in individuals without known musculoskeletal issues; secondly, to delineate a methodology for measurement, particularly focusing on myotonometry, to ameliorate the constraints encountered during real-world evaluations of runners; and lastly, to conduct a comprehensive evaluation of PMAS in runners across various scenarios, including metabolic testing trials, incremental running trials, and constant velocity runs.[20]

Firstly, we embark on an exhaustive examination of viscoelastic stiffness in humans, utilizing the wrist joint as a paradigmatic illustration. This scrutiny of viscoelastic characteristics yields normative data encompassing a wide demographic spectrum, spanning from young children to the elderly. Such insights hold potential in establishing baseline PMAS values among healthy individuals and facilitating comparisons across diverse athletic or pathological contexts.

Secondly, this thesis delves into a comprehensive elucidation of the foundational principles underpinning myotonometry and its practical applications. Despite the current utilization of this technology and device in research, scant attention has been directed towards conducting thorough investigations to ascertain the precise parameters that can be measured and the pragmatic utility of these measurements. Our findings indicate that while the MyotonPro device is a reliable instrument for estimating elastic stiffness in humans, it falls short in assessing viscoelastic properties, including viscous stiffness. This discrepancy underscores the imperative for further fundamental research in this domain to fully comprehend and exploit the potential of the MyotonPro device across diverse research and clinical settings.

Lastly, this thesis presents a clinical inquiry into the utilization of myotonometry for assessing running biomechanics. By applying the principles of the spring mass damper

(SMD) model, this study seeks to augment our comprehension of how elastic stiffness, particularly the rheological characteristics of lower limb tissues, can impact running mechanics. This approach underscores the utility of evaluating PMAS as a valuable asset in sports science, offering insights that could catalyze enhancements in athletic performance and the development of more efficacious injury prevention strategies. Although this remains predominantly observational, it provides a framework for employing comprehensive measurement strategies to aid clinicians and practitioners. Consequently, a thorough understanding of the rheological properties of musculoskeletal tissues such as muscle, fascia, or tendon holds promise, particularly in the realm of rehabilitation.

6.2. The importance of methodology in measurement devices

6.2.1. Limitations of myotonometry

As delineated in Section 03, myotonometry facilitates a deeper comprehension of the rheological properties of musculoskeletal tissues directly in the field, enabling assessments of larger cohorts. Nonetheless, this method exhibits noteworthy limitations. Despite the MyotonPro device's assertion of measuring viscoelastic properties, it falls short of furnishing comprehensive insights into the viscous component, casting doubt upon its scientific reliability. [20] Furthermore, while its measurements of the elastic component appear valid when compared to a rheometer, it merely approximates elastic stiffness without directly providing the Young's modulus. Essentially, myotonometry estimates the stress or load experienced by a tissue rather than directly measuring the stiffness coefficient (k). This aspect assumes particular significance in a clinical milieu, where observing tissue responses to loading and identifying potential impairments are paramount. By employing diverse methods and measurement points, a broader and more nuanced depiction of estimated stiffness behavior in human tissues can be attained. Much like weather perceptions hinge upon comparative or culturally established norms, interpreting stiffness devoid of context may lack significance.

For runners, a solitary stiffness measurement may fail to convey meaningful information unless juxtaposed with a normative value or another measurement. Assessing PMAS through myotonometry in both resting and loaded conditions proffers insights into tendon adaptability in runners, which is indispensable for optimizing performance and averting injuries.[20]

6.2.2. Alternatives devices

Various methodologies are available for quantifying PMAS, each presenting unique benefits and challenges. This thesis has primarily focused on myotonometry for its convenience in assessing superficial myofascial structures.

Another prominent tool for PMAS assessment is SWE, a specialized form of ultrasonography that quantitatively measures the elastic stiffness of musculoskeletal tissues by gauging the speed of shear waves induced within the tissue, which correlates with tissue stiffness. SWE is prized for its objective, quantitative insights into tissue elasticity, allowing for the

identification of specific structures being assessed and facilitating deep myofascial evaluations. [249] However, SWE's implementation, both in research and clinical settings, requires extensive training and structured logistics, [249] alongside standardization of assessment techniques, including probe positioning [250] and operator applied pressure. [251]

In comparison, the MyotonPro device offers simplicity and ease of use, demonstrating good reliability in assessing the gastrocnemii and the Achilles tendon. [216] Nevertheless, it has been noted that myotonometry is more adept in at detecting changes across different muscle contraction levels than SWE. [137] While SWE has shown excellent reliability in assessing tissues like the skin, breast, or liver, [252–254] its efficacy in tendon evaluation, such as the Achilles or Patellar tendons, varied from poor to moderate. [255]

EOD technology, similarly, does not hinge on extensive operator experience, positioning it as a dependable PMAS measurement method. The MyotonPro, with its quick operation and portability, offers practical advantages over EOD, albeit at the cost of lesser accuracy in PMAS assessments. Crucially, the MyotonPro may fall short in providing comprehensive insight into viscous stiffness, limiting its application in thorough viscoelastic property evaluation.[20]

Ultrasonography emerges as a valuable tool for in-depth PMAS insights in vivo, offering a validated and reliable means for measuring stiffness in structures like the Achilles tendon or plantar fascia. It enables immediate visualization and quantification of tendons and fasciae rheology, enriching our understanding of dynamics under different various conditions and contributing significantly to the biomechanical comprehension of tendons and fasciae across a spectrum activities, treatments, or pathological states[186] (Figure 34).

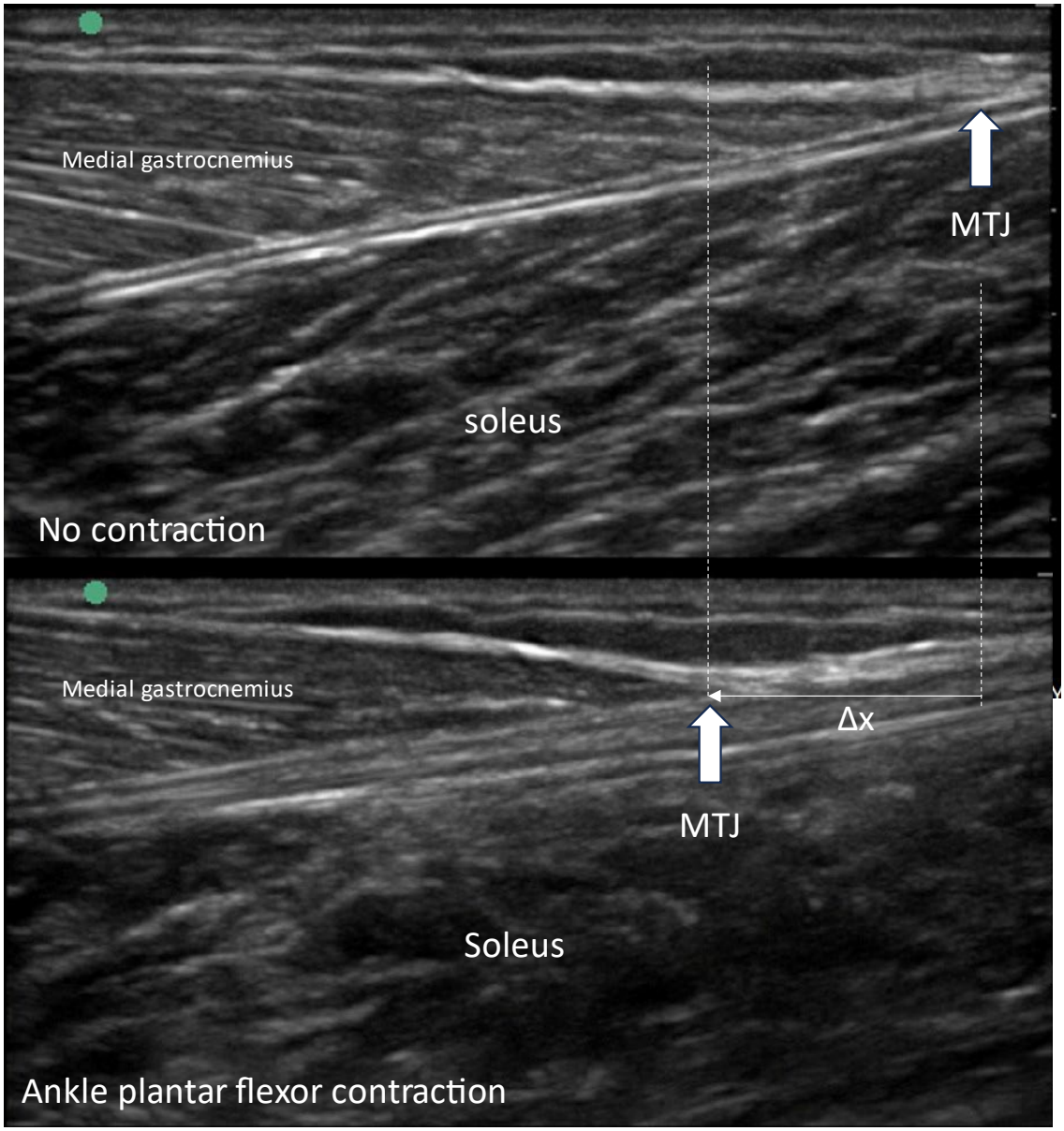

Figure 34. *The observation of the elastic stiffness of the Achilles tendon in a healthy 20-year-old male can be accurately assessed through ultrasonography. This method allows for the visualization of the Muscle-Tendon Junction (MTJ) of the medial gastrocnemius. During the assessment, the subject's ankle is maintained in a neutral position. When the subject performs an isometric contraction, a force F in Newtons is generated, which can be recorded by a force plate positioned under the subject's foot. Concurrently, the displacement Δx, measured in millimeters, can be measured using ultrasonography. According to Hooke's Law, as mentioned in the mechanical engineering section, the elastic stiffness of the Achilles tendon (k) can be calculated using the formula:* $k = \frac{F}{\Delta x}$ *where k, expressed in Newtons per millimeter (N/mm), provides a quantifiable measure of the tendon's ability to resist deformation under load.*

Operating this system poses challenges due to the requisite extensive operator training and clear guidelines to ensure reliable measurements. One of the intricate tasks involves standardizing the methodology and ensuring that participants consistently apply force. To measure stiffness (k), both the strain applied, visualized through ultrasonography, and the stress exerted on the tendon by the contraction of the triceps surae must be meticulously recorded. This stress is particularly pronounced beneath the foot where the force of the triceps surae is exerted.

A standardized device engineered to accommodate the patient's foot in a repeatable and clinically accessible manner is indispensable for precise measurement. The patient could be positioned either seated or supine, with their foot resting on a plate designed to capture the stress applied by or to the foot. This plate should allow rotation around the pivot point of the ankle, typically the medial malleolus, and be capable of recording stress in both static and dynamic positions. It should enable the patient to exert pressure against the plate while also permitting the measurement of external force applied by an operator, who manipulates the plate, thereby inducing passive stretching of the calf and the Achilles tendon (Figure 35).

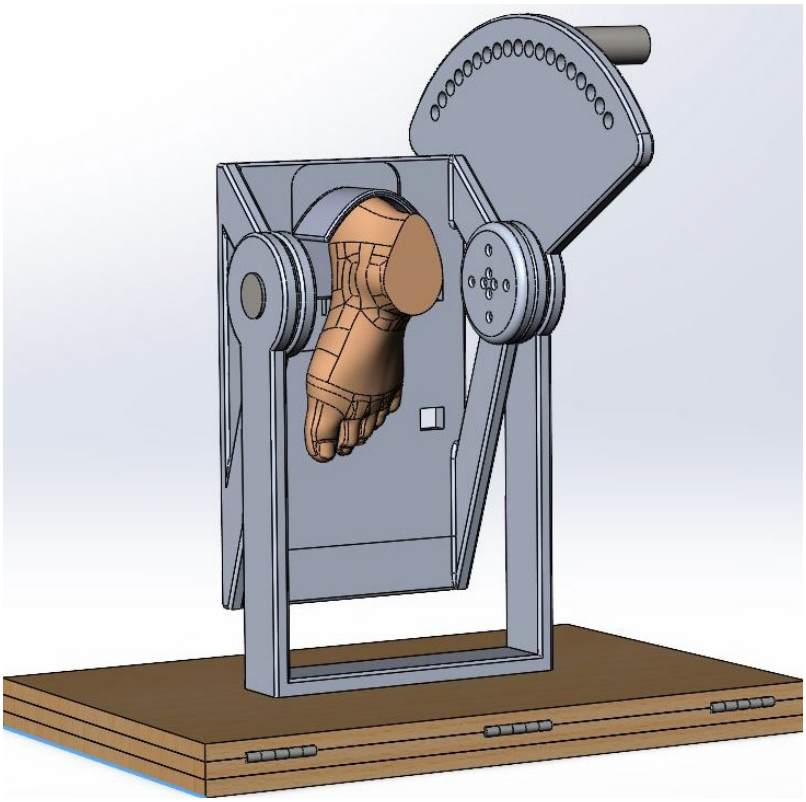

Figure 35. *Prototype of the measurement device for recording ankle amplitude and torque. An electronic goniometer can record the position of the ankle. This position can be either locked or left free to observe PMAS in static or dynamic conditions. A torquemeter can be placed on the axis of the device to record the pressure force applied by the resistive movement during dynamic assessments, or by the ankle plantar flexors, predominantly the triceps surae, in static conditions.*

To the best of our knowledge, no device has achieved perfect quantification of PMAS. Each method offers distinct advantages and limitations. The EOD stands out as the sole technology capable of measuring both elastic and viscous stiffness. However, its application is subject to notable constraints. Principally, the EOD is restricted to evaluating distal joints such as the ankle, wrist, or finger due to two primary factors. Firstly, the EOD necessitates the mobilization of the joint across a range of frequencies, particularly above 8 Hz, to capture the resonance frequency prevalent in musculoskeletal tissues. To accurately detect pathological variations in PMAS, a broad frequency spectrum encompassing values both below and above this resonance frequency is imperative. Consequently, the EOD's application is confined to lighter body segments to mitigate logistical challenges.

Furthermore, precise PMAS measurement with the EOD requires inertia reduction, rendering it suitable and reliable for distal, lighter segments. However, this criterion may not be applicable for larger body segments like the leg, arm, or thigh, where inertia significantly increases.

On the contrary, ultrasonography combined with elastography presents an opportunity to delve deeper into and measure elastic stiffness more comprehensively. These technologies furnish precise information about the specific tissue type under examination, which is immensely advantageous. In contrast, the EOD measures stiffness in a more generalized manner without delving into the specific rheological properties of individual tissues. Moreover, ultrasonography excels at assessing deeper tissues, a capability lacking in myotonometry and other methods, which primarily focus on superficial tissue examination. A significant drawback of ultrasonography and elastography is the requirement for an experienced operator. Ensuring consistent and reliable results with ultrasonography often entails relying on the same operator or investing significantly in training new operators. Ultimately, no single device emerges as superior for PMAS measurement; each one possesses its unique set of advantages and challenges (Table 19).

Table 19. Comparison of stiffness measurement devices

	Electromechanical Oscillation	Shear Wave Elastrography	Myotonometry	Ultrasonography
Methodology	Quantifying the torque at different frequencies (from 3 Hz to 12 Hz)	Quantifying the velocity of shear wave in the tissue	Quantifying the maximal acceleration of the tissue after a single external force	Quantifying the displacement of the tendon under a known force (muscle contraction)
Elastic Stiffness	yes	yes	yes	yes
Viscous Stiffness	yes	no	no	no
Where	Distal joints Ankle & Wrist	Any tissue under ultrasonography imaging, including deeper tissues	Superficial tissues (max 2cm under the skin)	Tendons
Validity	yes	yes	yes	yes
Reliability	yes	yes	yes	yes
Requirement	Untrained operator	Highly trained operator	Untrained operator	Highly trained operator
Duration	Less than 2 minutes per measurement	Time consuming	Less than 1 minute per measurement	Time consuming
Portable	Yes, but requires a heavy electrical supply	Yes, but requires a heavy electrical supply	Portable battery	Portable battery
On-field use	Possible but presents logistical challenges	Possible but presents logistical challenges	Easy to use	Possible but presents logistical challenges

6.3. "Stiffness" and running.

"What cannot be defined, cannot be discussed."

Georges Orwell, 1984

6.3.1. Terminology

In the realm of running and physical health, the term 'stiffness' is often misunderstood, leading to confusion between stiffness and mobility. Mobility refers to the ability to move within a specific ROM, either passively or actively, and is a key focus for healthcare professionals and sports performance specialists due to its direct impact on movement and functional activities. Contrary to common belief, stiffness and mobility should not be conflated; notably research has shown that elderly individuals can exhibit both decreased ROM and reduced PMAS [110], challenging the misconception that aging invariably leads to increased stiffness. This suggests that enhancing PMAS and myofascial tissue quality could be beneficial for that population. [256]

In the context of running, the concept of "stiffness" is frequently misapplied. Typically, reference to "stiffness" in running literature pertain to "dynamic stiffness" (Figure 16), a term used to assess the biomechanical response of the leg, including leg stiffness, vertical stiffness (Kvert), and joint stiffness (Kjoint), though the spring-mass-damper (SMD) model. These measurements assess the leg 'response to force and compression, vertical movements such as jumping, and the angular displacement due to joint moments, respectively. The misuse of terminology can lead researchers and clinicians to misinterpret the relationship between running stiffness, performance, or injury risk. The concept of stiffness in relation to running has been explored in the literature, but the use of varied stiffness types leads to different recommendations. Moreover, the employment of multiple terms to describe stiffness often results in ambiguous outcomes or guidance. Consequently, identifying the precise level of stiffness to enhance running performance or aid in recovery is currently not feasible. We advocate for the use of specific stiffness terminology to ensure better external validity (Figure 36).

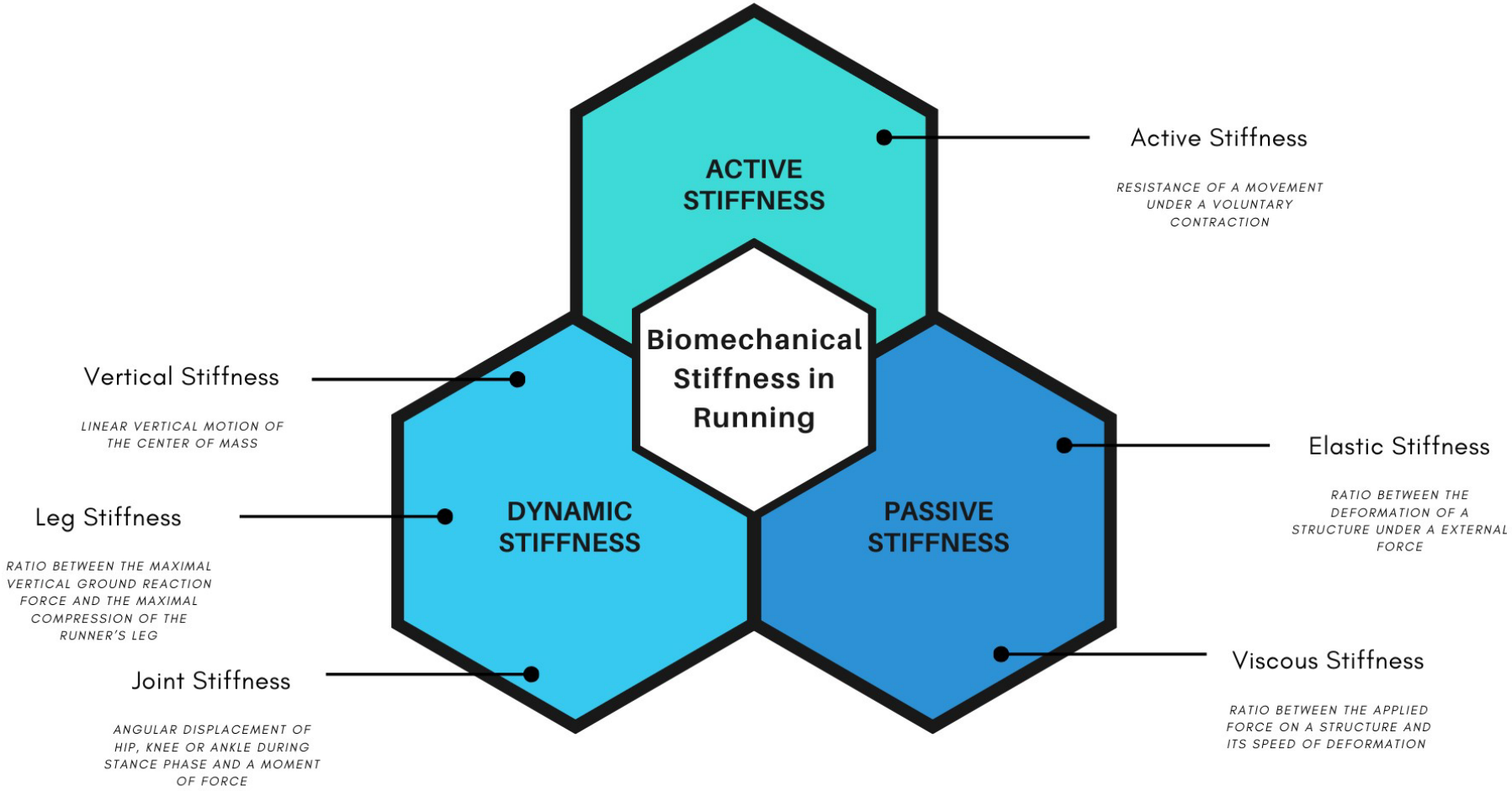

Figure 36. *Representation of biomechanical stiffness in running.*

6.3.2. To be or not to be … Stiff? (Figure 37)

Evidence indicates that stiffness, as evaluated by the SMD model, significantly impacts running performance. For instance, a study observed a negative correlation (r= -0.8) between vertical stiffness and oxygen consumption among middle-distance runners on a treadmill at 90% of their maximal aerobic velocity, suggesting less efficient runners exhibited more compliant (lower) "stiffness." [257] Conversely, leg stiffness did not correlate with aerobic performance in the same way vertical stiffness did, [46] suggesting humans can adjust their leg stiffness within a wider range compared to other forms of stiffness. Indeed, leg stiffness can double at constant speed simply by increasing step frequency, [59] highlighting the importance of adjusting stance according to the surface compliance for efficiency in endurance running.

Tam et al. noted a correlation between decreased ankle joint stiffness (r= 0.52) and increased knee joint stiffness (r= 0.38) with reduced oxygen consumption while running at 3.3 m s^{-1}. [258] This suggest that while knee joint stiffness varies with velocity, ankle joint stiffness remains relatively constant, highlighting the role of the triceps surae's muscle-tendon unit during the stance phase. [259]

Studies on running to exhaustion indicate that fatigue may lead to increased knee joint stiffness and decreased ankle stiffness. [260] [261] The reduced ankle stiffness in fatigued runners could serve as a compensatory mechanism to reduce stress on the Achilles tendon. [262] These findings underscore the complex relationship between muscle and tendon mechanics during running and the adaptations occurring with fatigue. [245]

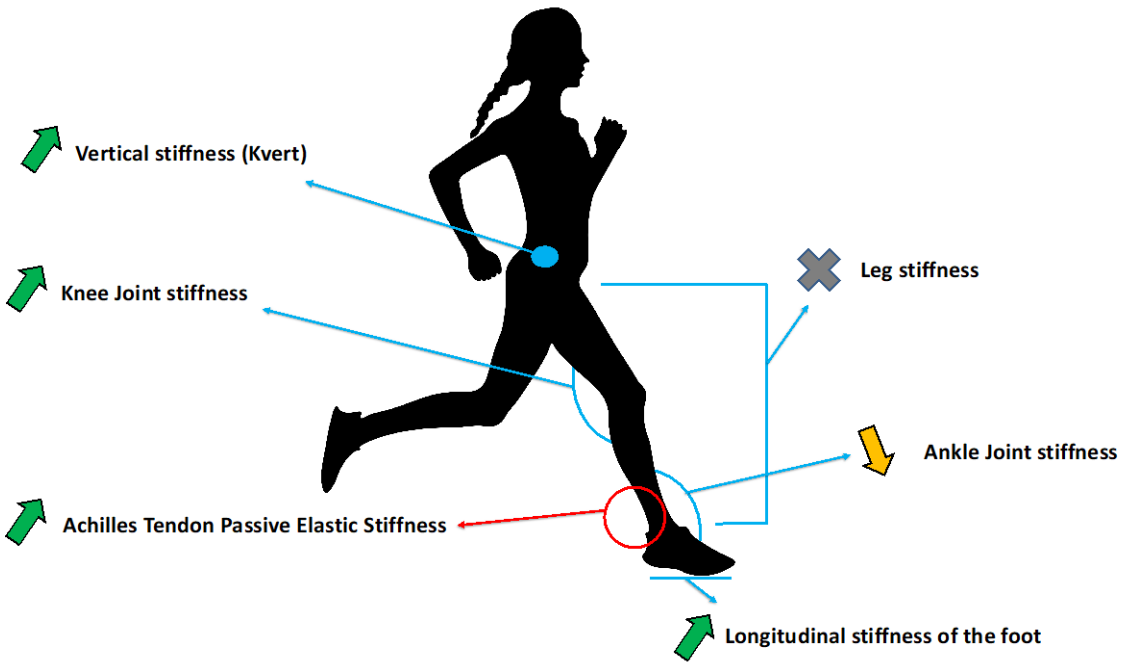

Figure 37. *Representation of stiffness interest in running performance. Green arrow represents positive correlation with running performance. Yellow arrow represents negative correlation with running performance. Grey cross represent no correlation with running performance.*

6.3.3. Footwear and artificial stiffness

Footwear may contribute to the passive stiffness experienced by runners, particularly around the ankle, though its impact on running performance remains a subject of debate. [263] In 2006, Roy and colleagues observed improvements in running performance with stiffer midsole running shoes among experienced runners (less than 40 minutes for a 10k). [264] Recently, there has been a surge in the popularity for modern, performance-focused running shoes within the running community. [265] Numerous shoe companies have initiated marketing campaigns to promote their products as advantageous for achieving faster running times. For example, Nike's 4% series was marketed as part of the Breaking2 project, aimed at breaking the two-hour marathon barrier. Shoe mass is acknowledged as a significant factor influencing the metabolic cost of running; heavier shoes can negatively affect running economy and performance. [266] Moreover, an optimal midsole stiffness may lead to reduced metabolic cost. [267] The industry has also shifted focus towards shoes that include a carbon plate, enhanced midsole stiffness, and specialized foam, which could potentially improve running oxygen consumption by up to 4% in elite runners in lab settings. [268] However, for recreational runners, these “super-shoes”

have not proven to be more economical than very light racing flats. These "super-shoes" blend traditional design with a more flexible midsole and curved carbon fiber plate [269], leading to increased longitudinal bending stiffness, which may improve a distance runner's economy. [270]

6.3.4. The foot

In addition to footwear, the ankle plantar flexor muscles, as well as the foot and its muscles, contribute significantly to the biomechanics of running by using potential (elastic) energy, thereby employing the principle of elastic stiffness. This process aids in minimizing energy expenditure and enhancing propulsion. Specifically, structures with the foot are responsible for between 8 and 17% of the mechanical work on the center of mass during acceleration and deceleration movements. [271]

The foot plays a pivotal role in the overall lower limb stiffness during running. Based on the "arch-spring" mechanism described by [272] the medial longitudinal arch of the foot acts as a spring, cooperating with the entire lower limb. [273] observed that during the initial phase of stance, as ground reaction force increases, foot compression absorbs mechanical energy. Subsequently, the ankle plantar flexor muscles and the foot's intrinsic muscles restore the ankle joint to its initial position at the end of the stance phase, releasing the stored elastic energy [274].

This "foot-spring" mechanism facilitates the generation of propulsive force during running. [275] The intrinsic foot muscles influence the stiffness of the foot's longitudinal arch during the stance phase, becoming more active and thus stiffening the arch during faster running paces, enabling greater force transfer between the body and the ground over shorter ground contact times. [276] Additionally, theories such as the "windlass" mechanism suggest an increase in the plantar fascia strain during the propulsive phase of walking, potentially elevating the foot's arch and thus its stiffness, primarily influenced by the plantar fascia. [277,278] However, recent studies like those by Welte et al. (2021) [279] indicate that, while the "windlass" mechanism tensions the plantar aponeurosis, it does not necessarily increase foot stiffness. Instead, elastic stiffness, and hence potential elastic energy, may be stored in the intrinsic foot muscles rather than in the plantar fascia, as they actively manage the foot energy function by generating and dissipating mechanical energy during positive and negative work phases of locomotor tasks. [280]

The plantar fascia and the Achilles tendon, while often considered separately, are anatomically connected by a thin layer of periosteal fibers present in some individual [281]. They share a key biomechanical role in running to support and propel the runner during the stance phase [282]. The Achilles tendon, extensively studied for its unique features such as its 3-dimensional helical structure and high collagen content (type I collagen with 90% of the extracellular matrix [283], and the 1% of elastin [284]), exhibits remarkable biomechanical properties, capable of withstanding significant loads. [285] For instance, it can withstand up to 9000 N of external load, or twelve times the body weight. [286]. Despite traditionally being assessed through in vitro studies or sophisticated methods like MRI or shear wave elastography, the Achilles tendon's role in running performance underscores the need for translating comprehensive biomechanical, rheological, and anatomical insights into practical applications for clinical and rehabilitation purposes.

6.4. Stiffness in rehabilitation

The Achilles tendon appears to have a significant impact on the storage and release of energy during running, whereas the gastrocnemius muscle does not. The importance of passive tissue rheological properties for PMAS can be attributed to the quasi-static use of the ankle plantar flexor muscle during the stance phase.

This component may be relevant for both **performance** and the **prevention** of running-related injuries. In terms of **performance**, this highlights the importance of strength training, especially the necessity of adding weights to enhance performance. [287] Bodyweight exercises alone may not generate sufficient power to increase maximum force or place enough stress on the Achilles tendon to induce rheological changes. However, adding resistance to these exercises could effectively stimulate the necessary modifications in the Achilles tendon's inherent properties, potentially improving overall performance. Furthermore, these rheological changes could also have implications for **preventing** repetitive strain running related injury. By enhancing the passive tissues' ability to withstand stress and strain, individuals may decrease their risk of developing tendinopathies or other conditions associated with excessive loading on the ankle plantar flexor muscle during activities like running or jumping.

Regarding the process of returning to sport for injured runners, assessing rheological properties could aid healthcare professionals in providing tailored recommendations. The mechanical properties of tendons are important for the passive transmission of forces from muscles to

bones, offering protection against sudden stretching, and serving as a mechanical buffer that enhances the efficiency of muscle action. [35] The presence of decreased tendon stiffness in Achilles tendinopathy, as demonstrated by numerous studies, involves a complex interplay of biomechanical alterations with substantial clinical implications. [52–54] ; Arya & Kulig, 2010 and Chimenti et al., 2014 used dynamometry and real-time ultrasonography to assess stiffness in individuals diagnosed with Achilles tendinopathy. [55,288] They reported a reduction in tendon elastic stiffness that could lead to increased muscle shortening as a means of compensating for the tendon's increased compliance. [55] This decrease in elastic stiffness could hinder the ability to produce rapid movements and negatively impact the speed at which force is generated. [56] Moreover, changes in the Achilles tendon's ability to store and release elastic energy, observed in individuals with Achilles tendinopathy, represent a significant shift in the biomechanical efficiency of the musculoskeletal system. [289] These modifications lead to a decreased in energy conservation during motion and high-intensity activities, like jumping, and may induce an increased energy dissipation as heat. These findings not only deepen our understanding of the pathogenesis of Achilles tendinopathy but also open avenues for developing targeted treatment approaches aimed at restoring tendon elasticity and functional performance. Conversely, a tendon with reduced elastic stiffness undergoes a higher degree of strain, thus increasing the risk of developing small tears within the collagen fibers. Over time, the accumulative effect of these microtraumas could make the tendon more prone to injuries and prolong the healing process. [290] Although additional clinical research is necessary to establish the relationship between tendon elastic (or viscous) stiffness and injury, it could be beneficial to regularly assess tendon elastic stiffness to monitor any potential improvements, stability, or deterioration in tendon function.

As highlighted in the initial study of this thesis [110] , the influence of aging on musculoskeletal tissue, particularly its rheological properties, could be significantly relevant for overall health and rehabilitation. We observed a decrease in PMAS in older individuals, specifically a decrease in viscous stiffness. For instance, sarcopenia—the loss of muscle mass associated with aging—could impact PMAS in individuals over 60 years old. [291] Moreover, aging is associated with changes in muscle architecture that could account for the decline in muscle capabilities as well as the reduction in PMAS. Studies on tendon stiffness have found that tendons in older individuals tend to be more compliant, or less stiff, compared to those in younger individuals. Over the past two decades, the interest in understanding how muscle and tendon rheological properties change with age has led to specific recommendations, such as muscle strengthening

and neuromuscular rehabilitation for the elderly [292]. Regular exercise, including running, can enhance tendon elastic stiffness and counteract the age-related decline in muscle function. Furthermore, integrating strength training into one's fitness regimen can strengthen both muscles and tendons, thereby mitigating the risk of muscle loss and preserving overall physical performance. By maintaining an active lifestyle and keeping track of their PMAS, individuals can more effectively maintain their muscular capabilities and overall health as they age.

6.5. General conclusion

This thesis endeavors to integrate fundamental theoretical concepts with clinical applications, ensuring that insights from one domain enrich understanding in the other. Initially, it is imperative to conduct a thorough exploration of fundamental biomechanical concepts, laying the groundwork for a deeper comprehension of principles such as "stiffness".

This work delves into rheological stiffness, emphasizing its relevance in clinical conditions, encompassing neurological and orthopedic disorders. The extensive section on myotonometry is tailored to augment the knowledge of researchers and healthcare professionals regarding the terminology and technology associated with this technique. Myotonometry technology exhibits promise across various healthcare sectors. Its utilization in assessing musculoskeletal disorders, such as tendinopathies and muscle injuries, yields valuable insights. Employing myotonometry within the context of running offers a distinctive opportunity to enhance our understanding of running-related injuries and performance enhancements.

The Achilles tendon plays a pivotal role in running performance, particularly concerning net metabolic cost of running. Although further research is warranted to substantiate this hypothesis, monitoring rheological properties could unveil new avenues for performance enhancement and injury management, potentially propelling runners and stakeholders in the running community to unprecedented levels of success.

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