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Original article

Characterization and modelling of the musculoarticular complex mechanical behavior in passive conditions. Effects of cyclic and static stretching

Caractérisation et modélisation du comportement mécanique du complexe musculoarticulaire en conditions passives. Influence de protocoles d'étirements cyclique et statique

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Abstract

This work aims to model the mechanical behavior of the musculoarticular complex (MAC). First, the implementation and the validation of original methodologies have enabled to assess elasticity, viscosity and friction of the MAC. The effects of cyclic and static stretching on these properties were then assessed. Changes in MAC mechanical properties induced by static stretching could mainly be explained by an acute increase in muscle lengths, while dissipative properties and stiffness of the MAC are only modified after cyclic stretching. Our results enable to suggest and discuss about some mechanisms probably implied in these adaptations. Finally, a rheological model was proposed and validated to model the mechanical behavior and the adaptations induced by cyclic and static stretching assessed in our studies. This model could then be used to simulate the effects of specific protocols performed for instance in sports or functional rehabilitation.

Résumé

Les travaux de cette synthèse visent à modéliser le comportement mécanique passif du complexe musculoarticulaire (CMA). L'implémentation et la validation de méthodologies originales nous ont permis de caractériser les paramètres d'élasticité, de viscosité et de frottement du CMA. Puis les modifications de ces propriétés ont été déterminées à la suite d'étirements cycliques et statiques. Les étirements statiques induisent principalement une augmentation transitoire de la longueur des muscles alors que les propriétés dissipatives et la raideur du CMA ne sont modifiées qu'après les étirements cycliques. Nos résultats permettent de discuter des mécanismes potentiellement impliqués dans ces réponses spécifiques. Enfin, un modèle rhéologique a été développé et validé afin de modéliser le comportement mécanique et les adaptations observées dans ce travail. Ce modèle pourrait par exemple permettre de simuler les effets de protocoles réalisés dans le cadre de pratiques sportives ou de programmes de rééducations fonctionnelles.

Keywords: Elasticity; Viscosity; Friction; Rheological model

Mots clés : Élasticité ; Viscosité ; Frottement ; Modèle rhéologique

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1. Introduction

When a skeletal muscle contracts in order to generate a movement or to maintain a posture, it also behaves as a viscoelastic material, and these mechanical properties must be taken into account when muscle functions are modeled in order to simulate human motion. The viscoelastic behavior of a passive musculoarticular complex (MAC) can be studied by imposing a standardized stretching protocol and measuring the MAC's response [1–3]. Nevertheless, in the literature, this methodology has been used mainly in clinical studies to describe the flexibility of various populations and the changes in the maximal range of motion [4]. Thus, the understanding of the mechanical behavior of the passive MAC is limited, and very few models of this behavior have been published in the literature.

For this reason, the main objective of the work described in this article is to develop a model of the mechanical behavior of the passive MAC. The first part (Section 2) of the article characterizes the mechanical behavior of the studied system, which must be done before the model can be developed. The second part (Section 3) aims at determining the effects of passive stretching protocols, which are classically performed within the framework of rehabilitation or sports practices, on the mechanical properties that were characterized in the first part. Last, on the basis of the result obtained in the first two parts, a rheological model of the mechanical behavior of the MAC is developed in a third part (Section 4). This article presents a synthesis of eight studies and was awarded by the price of the Mecabio Conference 2007.

2. Characterization of the passive mechanical properties of the MAC in vivo

2.1. Introduction

In vivo, it is possible to increase the lengths of muscles by moving the articulations. Starting from a given articular angle (θ_0), muscles crossing an articulation resist stretching. It is then possible to characterize the passive mechanical properties of the articulation by mobilizing the articular angle beyond θ_0 and measuring the passive torque (pT) developed by the MAC in resistance to the motion. This requires the determination of muscle electromyographic activity levels to be sure they are negligible [1–3], i.e., generally lower than 1% of the levels obtained during a maximum contraction [5,6]. This condition was observed in all the studies presented in the present article. Therefore, nervous factors and reflex mechanisms, which occur when the muscle is stretched in vivo, will not be studied.

Classically, it is considered that the pT developed in resistance to stretching is produced mainly by the muscular structures that are internal or external to the sarcomere and that are placed parallel to the contractile elements of the muscle. However, some studies showed that the tendon [7,8] and the contractile proteins [9–12] may also make a non-negligible contribution. In addition, some other studies considered that the joint can also produce a passive resistance [13,14]. Thus, it was considered that passive stretching enabled us to characterize the global mechanical properties of the MAC, including the implied muscles, tendons, and the mobilized articulation.

The first study was aimed at validating the experimental device used to characterize the mechanical behavior of the passive MAC. Having achieved that, some specific mechanical properties of the MAC are highlighted in the two following studies.

2.2. *Validation of the isokinetic Biodex dynamometer® used in passive mode [15]*

In all the studies presented in the present paper, a Biodex® isokinetic dynamometer (Medical Biodex, Shirley, NY, USA) allowed us to perform the passive stretching protocols at a given angular velocity and to measure the articular angle as well as the pT produced in resistance to the stretch. Nevertheless, if the use of isokinetic dynamometers to assess contractile properties of the muscle in vivo has been validated in the literature [16–18], to our knowledge, the validity of their utilization to measure the classically-used parameters in passive condition has never been demonstrated. Yet, during passive movements, torque levels are largely lower than during active contractions. Thus, a specific experimental design has been set up to determine the accuracy (i.e., the validity and the reproducibility) of the Biodex dynamometer® used in passive mode. Five cyclic stretching repetitions were imposed on an elastic rubber band at 22 preset velocities (from 0.017 to 5.24 rad.s^{-1}) using the dynamometer. The torque produced was measured using the dynamometer and was recalculated using the force and the position measured using external sensors. This protocol was performed twice over two different days. Velocity patterns performed by the dynamometer were also characterized, and our results show that these patterns were reliable both in term of the mean angular velocity (interclass correlation coefficient [ICC] = 1.00; standard error in measurement [SEM] < 0.26°.s^{-1}; coefficient of variation [CV] < 0.7%) and mean angular acceleration (ICC = 1.00; SEM < 1.46°.s^{-2}; CV < 3.6%). However, some discrepancies between the programmed and the measured speed profiles were observed when approaching the speed limit of the system. The torque measured with the dynamometer and the sensors were reliable (ICC = 1.00; SEM < 0.76 N.m; CV < 12.83%), although significant differences were observed between both methods. However, the SEM of the torque measured using the dynamometer in comparison to external sensors was not velocity dependent and was lower than 0.33 Nm (CV < 6.53%). Moreover, regressions between the two torque measurements were very close to the axes-bisector (r = 1.00, slope: 1.01 ± 0.01, y-intercept: −0.36 ± 0.22 Nm). Finally, our results showed identical decreases in torque measured using the dynamometer and the external sensors during the five cycles, indicating that these decreases were not due to the dynamometer. It can be concluded that the dynamometer performed valid and reliable torque measurements in passive mode, and was an accurate tool for determining the passive mechanical properties of the MAC.

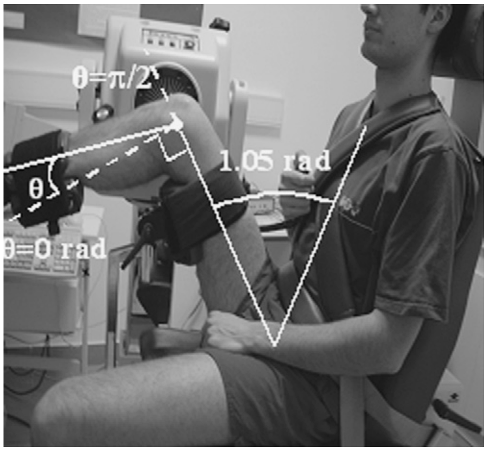

Fig. 1. Position of the subjects on the dynamometer for the studies of Sections 2. and 3. The angle trunk-thigh was set at 1.05 rad. The hamstring muscles were stretched by imposing a knee extension starting from the reference position (leg perpendicular to the thigh, $\theta = 0$ rad).

2.3. Musculoarticular stiffness (MAS) characterized by passive stretching: influence of the use of different mathematical models [19]

Many studies have shown the non-linearity of the relationship between the articular angle and the pT developed [1–3]. The MAS is then calculated as the slope of this relationship. However, various calculation methods for the MAS can be considered, and different mathematical models can be used to fit the pT–angle relationship [5,14,19–21]. Therefore, the aim of this study was to compare the fit of three mathematical models on the pT–angle relationships and to compare the resulting calculations of the MAS. The three mathematical models were:

- a fourth-order polynomial model (P4);
- a second-order polynomial model (P2);
- an exponential model very similar to the model proposed by Sten–Knudsen [4,22] (SK, cf. equation (1)).

$$pT(\theta) = \left(\frac{A}{\alpha}\right)(e^{a\theta} - B) \tag{1}$$

where A, B, and α are constants.

Eight healthy males (26.4 ± 4.4 years; 73.4 ± 7.2 kg; 177.8 ± 9.8 cm) volunteered to participate in the study. The position of the subjects was similar to the position used in previous studies reported in the literature [2,19,21,23] and is described in Fig. 1. The subjects sat on the Biodex dynamometer®, and their right thigh was elevated in order to set the trunk-thigh angle at 1.05 rad. The rotation axis of the dynamometer was aligned with the estimated rotation axis of the knee in the sagittal plane. Due to maximal hamstring stretch, all the subjects were unable to completely extend their knees in this position. The tests began with the leg perpendicular to the thigh (reference angle: $\theta = 0$ rad), and the subjects were stretched until 80% of the previously determined maximum range of motion (ROM) was reached [24].

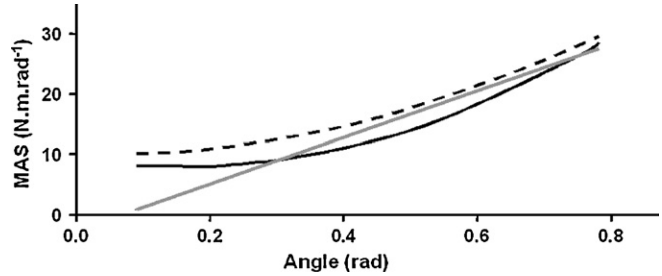

Fig. 2. Relationships between the musculoarticular stiffness (MAS) and the articular angle calculated with the polynomial model of the fourth order (black continuous line), the second order polynomial model (grey continuous line) and the model of Sten–Knudsen (dashed line).

The fits of the three mathematical models on the pT–angle relationships were very good (P4: $R^2 = 0.994 \pm 0.004$; P2: $R^2 = 0.988 \pm 0.007$; and SK: $R^2 = 0.993 \pm 0.005$). The P2 and SK models, which need three parameters for optimization whereas the P4 model requires five parameters, enabled us to calculate stiffness indices (i.e., the slopes of the relationships between MAS and angle or MAS and the pT). Therefore, the use of P2 and SK was preferred to the P4 model. However, significant differences in MAS calculations were obtained using the three models (Fig. 2). The P2 model provided MAS levels that were significantly lower than those provided by the other two models. That is why the SK model seemed to be more appropriate to model the pT–angle relationship and to calculate the MAS. In the third part (Section 4) of this paper, the SK model will then be used to model mechanical behavior.

2.4. Effects of stretching velocity on passive resistance developed by the knee musculoarticular complex: contributions of frictional and viscoelastic behaviors [25]

Considering the torque relaxation during a static stretching protocol [2,5,19,26] and the dissipation of energy during a loading–unloading cycle [21,27,28], it has been classically accepted in the literature that the MAC has a viscoelastic behavior [1–3,21,29]. Nevertheless, some other mechanical behaviors, such as plasticity and friction, can partially explain these phenomena on a theoretical basis. Since plasticity implies irreversible alterations, it can be considered that both the isolated muscle-tendon unit [30] and the in vivo musculoarticular complex [1,6,21,28,31] do not exhibit plastic behavior when muscles are stretched in a physiological range. However, the relative contributions of viscosity (i.e., fluid friction) and solid friction to the dissipation of energy by the MAC are still not well known [31]. The viscosity of the MAC should be highlighted by the dependence of the MAC's response at the speed of stretching, whereas solid friction implies a dissipation that is independent the loading rate [32,33]. However, the studies dealing with the effects of stretching velocity in the literature are relatively contradictory. Some studies indicated a large increase in the pT with the stretching velocity (e.g., a linear increase of 700% between stretching velocities in the range of 0.17 and 3.14 rad.s^{-1} [34]). Other studies indicated a much smaller dependence (e.g., a linear increase of about 15% between stretching velocities in the range of 0.09 and 2.09 rad.s^{-1} [35], and about 19% between 0.17 and 3.68 rad.s^{-1} [36]). To our knowledge, no study has determined the influence of the stretching velocity on the dissipative properties of the MAC.

Therefore, nine healthy males (age, 25.4 ± 3.0 years; height, 182.4 ± 7.5 cm; and weight, 76.6 ± 8.1 kg) performed passive extensions–flexions of the knee at five preset velocities (0.087, 0.52, 1.05, 1.57, and 2.09 rad.s^{-1}) in the position described in the previous study (Fig. 1). The results showed significant ($P < 0.001$) increases as the stretching velocity was increased of pT (between +17.6 and +20.8%, depending on the considered knee angle); of the potential elastic energy stored during the loading (i.e., the area under the loading pT–angle relationship, E: +22.7%); and of the dissipation coefficient (i.e., the energy dissipated during the cycle normalized by E, DC: +22.8%). These results suggest that the effect of viscosity on the MAC's mechanical behavior is limited. A linear model fitted the torque-velocity ($0.93 < R^2 < 0.98$), E-velocity (Fig. 3A; $R^2 = 0.93$), and DC–velocity (Fig. 3B, $R^2 = 0.99$) relationships quite well. The linear relationship between DC and velocity indicated that the DC does not tend towards zero for the slowest velocities and that the dissipative properties of the MAC can be modeled by combining linear viscosity and friction.

2.5. Conclusions

In this first part (Section 2), a methodology of characterization of the mechanical properties of the passive MAC was implemented and validated. First, this methodology allowed us to model the non-linear relationships between the pT and the articular angle in order to calculate the passive MAS. Afterward, it was shown the dissipation of energy by the MAC can be explained by combining linear viscosity and friction. These two studies are the first step towards the implementation of our model of the MAC's mechanical behavior (Section 4).

To account for the possibility of preconditioning, some loading–unloading cycles generally precede the characterization of the mechanical properties of a material or a structure. In the case of some viscoelastic materials, this step prevents taking into account a possible accommodation during the first loading cycles. Thus, some authors performed a preconditioning before assessing the mechanical properties of the MAC in vivo [31,37] or the passive isolated muscle ex vivo [38,39]. In the studies of the Section 2, no preconditioning was performed before the measurements of the mechanical properties. Indeed, this kind of protocol applied to the musculoarticular complex can be assimilated into a passive stretching protocol similar to the protocols usually performed within the framework of sporting practices and functional rehabilitations. In this second part (Section 3) of the present article, the changes induced by the passive stretching protocols were studied. These studies (Section 3) should contribute to a better understanding of the mechanisms related to the acute effects of passive stretching. These changes will also be integrated into the mechanical model of the passive MAC developed in the third part (Section 4).

3. Acute effects of cyclic and static passive stretching protocols on the passive mechanical properties of the MAC

3.1. Introduction

Passive stretching protocols are classically performed in many sporting practices or rehabilitation programs. Nevertheless, the effects of passive stretching protocols are still not well understood, and these effects remain a topic of continued interest for researchers. Several studies have shown that the pT [2,5] and the MAS [5,24] were decreased after some passive stretching protocols. But in order to explain such changes, three main mechanisms have been proposed in the literature:

- an increase in muscle-tendon resting length [30];
- a reorganization of the tissues of the muscle-tendon unit [40–43];
- a thixotropic behavior explained by a redistribution of the fluids of the MAC [5,6].

Nevertheless, to our knowledge, no previous study has determined the relative contributions of these mechanisms in the effects of passive stretching protocols.

Therefore, using the methodologies developed in the Section 2, the acute changes in the MAC's mechanical properties induced by passive stretching protocols performed in vivo were assessed to obtain a better understanding of the effects of these

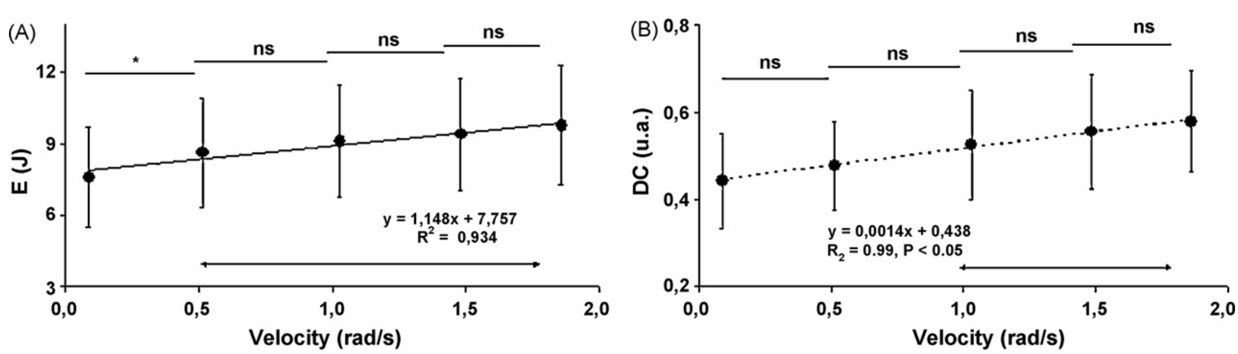

Fig. 3. A. Potential elastic energy stored during the loading (E). B. Dissipation coefficient (DC) as functions of the average angular velocity during the loading. ***: $P < 0.001$, ns: $P > 0.05$. The arrows correspond to the range velocity on which the parameter is significantly higher ($P < 0.05$) than the lowest speed tested.

protocols. In addition, this study should also enable us to adapt our modelling of the mechanical behavior of the MAC in order to take into account the effects of passive stretching. The two main passive stretching protocols, i.e., static stretching (maintaining a specific level of stretching for a given time) and cyclic stretching (undergoing cycles of loading–unloading), were studied.

3.2. Protocol

The position of the subjects was the same as previously indicated (Fig. 1). Eight healthy male volunteers (age, 23.3 ± 1.9 years; height, 181.3 ± 7.0 cm; and weight, 74.3 ± 4.7 kg) performed five loading–unloading cycles, up to 80% of the maximal ROM. The effects of the cyclic stretching were then assessed by comparing the first cycle and the fifth cycle. A 10-minute rest period was then allowed, and the subjects repeated 30 seconds of static stretching six times. A post-test loading–unloading cycle was then immediately performed, and the results were compared with the first (pretest) cycle of the cyclic stretching protocol in order to determine the effects of the static stretching.

3.3. Effects of cyclic and static stretching on the pT

In the literature, the effects of the stretching on the pT were mainly determined at a given articular angle. Therefore, this first study aimed at determining the effects of passive stretching on the whole pT–angle relationship. In accordance with the data of the literature [2,5], significant decreases of pT were observed following the two stretching protocols. During cyclic stretching, the pT decreased ($P < 0.05$) for all of the articular angles during the loading, but the range of the decreases was more important at the beginning of the ROM (knee flexed) than at the end (Fig. 4A). During the unloading, the decrease was significant ($P < 0.05$) only at the beginning of the knee flexion and was of very low amplitude (Fig. 4A). After static stretching, pT decreased ($P < 0.05$) for all the articular angles during the loading, but the magnitude of the decrease was more important at the end of the ROM than at the beginning (Fig. 4B). The decrease was also significant ($P < 0.05$) during the unloading. Thus, our results showed that the changes in pT after passive stretching depend on the articular angle considered. Our results also showed that these changes were qualitatively different after these two stretching protocols. Therefore, we hypothesized that the mechanisms during the two stretching protocols were also different.

3.4. Effects of passive stretching on the dissipative properties of the passive MAC [28]

In the same way as in Section 2.4, the DC was calculated for the first cycle, the fifth cycle, and after the static stretching protocol. The DC decreased significantly (−17.8%) during cyclic stretching. This decrease can be explained because of the systematic decrease in pT during the loading after cyclic stretching, which induced a decrease in the potential elastic energy stored (E) and because of the small changes in pT during the unloading (Fig. 4). Then, the ratio of dissipated energy and E increased, which induced a decrease in the DC. Consequently, the effects of cyclic stretching can be modeled based on the changes in fluid friction (viscosity) and solid friction already studied in the Section 2.4 (Section 2). An additional study will be required to assess the effects of cyclic stretching on these two mechanical behaviors. In contrast, the DC was not changed significantly after the static stretching. This result is explained by concomitant decreases in E and dissipated energy (Fig. 4). The assumption of an increase in muscle resting length was then formulated to explain the decrease in the pT quantified in the previous study, and this assumption is discussed in the following study.

3.5. The decrease in the pT after passive stretching can be modeled by a shift of the pT–angle relationship? [44]

The increase in muscle resting lengths would imply a shift to the right of the pT–angle relationships without changes in their shapes. Therefore, we developed a procedure to overlay a constant angle on the pT–angle relationships of the fifth cycle and the post-test on the relationship of the first cycle. The overlaid pT–angle relationship of the fifth cycle was significantly different ($P < 0.05$) from the relationship of the first cycle (Fig. 5A). As a result, it can be concluded that the changes in pT induced by cyclic stretching cannot be modeled by a shift of the pT–angle relationship, indicating that the shape of

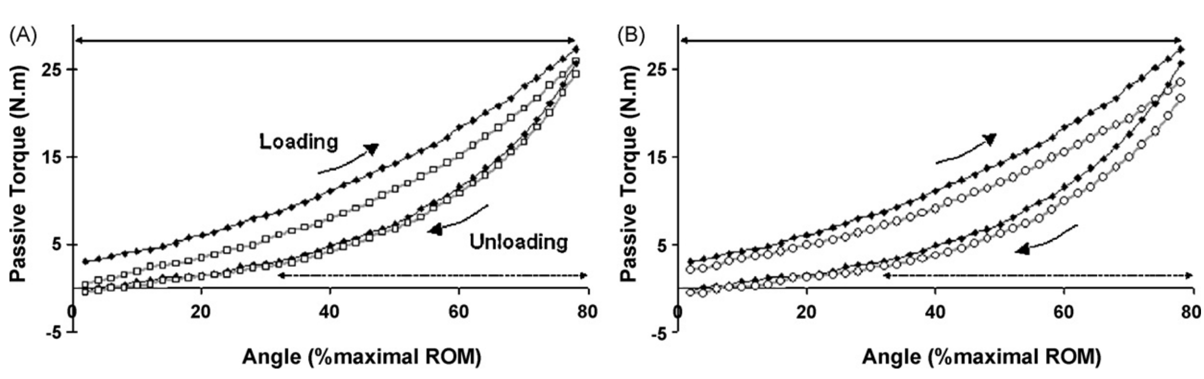

Fig. 4. Effects of cyclic stretching on the passive torque. A. Passive torque-angle relationships of the first cycle (◆) and the fifth cycle (□). B. Passive torque-angle relationships of the pretest (◆) and the post-test (○). The articular angle is expressed as a percentage of the maximal range of motion (ROM). Continuous arrow: significant decreases ($P < 0.05$) during the loading; dotted arrow: significant decreases ($P < 0.05$) during the unloading.

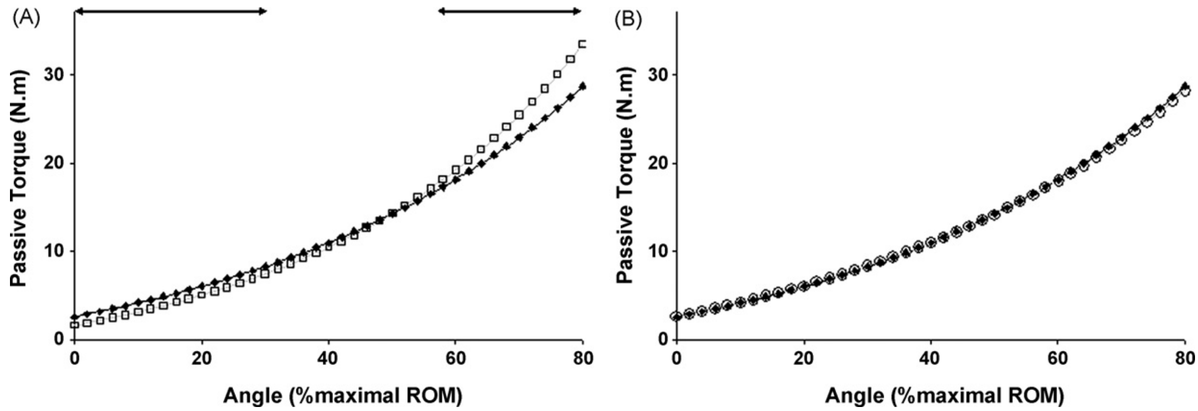

Fig. 5. A. Relationships between the passive torque and the articular angle (expressed as a percentage of the maximal range of motion, ROM) of cycle 1 (◆) and of cycle 5 readjusted of a constant articular angle (□); ●: significant difference ($P < 0.05$). B. Relationships of the pretest (◆) and of the post test readjusted of a constant articular angle (○); no significant difference was noted between the two curves.

this relationship has changed. On the other hand, there was no significant difference between the pretest relationship and the overlaid post-test relationship (Fig. 5B). Therefore, a shift to the right perfectly modeled the decrease in pT after static stretching. Contrary to the effects induced by cyclic stretching, the effects of the static stretching can thus be explained by increases in the lengths of the resting muscles, indicating that their intrinsic mechanical properties would not have been changed.

In order to confirm this last assumption, we measured local muscle elasticity using the transient elastography technique [45]. Our results did not show any significant change in the local elasticity of the *gastrocnemius medialis* muscle after 10 minutes of static stretching [46]. These results confirmed that the effects of static stretching could be explained mainly by increases in the lengths of the resting muscles without changes in their intrinsic mechanical properties.

3.6. Effects of cyclic stretching on passive mechanical properties of the knee joint: influence of stretching velocity [47]

The previous study showed that the increase in muscle resting length cannot explain the effects of cyclic stretching. The changes in dissipative properties assessed in Section 3.4, above, could be modeled by decreases in fluid friction (viscosity) and/or solid friction already studied in the first part (Section 2). Considering the mechanisms proposed in the Section 3.1, changes in fluid friction should reflect mainly the redistribution of fluids, while changes in solid friction should reflect changes mainly located in the tissues of the muscle-tendon unit. It has already been shown that fluid friction and solid friction (Section 2.4) can be assessed as the slope and the y-intercept, respectively, of the linear relationship between the DC and the stretching velocity [25]. Therefore, nine healthy male volunteers (age, 25.4 ± 3.0 years; height, 182.4 ± 7.5 cm; and weight, 76.6 ± 8.1 kg) performed five loading–unloading cycles at five different velocities (0.087, 0.52, 1.05, 1.57, and 2.09 rad.s^{-1}) in the position described in Fig. 1. Our results (Fig. 6) show that the effects of cyclic stretching were similar at the different velocities. In addition, the slope of the DC–velocity relationship was not significantly changed, whereas the y-intercept decreased significantly (Fig. 6B, 1st cycle: 0.432 ± 0.117, 5th cycle: 0.379 ± 0.112), indicating that the effects of the cyclic stretching could be modeled by a decrease in solid friction. Fluid redistributions can then be rejected as an explanation for the changes induced by cyclic stretching, and mechanisms, concerning muscular and tendinous tissues, could be proposed as an explanation.

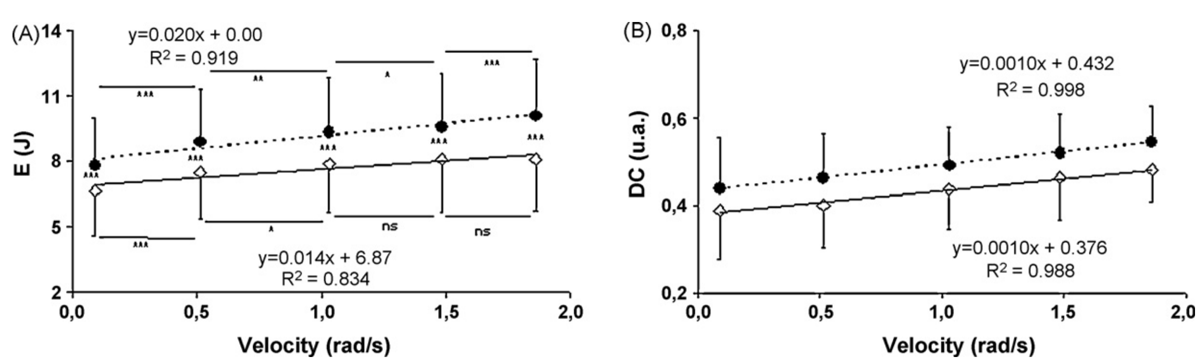

Fig. 6. A. Relationships between the mean velocity during the loading and the stored energy (E). B. Relations between the mean velocity during the loading and the dissipation coefficient (DC). ***: $P < 0.001$; **: $P < 0.01$; *: $P < 0.05$; ns: $P > 0.05$. ●: ◇: first cycle; □: fifth cycle.

McNair et al. [6] showed that pT and MAS decreased more at the highest stretching velocity. However, McNair et al. tested both angular velocities over a 2-minute period, and, hence, the number of cycles completed was quite different across their two angular velocities. The large changes observed by McNair et al. showed that repeated cycles of motion were very influential in inducing decreases in torque and stiffness. Our results showed that, when the same number of cycles is performed, the effects of cyclic stretching do not depend on the stretching velocity. The number of stretching cycles, not the rate of the loading, seems to determine the effects of cyclic stretching. This conclusion corroborates the implication of reorganizations of muscular and/or tendinous tissues.

3.7. Conclusions

On the basis of the methodology developed in the first part (Section 2), this second part (Section 3) has highlighted that the mechanisms implicated in the acute effects of cyclic and static stretching are probably different. Cyclic stretching induces a change in the dissipative properties of the MAC, and this change can be modeled by a decrease in friction. Then, this change would be attributed to ruptures of the actin-myosin bounds [43] or intramolecular connections [42] and/or to a reorganization of the collagen structures [40,41]. The relative contributions of these mechanisms to the effects of cyclic stretching remain to be determined. On the other hand, static stretching induces a shift to the right of the pT–angle relationship without changing the intrinsic mechanical properties of the MAC, which could be explained mainly by an increase in the lengths of the resting muscles.

The studies described in this section enabled us to make some assumptions concerning the mechanisms involved in the changes of MAC's passive mechanical properties. However, we cannot check these assumptions using the methodologies developed in the Section 2 and they cannot be used to clarify the mechanisms observed at the origin of the adaptations, e.g., to extrapolate the functional effects of stretching within the framework of a specific movement. By combining pT measurements and inverse dynamic analysis, it is impossible to determine the contribution of pT in the realization of a complex movement [48]. A mechanical model of these properties would then be useful, for example, to simulate the effects of these changes during the realization of complex tasks (e.g., walking, running, and jumping).

The last part of our work aimed at implementing a model in order to simulate the mechanical behavior of the passive MAC. Some adaptations of this model could make it possible to take into account the effects of the passive stretching protocols determined in the Section 3.

4. Modelling of the mechanical behavior of the passive MAC in vivo [49]

4.1. Introduction

Some studies have developed models of the MAC's behavior in passive conditions [31,48,50–52]. However, most of those studies did not take into account dissipative properties of the MAC or the dependence on the stretching velocity [48,50–52]. One study [31] implemented a complete model based on the quasi-linear, viscoelastic theory of Fung [39]. This approach is purely mathematical, and it does not reflect mechanical properties (i.e., elasticity, viscosity, and friction) in the same way as the approach of the Sections 2. and 3. Moreover, most of models of muscular behaviors or MAC behaviors during contractions are based on rheological models [4]. Therefore, in order to be able to combine, in the long term, the models of the CMA in active and passive conditions, we chose the rheological modelling approach.

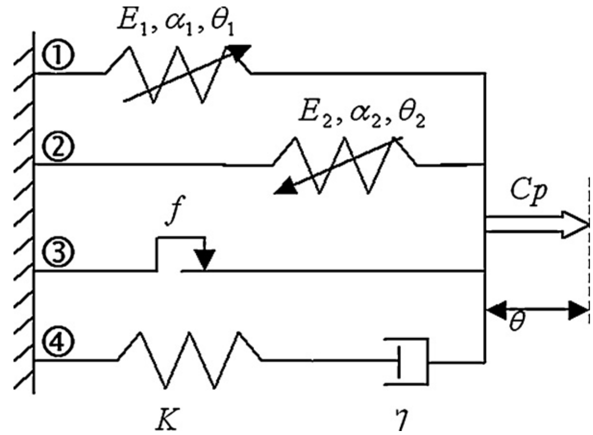

Fig. 7. Rheological model of the mechanical behavior of the musculoarticular complex developed in order to simulate the relationships between the passive torque (pT) and the articular angle (θ). The nine parameters (E_1, α_1, θ_1, E_2, α_2, θ_2, K, η, F) were determined by numerical optimization.

The rheological model presented in Fig. 7 was set up on the basis of the results obtained in the Section 2. The two, nonlinear springs modeled increases in pT at the two sides of the equilibrium point of the ankle joint (in dorsal and plantar flexion) on the basis of the exponential model studied in Section 2.3. The association between linear viscoelasticity and solid friction was used to take into account the dissipative properties of the system (Section 2.4). Initially, our experimental results were compared with the outputs from this model, and the results are presented in this paper. Then the model was modified so that it could simulate the realization of stretching protocols studied in the Section 3.

4.2. Fitting of the model

Nine healthy male subjects (age, 25.6 ± 2.7 years; height, 182.0 ± 8.3 cm; and weight, 75.1 ± 5.7 kg) volunteered to participate in this study. First, they performed five loading–unloading cycles at $0.175\,\mathrm{rad.s^{-1}}$ (from 0.70 rad in plantar flexion until 80% of the maximal ROM in dorsal flexion) using the isokinetic dynamometer shown in Fig. 1. Then, the subjects performed five passive cycles at five stretching velocities (0.035, 0.52, 1.05, 1.57, and $2.09\,\mathrm{rad.s^{-1}}$). Experimental pT–angle relationships obtained were used to develop the model.

The model shown in Fig. 7 was implemented with Simulink® software (The Mathworks, Natick, USA). The numerical resolution of the equations enabled us to use the articular angle that

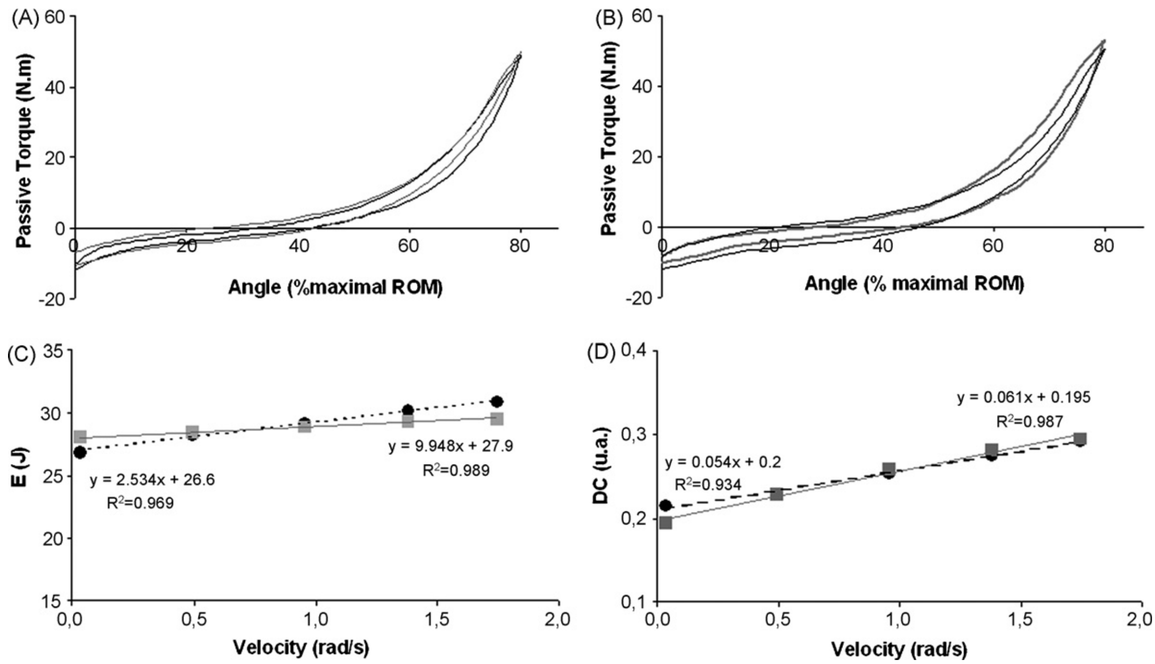

Fig. 8. Relationships between the passive torque and the articular angle (expressed as a percentage maximal range of motion, ROM) obtained during the experimentations (–) and using the model ( ) obtained with the model A at a velocity of 0.035 rad.s^{-1} and B at a velocity of 1.571 rad.s^{-1}. Relationships between C, the potential elastic energy stored during the loading (E), D, the dissipation coefficient (DC) and the average angular velocity calculated during the cycle. ●: experimental results, ■ model.

was actually imposed on the articulation as input to the model, thereby taking into account the real history of the loading. All the parameters of the model were obtained by minimizing the difference between the experimental and modeled data thanks to a nonlinear least squares method (Levenberg–Marquardt algorithm). A first optimization was used to determine the six elastic parameters of the model. The three parameters concerning the model of dissipation (linear viscosity associated with a solid friction) were determined during a second optimization.

Averaged pT–angle relationships obtained at slow and fast velocities are presented in Fig. 8A and B. Our results highlighted a good agreement between the experimental and modeled relationships with an average quadratic error of 1.34 ± 0.29 Nm (variation range: 0.80–1.89 N.m). The relationships obtained between the stretching velocity and E or DC also indicated good agreement between the model and the experimental results (Fig. 8C and D, respectively). For clarity reasons, the results presented were the average obtained from the nine subjects, but similar results were obtained for each subject.

Consequently, the rheological model presented in Fig. 7 could be used to accurately simulate the mechanical behavior of the MAC placed in passive conditions. However, it should be noted that the thickness of the hysteresis was slightly overestimated at the beginning of the ROM and was underestimated on the end (Fig. 8C and D). A non-linear model of viscosity according to the articular angle would then improve our model [53]. Nevertheless, it would require the development of new experimental procedures within the framework of the passive stretching, e.g., sinusoidal perturbations [4], in order to characterize the relationships between dissipation, stretching velocity, and the articular angle.

4.3. Adaptations of the model to take into account of the effects of static and cyclic stretching

According to the results of the Section 3, we propose to model the adaptations during five cycles performed at 0.175 rad.s^{-1} (cyclic stretching) by a change in friction using another friction element placed in parallel with the existing element in the model. This friction element was adjusted to simulate the additional friction during the loadings that depends on the considered cycle. This model of the effects of the cyclic stretching enabled us to simulate, in a very satisfactory way, the changes in the pT–angle relationships (Fig. 9A) as well as the changes in E and DC (Fig. 9B).

In the Section 3 it was shown that the changes in pT after static stretching can be modeled by a shift to the right of the CP–angle relationships. This shift was explained by an increase in the lengths of resting muscles. In the model, the effects of static stretching will be taken into account by changing the elasticity equilibrium angle θ_1 (Fig. 7), which will result in a shift of $\Delta\theta_1$. A 10-minute static stretching protocol was then performed immediately after the cycles at the various stretching velocities. Then, the same protocol was after static stretching than before static stretching (5 loading–unloading cycles at 0.175 rad.s^{-1} and five cycles at 0.035, 0.52, 1.05, 1.57, and 2.09 rad.s^{-1}).

Therefore, during the 5th cycle performed during the post-test, parameter values were the same as the values obtained during the modelling step previously described, except for θ_1, which was estimated using a numerical optimization. A simulation of the post-test was then performed. Very good agreement between the experimental results and the model's results was then found (Fig. 9C and D).

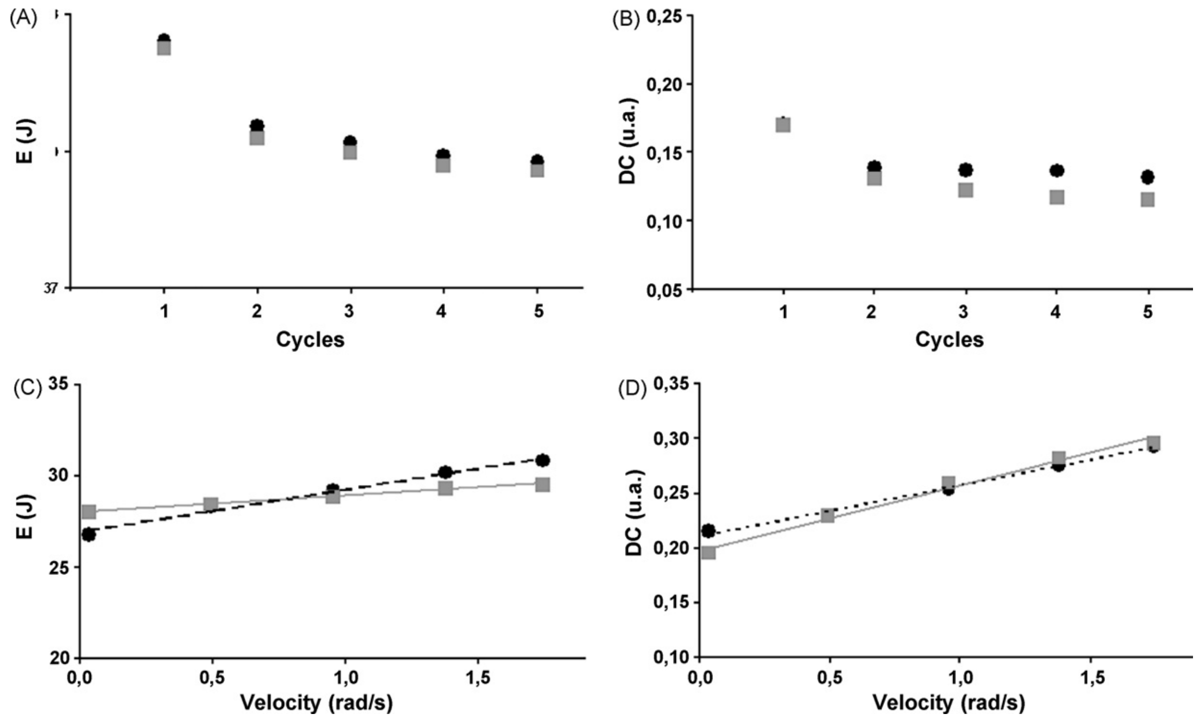

Fig. 9. Changes in A, the elastic potential energy stored during the loading (E), and B, the dissipation coefficient (DC) during the five loading–unloading cycles. C. Relationships between E and averaged angular velocity calculated during the cycle after the static stretching (post-test). D. Relationships between DC and the average angular velocity calculated in post-test. ● : experimental results, ■ : model.

4.4. Conclusions

A rheological model (Fig. 7), composed of non-linear elastic elements, a linear viscoelastic element, and a friction element, was developed. Very good agreement between the model's results and experimental results was shown, indicating that our model is valid. This model was then modified to take into account the effects of cyclic and static stretching. Very good agreement between the model's results and the experimental results was also highlighted, validating the choices that were made and showing that our model can be used to simulate a passive stretching protocol.

5. Conclusions and perspectives

This work, based on an original approach for dealing with the MAC's passive mechanical properties, enabled us to better understand this mechanical behavior and the acute adaptations during a passive stretching protocol. Our model could be used to simulate the effects of specific protocols performed, for example, within the framework of sport practices or functional rehabilitations. Recommendations could thus be made concerning the effectiveness of various stretching protocols. The model of the MAC's mechanical behavior could also be integrated into the models that simulate human motions, allowing the evaluation of the contribution of passive structures during various tasks. In the work presented in this paper, we were limited in the extent to which we could integrate the geometry of the MAC into our model. This limitation could be eliminated through the use of various imaging techniques, such as MRI, ultrasonography and EOS.

Conflict of interest statement

All authors agree that there are no conflict of interest issues in this research.

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