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Acute effects of static stretching on passive stiffness of the hamstring muscles calculated using different mathematical models

Antoine Nordez ^a, Christophe Cornu ^{a,*}, Peter McNair ^b

^a Laboratoire “Motricité, Interactions, Performance”, JE 2438, UFR STAPS, Université de Nantes, France

^b Physical Rehabilitation Research Centre, Auckland University of Technology, Private Bag 92006, Auckland 1, New Zealand

Abstract

Background. The aim of this study was to assess the effects of static stretching on hamstring passive stiffness calculated using different data reduction methods.

Methods. Subjects performed a maximal range of motion test, five cyclic stretching repetitions and a static stretching intervention that involved five 30-s static stretches. A computerised dynamometer allowed the measurement of torque and range of motion during passive knee extension. Stiffness was then calculated as the slope of the torque–angle relationship fitted using a second-order polynomial, a fourth-order polynomial, and an exponential model. The second-order polynomial and exponential models allowed the calculation of stiffness indices normalized to knee angle and passive torque, respectively.

Findings. Prior to static stretching, stiffness levels were significantly different across the models. After stretching, while knee maximal joint range of motion increased, stiffness was shown to decrease. Stiffness decreased more at the extended knee joint angle, and the magnitude of change depended upon the model used. After stretching, the stiffness indices also varied according to the model used to fit data. Thus, the stiffness index normalized to knee angle was found to decrease whereas the stiffness index normalized to passive torque increased after static stretching.

Interpretation. Stretching has significant effects on stiffness, but the findings highlight the need to carefully assess the effect of different models when analyzing such data.

1. Introduction

Passive stretching exercises are commonly performed in sports and rehabilitation. The acute effects of stretching on biomechanical properties and force production capacities of a muscle–tendinous complex are a topic of continued interest to researchers. In the literature, both static stretching (Bandy et al., 1997, 1998; Bressel and McNair, 2002; Magnusson et al., 1995, 1996a,b, 1998; McHugh et al.,

1992; McNair et al., 2001) and cyclic stretching (Bressel and McNair, 2002; Magnusson et al., 1998; McNair et al., 2001, 2002) have been studied. Two mechanisms have been advanced for observed increases in range of motion following stretching. Firstly, stretch tolerance involves changes in the perception of stretch most likely through modification in the sensitivity of pain receptors (Bandy and Irion, 1994; Bandy et al., 1997, 1998; Halbertsma and Goeken, 1994; Halbertsma et al., 1999; Magnusson, 1998). Secondly, both animal (Taylor et al., 1990, 1997) and human data (Magnusson, 1998; McNair et al., 2001) have shown decreases in muscle–tendon passive tension and stiffness immediately after passive stretching.

* Corresponding author. Present address: UFR STAPS, 25 bis, bd Guy Mollet, BP 72206, 44322 Nantes Cedex 3, France.

E-mail address: christophe.cornu@univ-nantes.fr (C. Cornu).

In humans, the passive viscoelastic properties of a musculo-articular complex can be assessed using passive loading and unloading torque (T)–angle (θ) responses (Magnusson, 1998; Magnusson et al., 1998; McNair et al., 2001; Riemann et al., 2001). To date, the variables of most interest from torque–angle data have been maximal torque and angle, energy stored, together with stiffness, which is measured as the slope of the torque–angle data. For stiffness calculations, mathematical models are typically fitted to the experimental data. A number of different models could be fitted to the torque–angle data. In this respect, it appears necessary to compare such models since assessments of stiffness may be notably different across models, and influence the findings.

Stiffness can be measured across different torque or angles, and some authors (Cornu et al., 2001; Lambertz et al., 2001) have derived a stiffness index from such data using the linear relationship between stiffness and isometric torque. This technique has been confined to measures of active stiffness, and has not yet been used to evaluate the passive responses of muscle to stretch. Yet, it could be quite valuable to assess passive stiffness indices normalized with torque and/or angle to characterize muscle–tendon unit elastic behavior across the whole range of motion. This would allow an assessment of adaptations as a result of training or rehabilitation using only one parameter that is independent from a particular value of angle or passive torque.

Therefore, the objectives of this study were (i) to examine the effects of different models of fit for torque–angle data to assess passive stiffness from loading of the knee flexor muscles during knee extension; (ii) to calculate passive stiffness indices allowing the assessment of hamstring stiffness changes across passive resistive torque and knee angles; (iii) to study hamstring stiffness and stiffness indices changes induced by static stretching.

2. Material and methods

2.1. Subjects

Eight subjects (26.4 (4.4) years, height 177.8 (9.8) cm, weight 73.3 (7.2) kg) volunteered to participate in this study. All the subjects were informed of the nature and the aim of this study before they signed an informed consent form. This study was conducted according to the Helsinki Statement (1964). Subjects practiced recreational sports, but did not participate in any strength or flexibility training at the time of the study.

2.2. Measurement techniques

A Biodex system 3 Pro[®] (Biodex medical, Shirley, NY, USA) isokinetic dynamometer was used for this study. Subjects were seated and the thigh was fastened to a thigh pad using velcro straps. The trunk–thigh angle was adjusted at 60°, and the input axis of the dynamometer

was aligned with the approximate sagittal-plane axis of rotation of the knee joint. All subjects were unable to reach full knee extension in this position. All procedures began with the lower leg perpendicular to the thigh. This position was our reference angle and defined as 0°. The torque (T), joint angular position (θ) and the joint angular velocity signals (ω) were sampled at 100 Hz.

Surface electromyography (sEMG) of the medial hamstring muscles, sampled at 1024 Hz, was visualized in real time by the experimenter and the subject (Myodata Compact[®], Electronique du Mazet, Le Mazet, France). Ag/AgCl surface electrodes (Controle Graphique Medical[®], Controle graphique SA, Brie–Comte–Robert, France) were placed with an 8 mm inter-electrode distance according to SENIAM recommendations (Hermens et al., 2000). The sensor gain was adjusted in order to see sEMG activity on the screen during low-level contractions, and muscle activity was monitored to ensure that the stretching procedures were passive.

2.3. Experimental protocol

A familiarization session was performed at least 1 day before the main testing session in order to prepare the subjects for the test protocols. During the main testing session, a baseline-test was followed by a static stretching protocol, and thereafter a post intervention test was performed. In order to minimize the effects of the baseline-tests, a 15 min rest period was undertaken before the stretching intervention. Baseline and post intervention tests were similar and included the following: (i) one maximal knee range of motion (RoM) measurement. In this test, the lower leg was passively extended ($\omega = 5^\circ \text{ s}^{-1}$), and the subjects used a stop switch when they perceived the maximum tolerable hamstring muscle stretch. This point was operationally defined as maximal knee extension. The leg was then immediately returned to the starting position; (ii) five cyclic ($\omega = 5^\circ \text{ s}^{-1}$) passive repetitions to 80% of the RoM measured above. Eighty percent of the RoM was used because muscles are stretched at this point in the range motion, and furthermore, it has been shown that electromyographic responses increase significantly when muscles are stretched past this point (McNair et al., 2001).

The intervention involved a static stretching protocol of five 30 s-static hamstring stretches. Specifically, the leg was passively extended to 80% of the pre-test RoM ($\omega = 5^\circ \text{ s}^{-1}$), maintained for 30 s in this position and unloaded to the initial position ($\omega = 5^\circ \text{ s}^{-1}$). No rest period was provided between each stretching repetition.

2.4. Data analysis

All the data were processed using a standardized program computed with Matlab[®] (The Mathworks, Natick, USA). Recorded torque was gravity corrected (Aagaard et al., 1995) and data were filtered using a Butterworth second-order low pass filter (10 Hz).

Three different models were fitted to the T - θ data and the determination coefficient (R^2) between modeled and experimental curves was calculated for each passive repetition:

- (i) a second-order polynomial (SOP) model (Eq. (1)):

$$T(\theta) = a\theta^2 + b\theta + c \quad (1)$$

where a , b and c are experimental constants;

- (ii) an exponential model (Eq. (2)) similar to the Sten-Knudsen (SK) model usually used to fit stress-strain relationship of isolated muscle (Goubel and Lensele-Corbeil, 1998; Sten-Knudsen, 1953):

$$T(\theta) = \frac{A}{\alpha} (e^{\alpha\theta} - B) \quad (2)$$

where A , B and α are experimental constants;

- (iii) a fourth-order polynomial (FOP) model (Eq. (3)) classically used in the literature to assess in vivo musculo-tendinous stiffness (Magnusson et al., 1996a; Magnusson, 1998; Magnusson et al., 1998; Riemann et al., 2001):

$$T(\theta) = m\theta^4 + n\theta^3 + o\theta^2 + p\theta + q \quad (3)$$

where m , n , o , p and q are experimental constants.

The fit of experimental models were processed using optimization software based on a non-linear least squares method (Levenberg-Marquardt algorithm: Marquard, 1966). The number of data points used for each fit was about 1200.

Using both SOP and SK models, musculo-tendinous stiffness (MTS) was calculated and related to θ (Eq. (4)) and T (Eq. (5)). This enabled the calculation of two passive stiffness indices from the SOP and SK models, which were determined from the slope of the linear relationships (Eqs. (6) and (7), respectively).

$$\text{MTS}_{\text{SOP}}(\theta) = 2a\theta + b \quad (4)$$

$$\text{MTS}_{\text{SK}}(T) = \alpha T + E \quad (5)$$

$$\text{SI}_{\text{SOP}} = 2a \quad (6)$$

$$\text{SI}_{\text{SK}} = \alpha \quad (7)$$

Using the fourth-order polynomial model (Eq. (8)), stiffness (MTS_{FOP}) was calculated from the slope of the T - θ data.

$$\text{MTS}_{\text{FOP}}(\theta) = 4m\theta^3 + 3n\theta^2 + 2o\theta + p \quad (8)$$

Stiffness was also assessed at three knee angles (5°, 25° and 45°), and compared across the different models.

After checking the distribution of data, parametric statistical tests were undertaken using Statistica® software (Statsoft, Inc., Tulsa, USA). A paired t -test was done to compare the RoM between baseline and post intervention tests. Determination coefficients (R^2) were calculated for MTS calculated across all the models. Comparisons of stiffness were made using a $3 \times 2 \times 3$ (model \times test \times angle) repeated measures analysis of variance (ANOVA). In this analysis, stiffness was averaged across the five passive cyclic

stretching repetitions. Finally, two paired t -tests were used to assess the effects of the static stretching interventions on the two stiffness indices averaged across the five cyclic stretching repetitions. Newman-Keuls post-hoc analysis was conducted where appropriate. The critical level of significance in the present study was set at $P < 0.05$.

3. Results

3.1. Comparison of models

A high correlation was found between experimental data and SOP model fit (R^2 : 0.984 (0.012), range: 0.946–0.997), experimental data and FOP model fit (R^2 : 0.988 (0.011), range: 0.954–0.999), and experimental data and SK model fit (R^2 : 0.991 (0.006), range: 0.975–0.999). Stiffness calculated using the three models were highly correlated (MTS_{SOP} vs. MTS_{FOP} : $R^2 = 0.919$, $P < 0.001$; MTS_{SK} vs. MTS_{FOP} : $R^2 = 0.945$, $P < 0.001$; MTS_{SK} vs. MTS_{SOP} : $R^2 = 0.894$, $P < 0.001$) (Fig. 1).

A main effect of angle was found indicating that stiffness increased with knee angle (5° vs. 25°: 57.4%, $P < 0.05$; 25° vs. 45°: 51.8%, $P < 0.001$). Besides, a significant effect of model was found ($P < 0.001$). More precisely, for all angles, stiffness was significantly different across the models (MTS_{FOP} vs. MTS_{SOP} : 11.5%; MTS_{FOP} vs. MTS_{SK} : 13.0%, MTS_{SK} vs. MTS_{SOP} : 21.6%). The shapes of stiffness-angle curves calculated using the three models were also different, particularly for the SOP model in the beginning of the RoM (Fig. 2A). The interaction effect for model \times angle ($P < 0.001$) is shown in Fig. 2B. Significant differences were found for MTS_{FOP} vs. MTS_{SOP} and MTS_{SK} vs. MTS_{SOP} at the 5° knee angle; for MTS_{FOP} vs. MTS_{SK} and MTS_{SK} vs. MTS_{SOP} at the 25° knee angle. No significant differences were observed at the 45° joint angle. The mean differences at 5°, 25°, 45° were 95.3%, 20.1%, 2.9% for MTS_{FOP} vs. MTS_{SOP} ; 21.3%, 29.0%, 4.2% for MTS_{FOP} vs. MTS_{SK} ; and 96.1%, 6.9%, 6.8% for MTS_{SK} vs. MTS_{SOP} .

3.2. Effects of static stretching on hamstring viscoelastic properties

Knee flexion RoM significantly increased after the static stretching intervention (+11.5 (6.8)%, $P < 0.01$). Musculo-tendinous stiffness was significantly decreased after stretching (−27.6%, $P < 0.001$). An interaction effect for model \times test was found ($P < 0.05$, Fig. 3) indicating that across the three models MTS decreased by a dissimilar way after static stretching (−0.071 N m °^{−1}, −25.5% using FOP model; −0.069 N m °^{−1}, −27.8% using SOP model and −0.094 N m °^{−1}, −29.3% using SK model). An interaction effect for test \times angle ($P < 0.01$) is shown in Fig. 4. Musculo-tendinous stiffness was significantly decreased ($P < 0.001$) for each angle after static stretching, but the magnitude of the absolute decrease was larger at the end of the RoM (−0.046 N m °^{−1}, −43.2% at 5°; −0.042 N m °^{−1}, −31.7% at 25°; −0.118 N m °^{−1}, −23.7%

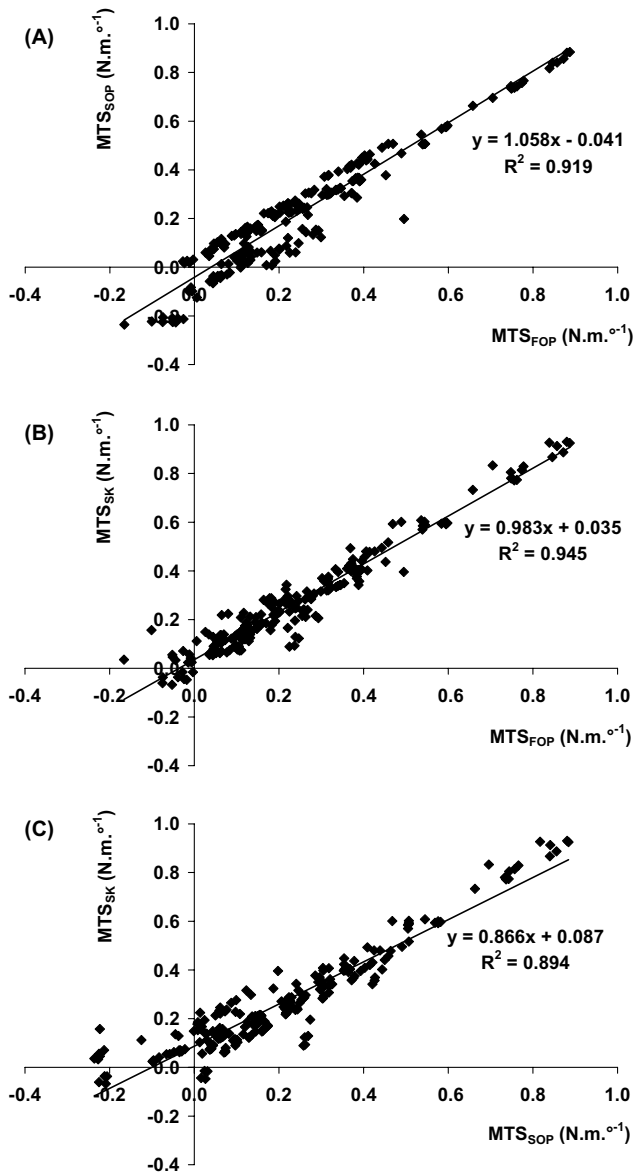


Fig. 1. Relationships between musculo-tendinous stiffness calculated using (A) fourth-order polynomial model (MTS_{FOP}) and second-order polynomial model (MTS_{SOP}); (B) fourth-order polynomial model (MTS_{FOP}) and modified Sten-Knudsen model (MTS_{SK}); (C) second-order polynomial model (MTS_{SOP}) and modified Sten-Knudsen model (MTS_{SK}).

at 45°). In respect to the stiffness indices, the stiffness index calculated with SOP model decreased (SI_{SOP} , -16.6 (9.7%), $P < 0.01$, Fig. 5A), whereas an increase in stiffness index calculated with the SK model was found (SI_{SK} , $+18.4$ (13.2%), $P < 0.05$, Fig. 5B).

4. Discussion

4.1. Comparison of models

The correlations between the experimental data and the three models used were similar or higher than those of Magnusson (1998) using the FOP model ($0.91 <$

$R < 0.99$). Moreover, our results show high correlations between MTS at given angles calculated using our two original models and the FOP model, the latter being used most often in the literature (Magnusson et al., 1996a, 1998; Magnusson, 1998; Riemann et al., 2001). A significant difference was found between stiffness calculated using the three methods, but this difference was not significant for the more extended RoM (Fig. 2B). For the more flexed angles in the RoM, the models do not fit in a similar manner leading to differences in stiffness assessments. Furthermore, the shape of stiffness curve with angle calculated using the SOP model was quite different to the other models (Fig. 2A). Therefore, it could be concluded that for low torque levels the SOP model was less appropriate. Nevertheless, stiffness differences averaged across the range of motion for SOP model vs. FOP model and for SK model vs. FOP were in the same range (11.5% and 13.0% , respectively). Overall, changes in stiffness following the static stretching intervention calculated using the three models were similar (-25.5% using FOP model; -27.8% using SOP model and -29.3% using SK model).

In the current study, utilizing the second-order polynomial and Sten-Knudsen models allowed the characterization of stiffness across the whole range of motion using only one parameter (SI). While this index has not been applied to passive torque and angle data previously, it has been used to assess the series elastic stiffness in active muscle of subjects with Duchenne Muscular Dystrophy (Cornu et al., 2001), prepubertal children vs. adult (Lambertz et al., 2003), older adults (Ochala et al., 2004), and in the examination of the long term effects of space flight (Lambertz et al., 2001). Classically, passive stiffness is assessed at a given angle (Magnusson et al., 1996b; Magnusson, 1998) or a given percentage of the RoM (Magnusson et al., 1998; McNair et al., 2001). However, interventions such as a stretching programme may change torque levels and angles and hence make it difficult to decide at what point (angle or torque or percentage of either) should the subject be measured after the intervention. To avoid those problems, the stiffness index allows the assessment of changes in stiffness independent of angle or torque changes.

4.2. Effects of static stretching on hamstring stiffness

Most often, the purpose of stretching is to increase range of motion, reduce tension and stiffness of the muscle-tendon unit (Magnusson et al., 1998; McNair et al., 2001; Witvrouw et al., 2004). Our results are for the most part in accordance with these thoughts. In respect to the effects of static stretching, our results are consistent with some studies that have found a decrease in stiffness at a given angle whatever the model (Magnusson et al., 1996a; Magnusson, 1998). The stiffness index calculated with the SOP model (SI_{SOP}) was also found to decrease. Interestingly, the stiffness index calculated with the SK model increased after static stretching. This discrepancy

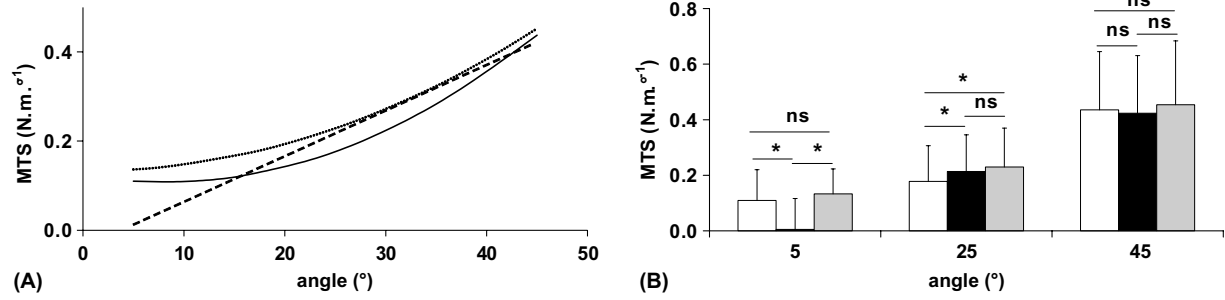


Fig. 2. (A) The shapes of the stiffness-angle relationships calculated using the fourth-order polynomial model (continuous line), the second-order polynomial model (dashed line) and the modified Sten-Knudsen model (dotted line); (B) stiffness across the three models at different knee joint angles calculated with a fourth-order polynomial model (□), a second-order polynomial model (■) and the modified Sten-Knudsen model (▒). * $P < 0.05$, ns: $P > 0.05$.

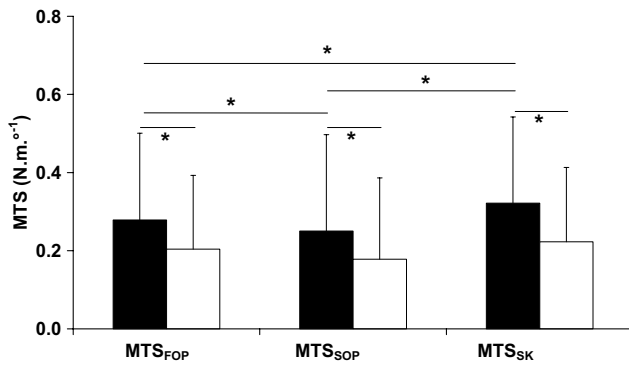


Fig. 3. Comparison of musculo-tendinous stiffness (MTS) averaged across knee angles calculated before (pre-test, ■) and after (post-test, □) static stretching using the fourth-order polynomial model (MTS_{FOP}), the second-order polynomial model (MTS_{SOP}) and the exponential model (MTS_{SK}). * $P < 0.05$.

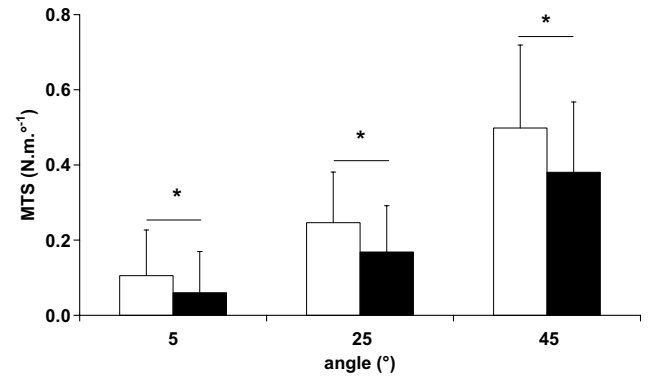


Fig. 4. Musculo-tendinous stiffness (MTS) calculated using the three models before (□) and after (■) the static stretching intervention at the three knee angles. * $P < 0.05$.

4.3. Conclusion and perspective

Three mathematical models were used in the present study to assess effects of stretching on viscoelastic properties of the musculo-tendon unit. They have enabled assessments of specific musculo-tendinous stiffness changes after static stretching. Stiffness calculated with these models was significantly different and particularly so for the second-order polynomial model at low torque levels. However,

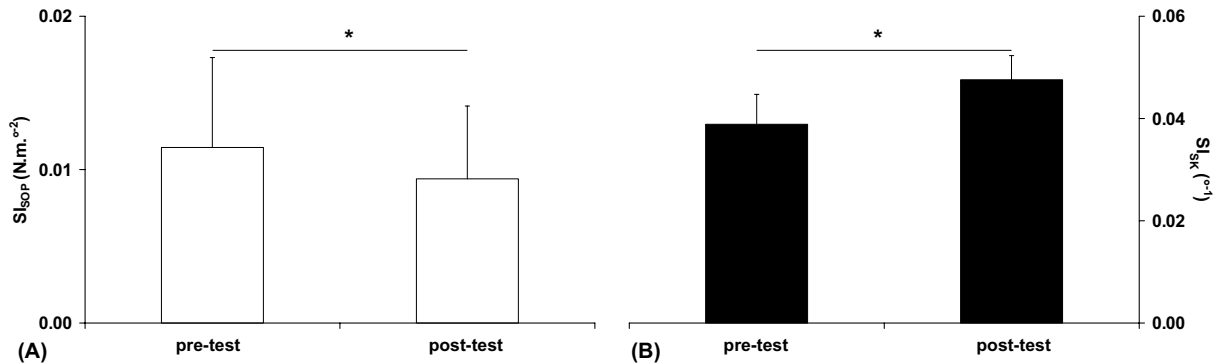


Fig. 5. (A) Stiffness index calculated using the second-order polynomial model (SI_{SOP}) before (pre-test) and after (post-test) the static stretching intervention; (B) stiffness index calculated using the modified Sten-Knudsen model (SI_{SK}) before (pre-test) and after (post-test) the static stretching intervention. * $P < 0.05$.

the three models provided more consistent stiffness measurements at the end of the RoM. Two different stiffness indices were calculated, and they provided dissimilar results in respect to the effects of stretching. Since changes in stiffness after stretching are different across the models, this study highlights the need to carefully assess the effect of different models when analyzing these data. At this time, no one model can be recommended exclusive of others. Further study is needed to identify those parameters which most accurately provide a measure of a particular model's validity.

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