



Prestress revealed by passive co-tension at the ankle joint

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ABSTRACT

This study was designed to test the assumption that elastic tissues of the ankle are prestressed, by investigating the presence of simultaneous opposite passive elastic moments and thus, passive co-tension, at the ankle joint. A prestressed two-spring model used to generate qualitative predictions of the effects of stretching the posterior elastic structures of the ankle on the net passive moment of this joint was used. Twenty-seven healthy individuals were subjected to passive evaluation of the net elastic moment of the ankle in the sagittal plane, with the knee positioned at 90°, 60°, 30° and 0° of flexion, in order to change the length of the posterior biarticular elastic structures. The placement of the knee in the more extended positions caused changes in the net passive moment as predicted by the prestressed model. The ankle position in which the net passive moment was equal to zero was shifted to more plantar flexed positions ($p < 0.001$) and there was a global increase in ankle stiffness since both passive dorsiflexion stiffness ($p \leq 0.037$) and passive plantar flexion stiffness ($p \leq 0.029$) increased. The normalized terminal plantar flexion stiffness also increased ($p \leq 0.047$), suggesting that biarticular posterior elastic structures are pre-strained and still under tension when the ankle is maximally plantar flexed and the knee is positioned at 60° of flexion. Resting positions were indicative of equilibrium between opposite passive elastic moments. The results revealed that there is passive co-tension at the ankle, demonstrating the existence of prestress in elastic structures of this joint.

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1. Introduction

The passive stability of synovial joints is used by the motor system to meet the demands of functional activities as it reduces control complexity and metabolic cost of active stability (Dickinson et al., 2000; Moritz and Farley, 2004). Passive stability is partly attributable to the nonlinear spring-like properties of the soft tissues (Loram et al., 2007).

Currently accepted models of joint stability often assume the existence of a range of motion around the joint position in which the net moment is zero, within which there would be negligible or no tension in the elastic structures (Crawford et al., 1998; Panjabi, 2003). However, positions with complete absence of passive elastic moments were observed only in *in vitro* studies in which muscle–tendon units and fascial connections of a given joint had been removed (Oda et al., 1992; Thompson et al., 2003). *In vivo*, joint motion takes place under passive moments throughout the range of motion and the net passive moment equals zero and

changes direction only at one specific joint position (McClure et al., 1998; Silder et al., 2007). In absence of concurrent opposite passive moments, this position would have to correspond to the slack lengths of both antagonistic groups of elastic structures (De Monte et al., 2006; Muraoka et al., 2002). This behavior is unlikely since every elastic structure and their subcomponents have their own slack lengths (Zajac, 1989). The complex architecture of muscle–tendon units and their intra-, inter- and extra-muscular fascial connections make these slack lengths dependent on each other (Epstein et al., 2006; Yucesoy et al., 2003). Distinct elastic structures of a synergistic group start developing tension at different joint positions and some of them may exert moments before the joint position in which the antagonistic elastic structures become slack. Thus, there will be passive co-tension, i.e. antagonistic elastic structures will be concurrently under tension and exert simultaneous opposite moments within the joint range of motion. The joint position in which the net elastic moment is zero is probably the position where these opposite moments are equal and offset each other (Fig. 1).

The possible existence of passive co-tension in joints leads to the hypothesis that the musculoskeletal system is a prestressed system. Prestress arises from pre-existing forces in components of a system, i.e. forces that are present when such a system is in its resting position (before it is subjected to additional forces

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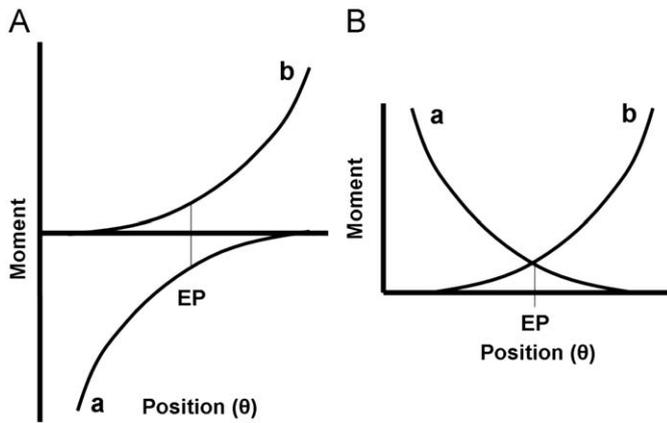


Fig. 1. Schematic representation of the passive moments produced by antagonistic groups of elastic structures (*a* and *b*), as a function of joint position (θ). Opposite moments *a* and *b* would be simultaneous in part of the range of motion, generating an equilibrium position (*EP*) where the net moment (not shown) equals zero. Graph B shows absolute values of the moments in order to facilitate visualization of the *EP* as the position where moment curves cross each other.

generated externally or internally) (Skelton et al., 2001; Sultan and Skelton, 2003). Prestress can arise due to the presence of passive tension in elastic components, such as myofascial structures, tendons and ligaments, configured with lengths beyond their slack lengths (pre-stretched), generating greater system stiffness compared to the absence of prestress (Skelton et al., 2001). Prestress also produces an equilibrium position, which would correspond to the joint's resting position, causing the joint to passively return to this position after a transient disturbance (Connelly and Back, 1998). Consequently, prestressed systems have intrinsic stability. Since musculoskeletal joints are systems with antagonistic tissues, prestressed elastic structures would create passive co-tension underlying the joints net passive moments and stiffnesses.

The present study tested the hypothesis that the net passive moment of a joint is resultant from simultaneous opposite elastic moments produced by prestress in the elastic tissues related to the ankle joint. By means of motions at the knee joint, the length of posterior structures of the ankle is changed without changing the length of the antagonistic structures. Qualitative predictions of the effects of stretching elastic structures of a synergistic group on the net passive moment of the ankle were generated by a prestressed two-spring model.

2. The prestressed two-spring model

This model consists of two nonlinear ideal springs attached in series to opposite sides of a massless body. The springs are modeled as elastic cables and thus, cannot produce compressive forces below their resting lengths (slack lengths). The springs represent the antagonistic groups of elastic structures of a joint, their resultant force is related to the net passive elastic moment, and the linear body position represents the joint angular position. Prestress is included in the model by stretching each spring the same arbitrary amount, shifting their slack lengths farther from the body mid position (Fig. 2).

The behavior of each spring has been modeled using a quadratic function according to the nonlinear behavior of elastic forces produced by passively stretched musculoskeletal soft tissues (Voigt et al., 1995), and the resultant elastic force of the system is calculated as the sum of the springs' forces

$$F_e = k(\Delta l)^2 \quad (1)$$

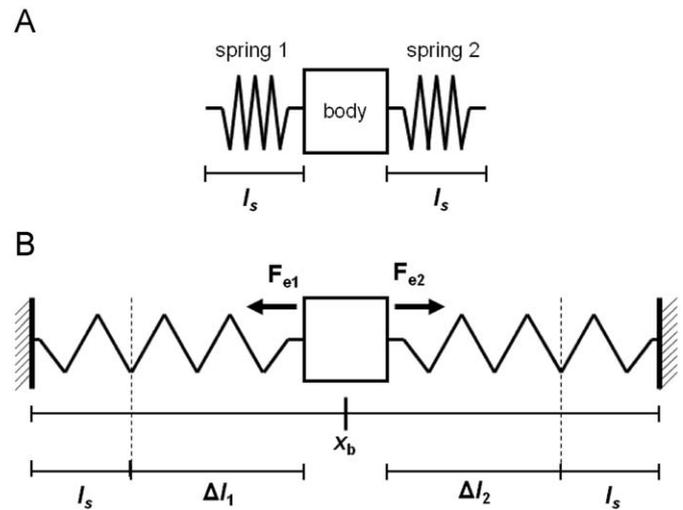


Fig. 2. The prestressed two-spring model of the joint passive elastic properties. (A) The model before inclusion of prestress, where the springs are at their slack lengths (l_s); (B) prestressed model where the springs are pre-stretched (before the body is subjected to any displacement). The stretching of the springs shifted their slack lengths farther from the body mid position and hence, the springs become slack when body displacement reaches their slack lengths (dashed lines). F_{e1} and F_{e2} : elastic forces of springs 1 and 2, respectively. Δl_1 and Δl_2 : deformations of springs 1 and 2, respectively. x_b : body positions within the range of displacement.

$$F_R = \sum_{i=1}^2 F_{ei} \quad (2)$$

where F_e is the elastic force produced by each spring, k their elastic modulus, Δl the deformation of each spring, and F_R the resultant elastic force in the system. The springs have equal arbitrary values for k . Since the forces exerted by the springs have opposite directions, F_R is resultant from subtraction between the springs' forces. The system forces are shown in Fig. 3.

3. Model behavior and predictions

In the model, an additional arbitrary stretching of spring 2 generated changes in the behavior of the resultant force, due to changes in the interaction between the nonlinear springs' forces (Fig. 4). The position where the resultant force is zero (equilibrium position) was shifted toward the spring 2. Moreover, there was a global increase of the system stiffness, represented by the increases in the slopes of the resultant force–position curve. Thus, unlike a slack system, stiffness was affected in both directions, even without changes in the parameters of the spring 1.

According to the changes in the resultant force of the prestressed model, qualitative predictions were generated for changes in the net passive elastic moment of the ankle, caused by placing the knee in more extended positions to progressively stretch ankle posterior structures: (a) the position in which the net passive elastic moment is zero (estimated resting position) will shift to a more plantar flexed position; (b) the dorsiflexion stiffness, at positions more dorsiflexed than the estimated resting position, will increase; and (c) the plantar flexion stiffness, at positions more plantar flexed than the estimated resting position, will also increase. Dorsiflexion stiffness is defined as the rate of increase in the ankle plantar flexion moment and plantar flexion stiffness is defined as the rate of increase in the ankle dorsiflexion moment. According to the prediction (c), stiffness increase during terminal plantar flexion would suggest that there is co-tension in this position (the posterior structures of the ankle do not become

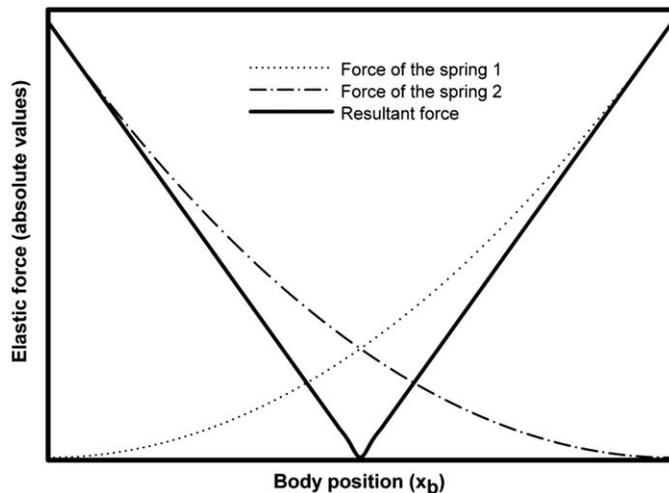


Fig. 3. Forces (absolute values) generated by the prestressed two-spring model at the body positions where the springs are co-tensioned and there are concurrent opposite forces. The elastic forces of the spring 1 (F_{e1}) and of the spring 2 (F_{e2}) represent the opposite moments produced by the antagonistic groups of elastic structures of a joint. The solid line is the resultant force (F_R), with a single body position where its value equals zero (equilibrium position), and represents the net passive moment of a joint. Absolute values are shown in order to facilitate visualization of the equilibrium position as the body position where the F_R curve “touches” the abscissa. Notice the greater stiffness (slope) of the resultant force in comparison to the stiffness of each spring. The linearity of the resultant force is due to the fact that the springs’ forces were determined by identical functions, which is unlikely for the asymmetric antagonistic elastic structures of joints *in vivo*. The linearity obtained does not influence the qualitative predictions of the model.

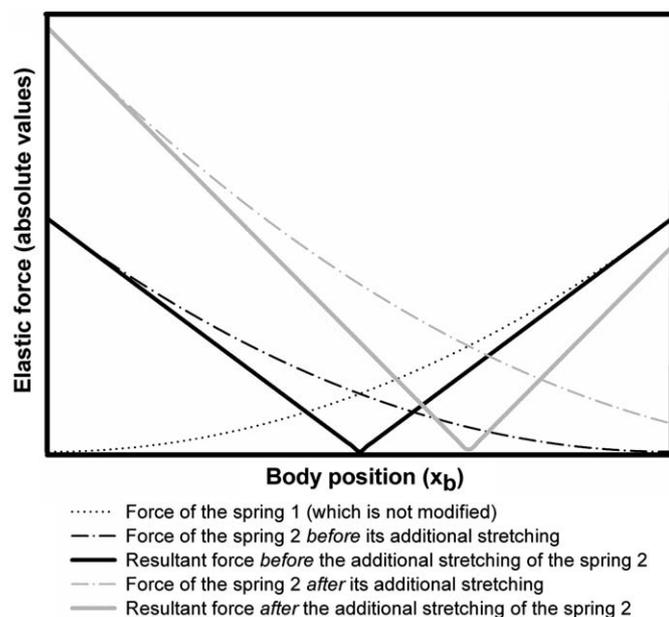


Fig. 4. Elastic forces of the prestressed system, before and after the additional stretching of the spring 2. The increases in the force produced by the spring 2 and in the slope of this force shift the equilibrium position toward the spring 2. The force of the spring 2 becomes greater than the force of the spring 1 within a larger number of positions, producing a greater range of body displacement within which the resultant force have the same direction of the force of the spring 2. The slopes of the springs’ forces at the new equilibrium position are greater (due to nonlinearity), resulting in a larger rate of change of the values corresponding to the subtraction between the springs’ forces. This last change increases the slope of the resultant force in both directions, indicating a global increase of the system stiffness. Notice that the force of the spring 1 was not modified.

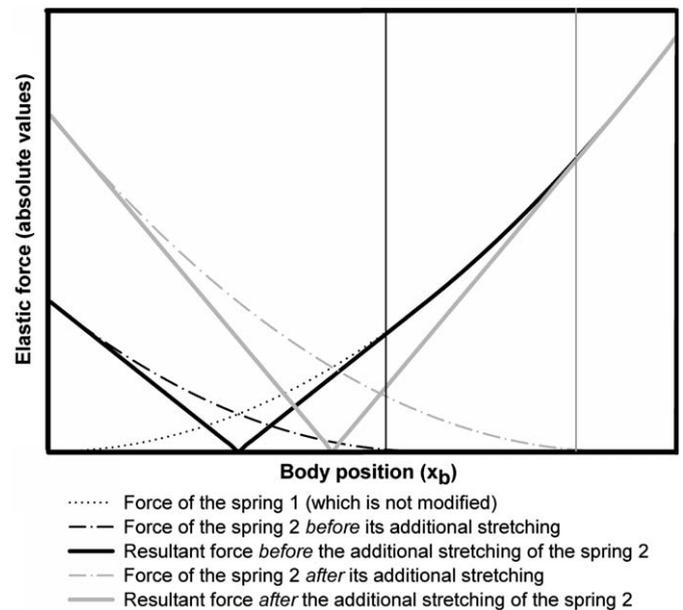


Fig. 5. Elastic forces of a prestressed system whose spring 2 becomes slack in the terminal body positions on the right side of the graph, both before and after the additional stretching of the spring 2. At the positions in which the spring 2 is slack (on the right side of the vertical lines), the resultant forces become equal to and superimpose the individual force of the spring 1. At these positions, the system stiffness is equal to the stiffness of the spring 1. Further, the absence of tension in the spring 2 and of co-tension causes the additional stretching of this spring not to result in increases in system stiffness at the terminal positions on the right side of the figure (differently to the prestressed system of the Fig. 4). In this sense, if the amount of stretching is sufficient to increase terminal system stiffness, it would indicate that this stretching caused the spring 2 to change from slack to tensioned in these positions.

slack in any joint position) (Fig. 4). In contrast, the absence of stiffness increases in this range would suggest that the posterior structures are not under tension at the terminal plantar flexion positions (Fig. 5).

4. Methods

4.1. Subjects

Twenty-nine young healthy subjects volunteered to participate in the study. During data collection, two subjects were excluded from the study due to inability to maintain the ankle muscles relaxed (according to the definition of muscle contraction described in the procedures). The data collected from the remaining 27 subjects (14 male, 13 female) were considered in the study. Their mean (\pm SD) age, mass and height were 23 ± 3.15 years, 62.85 ± 13.44 kg, and 169 ± 11 cm, respectively. The ankle passive range of motion of the dominant limb, in the sagittal plane, was $57.66 \pm 10.6^\circ$, with a maximum dorsiflexion of $14.66 \pm 6.73^\circ$ and a maximum plantar flexion of $43 \pm 7.32^\circ$, measured with the knee in full extension. Exclusion criteria were the presence of symptoms or history of injuries in the dominant lower limb. The participants signed an informed consent and this work was approved by the institution’s Ethics in Research Committee.

4.2. Procedures

Initially, maximum passive dorsiflexion and plantar flexion angles at the ankle were measured in the dominant limb of each subject, with the knee in full extension, using a universal

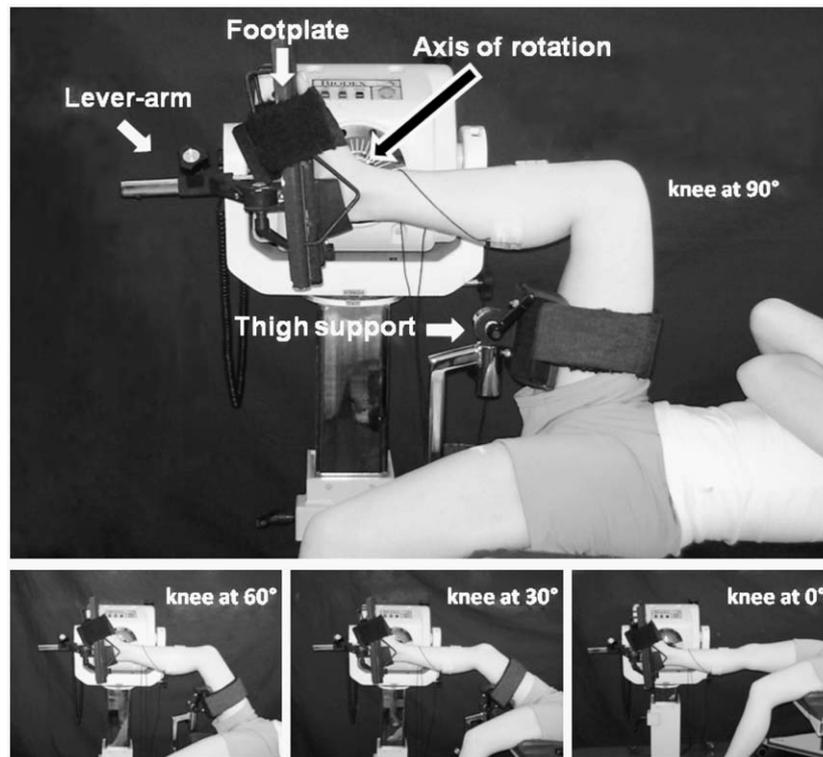


Fig. 6. Experimental setups for the tested knee conditions.

goniometer. Maximum positions were defined as the last ankle positions in which the subjects did not report discomfort related to tissue stretching. Subjects were then placed in supine position, barefoot, on an isokinetic dynamometer (Biodex System 3 Pro, Shirley, USA) with the lateral malleolus (as an estimate of the ankle axis of rotation) aligned with the axis of rotation of the equipment (Fig. 6). The foot was tightly fixed with a velcro® strap to the dynamometer's footplate and the thigh was supported to enable the subject to stay as relaxed as possible. The shank of the same limb was positioned horizontally, using an analog inclinometer, and maintained in this position during the procedure. The neutral position of the ankle (0°) was defined as an angle of 90° between the footplate and the shank, also measured with the inclinometer.

The ankle was passively moved by the dynamometer from maximum dorsiflexion to maximum plantar flexion, with a constant velocity of $5^\circ/\text{s}$. The moment acting on the axis of rotation was registered with a frequency of 100 Hz. This unidirectional motion was used in order to obtain single-line curves of the moment–position relationship, which were qualitatively comparable to the curve of the model's resultant force. The constant low velocity prevented inertial forces in the subject–equipment system, in accordance with the massless body of the model. One series of three repetitions was passively performed with the knee joint positioned at 90° , 60° , 30° and 0° of flexion. Before each series, a 5-repetition pre-conditioning series was performed to maximally reduce tissue viscosity and to permit attributing ankle passive moments mostly to tissues elasticity (Hoang et al., 2007). The subjects were asked not to resist or help ankle motion voluntarily. None of the subjects reported any discomfort during the trials.

Surface electromyography of the medial gastrocnemius, soleus and tibialis anterior were monitored to ensure that these muscles were relaxed during the tests. The signal was initially registered with the subjects lying supine, completely relaxed. After each repetition, data were processed using the software Matlab (The

Mathworks, Natick, MA), which enabled the identification of muscle contractions, defined as activity amplitudes equal to or greater than two standard deviations of the signal of the same muscle relaxed (Lamontagne et al., 1997). Repetitions with contractions were rejected and a new trial was performed. Electromyographic data were collected at a frequency of 1000 Hz with active electrodes and the MP100 system (Biopac System, Goleta, USA), and band-pass filtered with cut-off frequencies of 10 and 500 Hz.

4.3. Data reduction

The moment registered by the dynamometer was corrected for the moment generated by the weights of the lever-arm and footplate, measured after the tests of each subject with the same device. The resultant moment was further corrected by the moment produced by the weight of the foot, which was calculated based on anthropometric data (Winter, 2005). The final resultant moment represented the passive elastic moment of the ankle in the sagittal plane. Dependent variables were calculated from the moment–position relation: (1) ankle position, in degrees, where the net moment is zero ($\theta_{\text{moment-zero}}$); (2) passive dorsiflexion stiffness, in Nm/rad , calculated as the mean slope ($\Delta\text{Nm}/\Delta\text{rad}$) (Latash and Zatsiorsky, 1993) of the moment–position curve at positions more dorsiflexed than the $\theta_{\text{moment-zero}}$; (3) passive plantar flexion stiffness, in Nm/rad , calculated as the mean slope of the moment–position curve at positions more plantar flexed than the $\theta_{\text{moment-zero}}$; (4) normalized terminal passive plantar flexion stiffness, in Nm/rad^2 , calculated as the mean slope of the moment–position curve in the final 5° of plantar flexion, divided by the excursion corresponding to the positions more plantar flexed than the $\theta_{\text{moment-zero}}$. The mean slope of each repetition, used for calculating the dependent variables (2), (3) and (4), was defined as the mean value of the multiple point-by-point slopes obtained between each two subsequent points of the

moment–position curve (Bressel et al., 2004), at intervals of 0.05° . Finally, mean values of the dependent variables, resulting from the three test repetitions conducted at each knee position, were used for statistical analyses.

Since $\theta_{\text{moment-zero}}$ and thus, curve lengths on the right and on the left sides of the $\theta_{\text{moment-zero}}$ in the abscissa, could be modified among knee positions, caution had to be taken for the variables definition. The length of a nonlinear curve, from its origin to its end, influences the value of its mean slope and slope at its terminal positions. A longer nonlinear curve has greater mean and terminal slopes than a shorter nonlinear curve (defined by the same function), since the former curve has a steeper additional part at its end. Hence, the lengths of the curves, from the $\theta_{\text{moment-zero}}$ to maximum dorsiflexion or maximum plantar flexion, were taken into consideration when defining the dependent variables (2), (3), and (4). The dorsiflexion stiffness was calculated considering the smallest excursion, from the $\theta_{\text{moment-zero}}$ to maximum dorsiflexion, of each subject, obtained among all knee positions. The plantar flexion stiffness was calculated considering the smallest excursion, from the $\theta_{\text{moment-zero}}$ to maximum plantar flexion, of each subject, registered among all knee positions. Therefore, the curve lengths (beginning from the $\theta_{\text{moment-zero}}$ s) considered for calculation of dorsiflexion and plantar flexion stiffness for each subject were standardized. Finally, considering the influence of the origin-to-end length of a nonlinear curve on its terminal slope, the terminal plantar flexion stiffness was normalized by the length of the moment–position curve (from the $\theta_{\text{moment-zero}}$ to maximum plantar flexion) of each knee position. Therefore, it was possible to remove the influence of curve length changes and to identify only the influence of knee extension on these dependent variables.

4.4. Statistical analyses

Repeated measures analyses of variance (ANOVA) with one factor (knee position) and four levels (90° , 60° , 30° , 0°) were carried out for each dependent variable. Pre-planned Contrasts (90° vs. 60° , 60° vs. 30° , 30° vs. 0°) were the pairwise comparisons used to test for specific differences, when significant differences were identified for the ANOVAs' main effects. The significance level (α) was set at 0.05 (Keppel and Sheldon, 1989).

Intraclass correlation coefficients ($ICC_{3,1}$), corresponding to the between-repetition reliabilities, were determined to enable further calculation of the standard error of measurement (SEM) of each dependent variable.

5. Results

Main effects for all variables were significant ($p < 0.001$). Pre-planned Contrasts revealed that the placement of the knee in more extended positions significantly shifted the $\theta_{\text{moment-zero}}$ toward plantar flexion ($p < 0.001$) (Fig. 7). In addition, passive dorsiflexion stiffness ($p \leq 0.037$), passive plantar flexion stiffness ($p \leq 0.001$) and normalized terminal passive plantar flexion stiffness ($p \leq 0.047$) increased significantly with increased knee extension (Fig. 8). Means, standard deviations and p values are shown in Table 1. ICCs varied from 0.95 to 0.99, showing excellent between-repetitions reliabilities. SEM values varied from 0.4 to 2.06 and the estimates of error ranges were smaller than the corresponding mean differences, for each variable, demonstrating that the differences were not due to possible inherent measurement errors (Portney and Watkins, 2000).

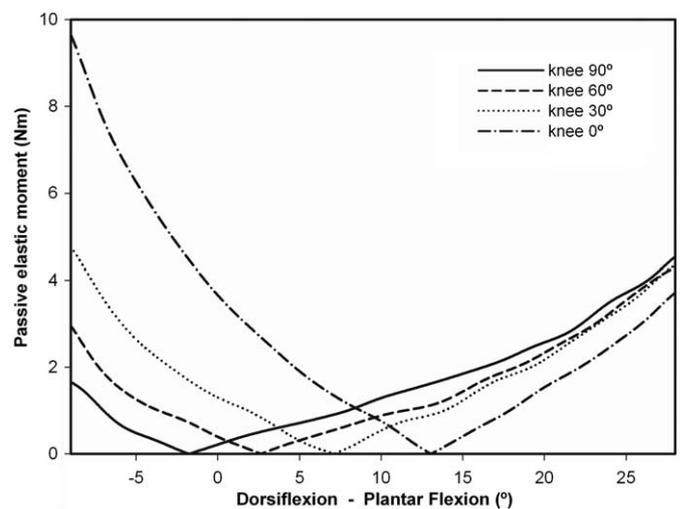


Fig. 7. Ankle net passive moments (absolute values) in relation to ankle joint position of a representative subject, showing the $\theta_{\text{moment-zero}}$ s (estimated resting positions in which the net passive elastic moments equal zero) and the slopes (estimated stiffness) of the moment–position curves. Each curve was obtained in one of the three repetitions carried out with the knee at each tested position. In each curve, the moment on the left side of the $\theta_{\text{moment-zero}}$ is plantar flexor; whereas the moment on the right side of the $\theta_{\text{moment-zero}}$ is dorsiflexor.

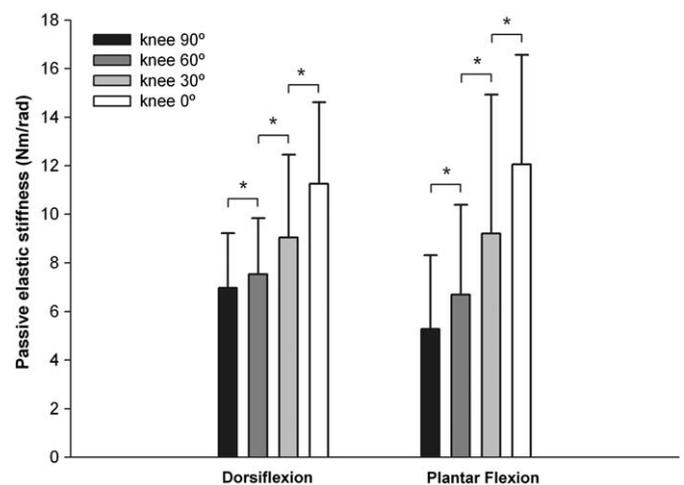


Fig. 8. Global increase in system stiffness caused by the placement of the knee in more extended positions, demonstrated by increases in both passive dorsiflexion stiffness (as usually expected) and passive plantar flexion stiffness (counter-intuitively for traditional assumptions), respectively, considering the corresponding $\theta_{\text{moment-zero}}$ s. * $p < 0.05$ demonstrated by the Pre-planned Contrasts.

6. Discussion

The changes observed in the passive mechanical behavior of the ankle joint are in accordance with the predictions of the prestressed two-spring model, demonstrating that the ankle net passive moment in the sagittal plane is, at least partly, resultant from simultaneous opposite elastic moments. The $\theta_{\text{moment-zero}}$ s (estimated resting positions without influence of external forces) correspond to the equilibrium of pre-existing opposite passive moments and not to lack of tension. Such findings indicate that the ankle is a prestressed joint and thus antagonistic elastic structures are co-tensioned in part of the ankle joint range of motion. Studies that attribute ankle resistance moment only to the elastic structures which are antagonistic to the unidirectional

Table 1Means (\pm SD) of the parameters of ankle net passive elastic moment and *p* values of the comparisons among knee positions.

	Knee 90	<i>p</i>	Knee 60°	<i>p</i>	Knee 30°	<i>p</i>	Knee 0°
$\theta_{\text{moment-zero}}$ (°) ^a	8.87 \pm 5.3	<i>p</i> < 0.001	12.26 \pm 5.42	<i>p</i> < 0.001	15.96 \pm 5.61	<i>p</i> < 0.001	20.12 \pm 6.54
Dorsiflexion stiffness (Nm/rad)	6.97 \pm 2.24	<i>p</i> = 0.037	7.53 \pm 2.31	<i>p</i> = 0.003	9.04 \pm 3.4	<i>p</i> < 0.001	11.26 \pm 3.35
Plantar flexion stiffness (Nm/rad)	5.27 \pm 3.04	<i>p</i> < 0.001	6.69 \pm 3.7	<i>p</i> < 0.001	9.2 \pm 5.73	<i>p</i> = 0.001	12.05 \pm 4.51
Normalized terminal plantar flexion stiffness (Nm/rad ²)	23.48 \pm 14.68	<i>p</i> < 0.001	27.22 \pm 15.82	<i>p</i> = 0.037	31.74 \pm 21.43	<i>p</i> = 0.047	37.64 \pm 19.78

p values of the Pre-planned Contrasts, corresponding to the comparisons between one knee position and the subsequent more extended knee position. The level of significance was set at 0.05.

^a Positive values indicate plantar flexed positions.

angular displacement analyzed (De Zee and Voigt, 2002; Kubo et al., 2001) should consider that this moment may also be influenced by the elastic structures that produce moments in the direction of the displacement. It is usually expected that changes in knee position affect only dorsiflexion stiffness as a result of tension changes in biarticular posterior structures (Silder et al., 2007); however, as demonstrated by the present study, there is a global stiffness change since plantar flexion stiffness was also modified (Fig. 8), considering the new $\theta_{\text{moment-zero}}$ s as references. These changes are known responses of prestressed mechanical systems when the tension of one or more of their elastic structures is asymmetrically modified (Skelton et al., 2001).

The changes studied were produced by tension increases only in the ankle elastic structures that are affected by knee position. These changes occurred in all modifications of knee position, including the modification from 90° to 60° of flexion. Thus, inferences can be made for the ankle only when the knee is at 60° of flexion or at more extended positions. The increases in the normalized terminal plantar flexion stiffness suggest that the posterior structures are still under tension in the final plantar flexion positions, since the tension of the ankle anterior elastic structures was not modified (Fig. 4). These findings indicate a large pre-strain. The ankle posterior elastic structures are under tension even when the ankle is maximally plantar flexed with the knee at 60° of flexion. This finding is comparable to those of Hoang et al. (2007), who demonstrated that the gastrocnemius muscle starts developing tension at knee positions close to 100°. Conversely, Muraoka et al. (2005) found that this muscle becomes tensioned when the knee is at 43° and the ankle is at 10° of dorsiflexion. These distinct results may be due to different methodology (e.g. ankle position) and/or to the possible inter-subjects variability related to tissue properties.

The passive co-tension resulting from prestress produces higher stiffness than the stiffness of each elastic component (Fig. 3) and may be exploited to accomplish active joint stability. This property might contribute to the production of the net passive joint stiffness without requiring each structure to have higher stiffness by possessing, for example, greater cross-sectional area and mass (Chleboun et al., 1997). As the ankle net moment is plantar flexor from the $\theta_{\text{moment-zero}}$ to maximum dorsiflexion, indicating predominant tension in posterior structures, and this tension is present until maximum plantar flexion, the posterior structures are under tension within the entire range of motion. Therefore, there is co-tension in the range within which it is possible to identify tension in the anterior structures. The dorsiflexor portion of the net moment curve indicates that these structures exert tension at least from the $\theta_{\text{moment-zero}}$ to maximum plantar flexion. Thus, considering the more dorsiflexed $\theta_{\text{moment-zero}}$ (observed with knee at 90°), it can be inferred that identifiable co-tension exists from 9° of plantar flexion to the maximum plantar flexion position (43°), approximately. The extent of co-tension toward dorsiflexion was not estimated. Even so, the identified range of co-tension at the ankle and the knee range studied are used during common activities (Novacheck, 1998).

Traditional models of joint stability do not consider myofascial tissues as part of the elastic structures that provide passive stability, which may mislead to the idea of intermediary ranges of motion without tissue tension, within which muscles would contribute only actively to stability generation (Crawford et al., 1998; Panjabi, 2003). The prestress which arises from the presence of myofascial tissues and connections may partly explain the contrast between the continuous intersegmental passive moments observed in the spine *in vivo* (Hasegawa et al., 2008) and the intermediary ranges of motion with no tissue tension observed only in spinal joints *in vitro* (with no myofascial tissues) (Oda et al., 1992). Even some *in vitro* biomechanical analyses did not find ranges of motion with complete tissue slackness, suggesting that isolated ligaments may also hold some amount of pre-tension (Heuer et al., 2007; Novotny et al., 1998). Further, the intricate fascial connections (Huijing, 1999; Yucesoy et al., 2003) are usually not considered in musculoskeletal models (Zajac, 1989). Passive properties of isolated muscles *in vitro*, in the absence of these connections, are often assumed (Hoy et al., 1990). A property commonly accepted is that each muscle in the system starts developing passive tension at the length (joint position) where this muscle exerts maximum active isometric force (optimum length) (Zajac, 1989), as observed *in vitro* (Woittiez et al., 1983). Hoy et al. (1990), for instance, calculated model-based estimates of the ankle position in which the gastrocnemius exerts maximum active isometric moment and found angles close to 15° degrees of plantar flexion. This would be an estimate of the joint position where this muscle starts developing passive tension; however, this conclusion is questionable according to our results. Mechanical properties originated from the *in vivo* architectural complexity, should be addressed to appropriately study the role of passive properties on joint stability and movement coordination. The inclusion of prestress (estimates of co-tension) in biomechanical models may reveal greater contributions of passive stiffness to production of joint stability than commonly found. It would emphasize the intrinsic stability of the musculoskeletal system, usually underestimated by theoretical models of stability control (Hodges, 2000).

Conflict of interest statement

The authors disclose that there is not any conflict of interest, which could have biased this work.

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