

The Effects of Sex, Joint Angle, and the Gastrocnemius Muscle on Passive Ankle Joint Complex Stiffness

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Objective: To assess the effects of sex, joint angle, and the gastrocnemius muscle on passive ankle joint complex stiffness (JCS).

Design and Setting: A repeated-measures design was employed using sex as a between-subjects factor and joint angle and inclusion of the gastrocnemius muscle as within-subject factors. All testing was conducted in a neuromuscular research laboratory.

Subjects: Twelve female and 12 male healthy, physically active subjects between the ages of 18 and 30 years volunteered for participation in this study. The dominant leg was used for testing. No subjects had a history of lower extremity musculoskeletal injury or circulatory or neurologic disorders.

Measurements: We determined passive ankle JCS by measuring resistance to passive dorsiflexion ($5^\circ \cdot s^{-1}$) from 23° plantar flexion (PF) to 13° dorsiflexion (DF). Angular position and torque data were collected from a dynamometer under 2 conditions designed to include or reduce the contribution of the gastrocnemius muscle. Separate fourth-order polynomial equations relating angular position and torque were constructed for each trial. Stiffness values ($Nm \cdot degree^{-1}$) were calculated at 10° PF, neutral (NE), and 10° DF using the slope of the line at each respective position.

Results: Significant condition-by-position and sex-by-position interactions and significant main effects for sex, position, and condition were revealed by a 3-way (sex-by-position, condition-by-position) analysis of variance. Post hoc analyses of the condition-by-position interaction revealed significantly higher stiffness values under the knee-straight condition compared with the knee-bent condition at both ankle NE and 10° DF. Within each condition, stiffness values at each position were significantly higher as the ankle moved into DF. Post hoc analysis of the sex-by-position interaction revealed significantly higher stiffness values at 10° DF in the male subjects. Post hoc analysis of the position main effect revealed that as the ankle moved into dorsiflexion, the stiffness at each position became significantly higher than at the previous position.

Conclusions: The gastrocnemius contributes significantly to passive ankle JCS, thereby providing a scientific basis for clinicians incorporating stretching regimens into rehabilitation programs. Further research is warranted considering the cause and application of the sex-by-position interaction.

Key Words: muscle, pathology, rehabilitation, range of motion, flexibility

Functional joint stability (FJS), the quality of possessing adequate joint stability to enable normal performance of a joint during functional activity, arises from complementary relationships existing between static and dynamic components.¹ Collectively, both components serve to maintain FJS throughout the physiologic range of motion by resisting potentially destabilizing forces.¹ An enhanced ability to diffuse destabilizing forces results in augmented FJS.^{2–4} Stiffness is the mechanical property that determines how effectively external forces delivered to the skeletal system are absorbed or transmitted (or both) by the articular soft tissues.^{5,6} In contrast to *muscle stiffness*, which describes the stiffness properties specifically exhibited by the tenomuscular tissues, *joint stiffness* encompasses contributions from all structures located within and over the joint (muscles, tendons, skin, subcutaneous tissue, fascia, ligaments, joint capsule, and cartilage).^{1,7,8} Because dynamic restraints are included in addition to static restraints, joint stiffness becomes a function of not only the passive factors (ie, viscoelasticity)

associated with each structure but also the level of neural influence over each articular muscle.^{2,5,6,8–11} Neural influences can be considered to exist intrinsically, represented by the level of muscle activation (number of actin-myosin cross-bridges) existing at an instant,^{2,5,10,12–15} and extrinsically, represented by the arrival of a reflexive activation in response to a sensory stimuli.^{2,10,16} Superimposed on these neural influences are the factors of single muscle fibers (ie, sarcomere length-tension and force-velocity relationships) as well as whole muscles (ie, arrangement of muscle fibers within a muscle and precise location of insertional sites).⁹

From an engineering perspective, stiffness may be defined in terms of elasticity, viscosity, friction, inertia, and plasticity.^{7,8} In normal finger and knee joints measured with passive arthrography, elastic stiffness has been credited as the largest stiffness component (more than half).⁸ Comprising ligaments, tendons, and muscle are varying concentrations of collagen, proteoglycans, water, and elastin.^{17–20} These structural components determine each tissue's characteristic viscoelastic be-

Table 1. Subject Demographics (Mean ± SD)*

	Men (n = 12)	Women (n = 12)
Age (y)	22.0 ± 3.3	19.6 ± 1.3
Height (cm)	177.9 ± 6.82	165.5 ± 7.9
Weight (kg)	77.5 ± 16.5	64.8 ± 16.5

*SD indicates standard deviation, and n, number of subjects.

havior during passive lengthening^{15,21} and the elastic and viscous stiffness properties.

A joint stability perspective suggests that increased joint stiffness is a desirable characteristic. Stiffer joints, arising from increased muscle stiffness, are theorized to have a heightened ability to absorb energy contained in destabilizing forces.^{2-4,22,23} Although destabilizing forces may not be countered entirely, many could potentially be lessened in magnitude, thereby reducing the incidence of joint subluxation or dislocation. In contrast, stiffer joints are also theorized to increase the risk for injuries^{5,23} or exacerbate the signs and symptoms associated with antagonistic muscle syndromes.^{24,25} For example, increased extensor muscular stiffness requires higher contractile forces to be developed by the flexor muscles for a given movement. Secondary to the requirement for higher force production are increased stresses to the bone-tendon interfaces and abnormal muscle hypertrophy. Over time, these alterations may potentially increase the predisposition for development of insertional tendinitis and compartment syndromes, respectively.

Quantifying the influence of the gastrocnemius on passive ankle joint complex stiffness (JCS) would provide practitioners with a scientific rationale for selecting clinically advocated rehabilitation and intervention strategies. Such strategies include common clinical techniques such as stretching and strengthening the lower leg musculature. In the one study conducted on the influence of the gastrocnemius muscle on passive ankle JCS,²⁵ researchers focused solely on sedentary women older than 21 years. Additionally, while sex differences in stiffness have been shown at the knee and elbow joints,^{7,26-28} no investigators have considered the existence of sex differences relative to passive ankle JCS. Therefore, our purpose was to determine the effects of sex, ankle and knee joint angles, and the gastrocnemius muscle on passive ankle JCS.

METHODS

Subjects

Twenty-four physically active subjects participated in this investigation (Table 1). The dominant leg, defined as the preferred leg for kicking a ball, was used for all data collection. Physically active was operationally defined as participation in physical activity for a minimum of 20 minutes, 3 times per week. None of the subjects had sustained a lower extremity musculoskeletal injury in their dominant leg or had a history of circulatory or neurologic disorders that could have potentially affected their JCS. Informed consent was obtained from all subjects in accordance with the university's institutional review board, which also approved the study.

Procedures

We determined passive ankle JCS by measuring the resistance to passive movement^{7,8,21,24,25,28-30} from 23° plantar

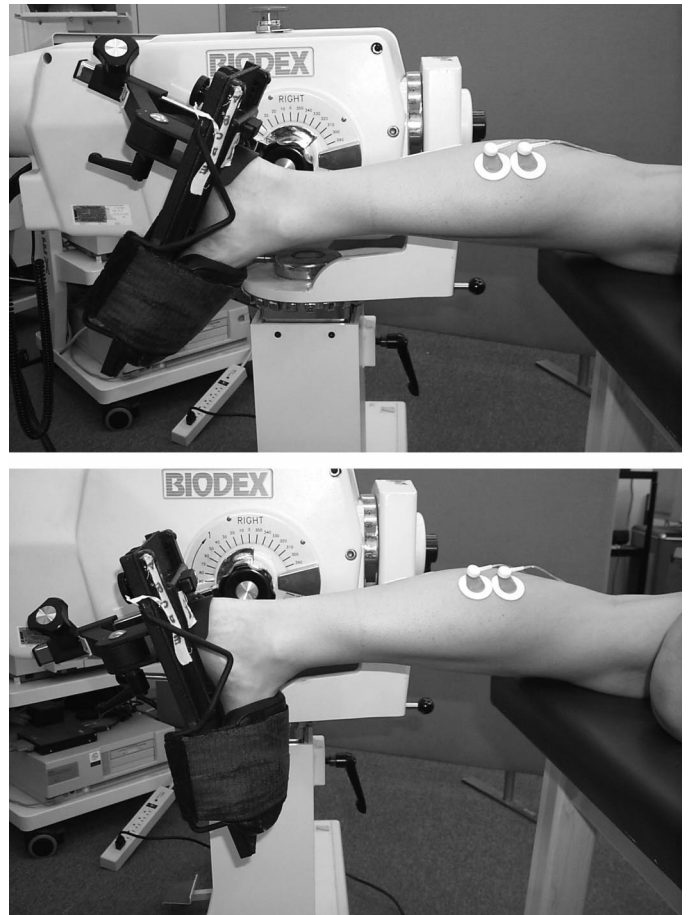


Figure 1. During each trial, the ankle was moved from a starting position of 23° plantar flexion (top) to an ending position of 13° dorsiflexion (bottom).

flexion (PF) into 13° dorsiflexion (DF) (Figure 1). Passive movements at an angular velocity of 5°·s⁻¹ were delivered^{14,21} using the Biodex System 2 Isokinetic Dynamometer (Biodex Inc, Shirley, NY) in a passive mode. The extra 3° ensured that constant velocity was achieved and maintained throughout the target range of 20° PF to 10° DF, thereby eliminating confounding changes in inertia. We used 2 straps (forefoot and midfoot) to fix the foot to the footplate once the axis of rotation was aligned with the lateral malleolus. During all testing, the ambient air temperature was maintained between 20.6°C and 21.7°C.

Trials were completed under 2 conditions designed to include or reduce the contribution of the gastrocnemius muscle. The first condition involved the subject in a prone position with the knee at 0° flexion, while the second condition involved the subject's maintaining a kneeling position with the knee held at 90° flexion. The order of the conditions was counterbalanced among subjects. During testing, we instructed subjects to relax all muscles in the lower leg and to not interfere with the passive movements. Before data collection, we gave each subject several familiarization trials under each condition. In addition to allowing the subjects to become familiar with the testing procedures, the familiarization trials decreased thixotropy³¹ and the stress relaxation phenomena described by Taylor et al.³²

Analogue data concerning angular position and torque from the potentiometer and load cell contained within the dyna-

Table 2. Reliability Analyses*

Condition	ICC	ICC 95% Confidence Intervals		SEM
		Lower Limit	Upper Limit	
Knee straight				
10° PF	.44	.07	.77	.03
NE	.75	.47	.92	.05
10° DF	.88	.71	.96	.08
Knee bent				
10° PF	.54	.18	.83	.02
NE	.87	.69	.96	.02
10° DF	.94	.83	.98	.05

*ICC indicates intraclass correlation coefficients; SEM, standard error of measurement; PF, plantar-flexed position; NE, ankle-neutral position; and DF, dorsiflexed position.

Table 3. Stiffness Values (degrees) by Sex (Mean ± SD)*

Condition	Men	Women
Knee straight		
10° PF	.0028 ± .058	-.0041 ± .034
NE	.4115 ± .127	.3162 ± .081
10° DF	1.3162 ± .305	.9724 ± .217
Knee bent		
10° PF	-.0691 ± .050	-.0670 ± .035
NE	.2833 ± .096	.1911 ± .064
10° DF	1.1176 ± .470	.7726 ± .256

*SD indicates standard deviation; PF, plantar-flexed position; NE, ankle-neutral position; and DF, dorsiflexed position.

momenter head were collected at 100 Hz via an analogue-to-digital converter (Keithley Metrabyte DAS1402, Keithley Instruments Inc, Tauton, MA) and stored on a personal computer for later analysis. Additionally, to ensure that reflexive or voluntary muscle activity was not being elicited during the passive movements, we monitored the activity of the soleus, medial and lateral heads of the gastrocnemius, and medial and lateral hamstring muscles using the Noraxon Telemyo Electromyography System (Noraxon USA Inc, Scottsdale, AZ). Signals from the muscles were collected using self-adhesive silver-silver chloride bipolar surface electrodes (Multi Bio Sensors Inc, El Paso, TX) and passed through a single-ended amplifier (gain 500) to an 8-channel FM transmitter worn by the subject. A receiver then filtered and further amplified the signals (gain 500, Butterworth 15-Hz low-pass and 500-Hz high-pass filters, common mode-rejection ratio of 130 db) before conveying the data to the analogue-to-digital card.

Data Reduction

Although 6 trials of data were recorded under each experimental condition, only the first 3 trials that matched the selection criteria were used in the subsequent analyses. The selection criteria included no muscle activity or alterations in the torque curves through visual inspection of the raw data. We developed customized software to complete all data reduction procedures. First, torque and angular position data were smoothed using a median 5 filter. Gravity corrections for each torque data point were then completed using the corresponding angular position. Factored into the gravity corrections were the mass of the footplate, mass of the foot,³³ and lever arm length.

Separate fourth-order polynomial equations relating angular position and torque were then constructed for each trial ($Y = ax^4 + bx^3 + cx^2 + dx + e$, where Y is the gravity corrected torque, x is the angular position, and a through e are constants). Stiffness values were calculated at 10° PF, neutral (NE), and 10° DF by using the first derivative (slope) of the equation ($dy/dx = 4ax^3 + 3bx^2 + 2cx + d$, where dy/dx is the stiffness) at each of the respective positions. This method of stiffness measurement, using the slope of the line relating torque and angular position during passive movement, has been previously described³⁰ and used.^{24,25,29} The stiffness represented by taking the slope of the line has been attributed to elastic stiffness at that point in the range of motion.^{25,34}

Reliability

We conducted a pilot study in conjunction with the current study to establish the reliability for our exact methods. Twelve subjects (8 men, 4 women, mean height = 176.6 cm, mean weight = 80.2 kg) participated in 3 repeated testing sessions. Twelve subjects were chosen based on an a priori power analysis.³⁵ All subjects conformed to the previously discussed inclusion and exclusion criteria. Intraclass correlation coefficients³⁶ (ICC) (2, K) ranged from .44 to .93 (Table 2). In addition, we established the confidence intervals and standard error of measurement associated with each ICC. Standard error of measurement ranged from .02 to .08, supporting the absolute reliability of the methods. These results are comparable with those previously reported using similar methods at the ankle.³⁷

Data Analysis

The stiffness values across the 3 trials under each condition were averaged and analyzed using a 3-factor, repeated-measures analysis of variance with sex as a between-subjects factor and condition and position as within-subjects factors. Statistical significance of $P < .05$ was set a priori for all analyses.

In an attempt to probe the cause of the significant sex effects, we conducted several post facto analyses. First, we performed Pearson bivariate correlational analyses between the stiffness values (each position and condition) and subject height, weight, and ponderal index (calculated by dividing height by the cube root of weight³⁸) across all subjects. Additionally, independent t tests were conducted on the height, weight, and ponderal index variables between the sexes. To reduce the Type I error rate, statistical significance for the t tests was adjusted to $\alpha < .01$. Last, we repeated the 3-factor analysis of variance using any of the demographic variables (height, weight, ponderal index) determined to be significantly related to stiffness and significantly different between the sexes as a covariate.

RESULTS

The data of 1 subject had to be disregarded for technical reasons. An example of the torque versus position data from 1 acceptable trial, as well as the constructed equation, is presented in Figure 2. Means and standard deviations are provided in Table 3. The results of the statistical analysis on the remaining subjects revealed significant condition-by-position ($F_{2,42} = 6.39, P = .004$) and sex-by-position ($F_{2,42} = 7.40, P = .002$) interactions, as well as significant main effects for

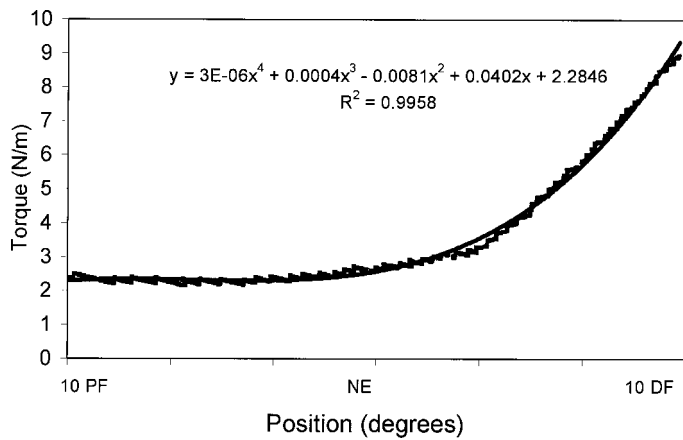


Figure 2. Separate fourth-order polynomial equations were constructed relating the angular position and torque data for each trial. Pictured are the original data (rough line) and line of the equation (solid).

sex ($F_{1,21} = 6.22, P = .021$), position ($F_{2,42} = 288.00, P < .000$), and condition ($F_{1,21} = 73.90, P < .000$). Tukey post hoc analyses of the condition-by-position interaction revealed significantly higher stiffness values under the knee-straight condition compared with the knee-bent condition at both ankle NE and 10° DF (Figure 3). Within each condition, stiffness values across positions were significantly higher as the ankle moved into DF (10° PF $<$ NE $<$ 10° DF). Tukey post hoc analysis of the sex-by-position interaction revealed significantly higher stiffness values at 10° DF in men (Figure 4). Lastly, Tukey post hoc analysis of the position main effect revealed that, as the ankle moved into dorsiflexion, the stiffness at each position became significantly higher than at the previous position (10° PF $<$ NE $<$ 10° DF).

Results of the post facto correlational analyses revealed significant relationships ($P < .05$) between height and the stiffness values at the NE and DF positions under both conditions (Table 4). The independent t tests revealed only height to be statistically different between the sexes ($t_{21} = 4.01, P = .00$). The results of the analysis of covariance using height as the covariate were identical to the analysis of variance except that the main effect for sex was nonsignificant.

DISCUSSION

The purpose of our investigation was to determine the effects of sex, joint angle, and the gastrocnemius muscle on passive ankle JCS. The most significant aspect of our study was the quantification of the gastrocnemius' influence on pas-

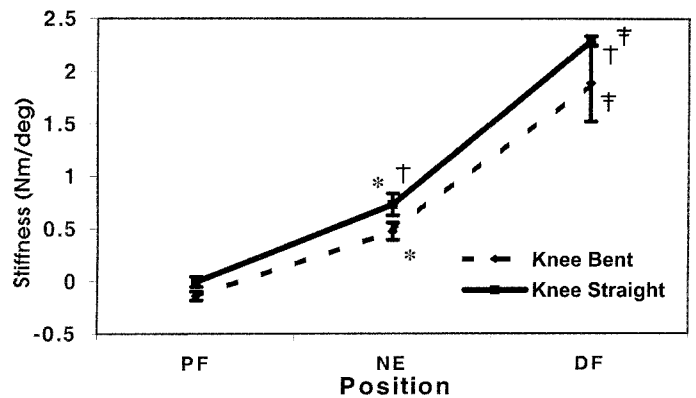


Figure 3. Position stiffness mean (\pm SD) for each condition illustrating the condition-by-position interaction.

sive ankle JCS. Collapsed across sex, our results suggest that the gastrocnemius significantly increased passive ankle JCS moving into DF as early as ankle NE. Clinically, this implies that the contribution of the gastrocnemius represents an important consideration during rehabilitation programs involving the lower extremity. Additionally, collapsed across condition, our results demonstrate significant sex differences in stiffness at 10° DF. The clinical significance and cause of this result warrant further investigation.

The speed with which we induced the passive displacements, $5^\circ \cdot s^{-1}$, was chosen to avoid eliciting stretch reflexes.¹⁴ Further, resistance to passive ankle displacement at this speed has been demonstrated as unchanged under ischemic conditions that block the Ia afferent fibers from the muscle spindles.¹⁴ By asking subjects to not intervene with the passive movements,^{5, 6, 10, 21, 23, 27, 29} we took advantage of the ability to abolish muscle activity through conscious relaxation,³⁹ thereby eliminating conscious muscle activation as a confounding factor. The relatively few trials in which increased electric activity occurred in our study, coupled with the ability to easily identify and eliminate these instances, support this presumption. Thus, it is reasonable to attribute the resistance measured in response to passive ankle joint displacement into DF to the intrinsic mechanical properties of the joint complex. The potential contributory sources, in addition to the gastrocnemius, include any of the structures spanning the joint: skin, ligaments, joint capsule, and anterior and posterior surrounding compartment muscles. After sequential resections of the tissues crossing the wrist joint, Johns and Wright⁴⁰ reported that resistance to passive movement arose primarily from the joint capsule (47%) and the muscles (41%). The remainder of the resistance was provided by tendons (10%) and skin (2%). With

Table 4. Correlational Analyses Among Demographic Variables and Stiffness Values (n = 23)*

	Knee Bent			Knee Straight		
	10° PF	Condition NE	10° DF	10° PF	Condition NE	10° DF
Height	$r = -.219$ $P = .316$	$r = .429$ $P = .041$	$r = .501$ $P = .015$	$r = .086$ $P = .696$	$r = .494$ $P = .017$	$r = .648$ $P = .001$
Weight	$r = -.394$ $P = .063$	$r = .289$ $P = .181$	$r = .373$ $P = .080$	$r = -.108$ $P = .015$	$r = .408$ $P = .054$	$r = .569$ $P = .005$
Ponderal index	$r = .287$ $P = .184$	$r = .044$ $P = .841$	$r = -.018$ $P = .937$	$r = .250$ $P = .251$	$r = -.076$ $P = .731$	$r = -.143$ $P = .516$

*n indicates number of subjects; PF, plantar-flexed position; NE, ankle-neutral position; DF, dorsiflexed position; r , bivariate correlation coefficient; and P , P value.

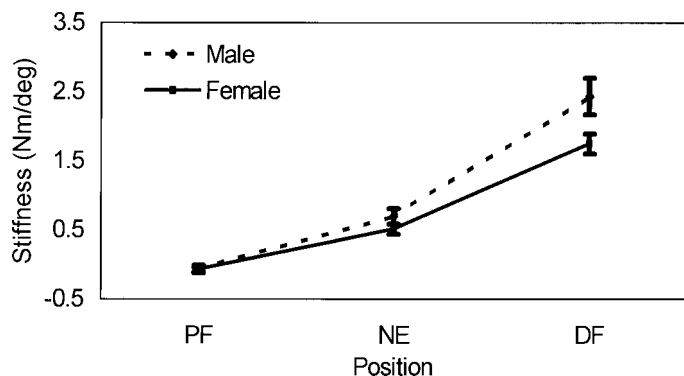


Table 5. Gastrocnemius Muscle Length Relative to Reference Position Length* Associated with the Knee And Ankle Angular Positions†‡

Condition	10° PF	NE	10° DF
Knee bent	-1.81%	0%	3.76%
Knee straight	4.48%	6.46%	10.05%

*The length of the gastrocnemius with 90° ankle and 90° knee flexion.
 †Derived from equations established by Grieve et al.⁴⁴
 ‡PF indicates plantar-flexed position; NE, ankle-neutral position; and DF, dorsiflexed position.

Figure 4. Position stiffness mean (\pm SD) for each sex illustrating the sex-by-position interaction.

respect to the elbow joint, Chleboun et al²⁸ noted that muscle volume accounted for 84% of the variance in elbow stiffness. Considered collectively, these studies suggest that the degree of contribution from each articular structure may be unique to a particular joint.⁷ We were unable to find similar studies comparing contributions of the various articular structures at the ankle joint, leaving some of the etiologic interpretation of our results to limited degrees of speculation.

Our experimental design took advantage of the biarticular span of the gastrocnemius muscle. Because the proximal attachment of the gastrocnemius resides above the posterior femoral condyles, flexing the knee to 90° shortens the distance between the distal and proximal attachment sites, thereby decreasing the potential passive resistance. A similar method has been used to determine the contribution of the gastrocnemius to maximal voluntary ankle torque production.⁴¹ Thus, comparing the stiffness values attained during the knee-bent condition with those attained during the knee-straight condition provided a means by which we could determine the relative influence of the gastrocnemius. Although changes in the tension offered by skin and associated connective tissues could have accompanied the change in knee position, we feel that it is reasonable to disregard such effects as minimal in light of the paper by Johns and Wright.⁴⁰

Both testing conditions included the resistance offered by the uniarticular muscles crossing the axis of the ankle joint, posterior ankle joint capsule, and ligaments. Measurement of ankle ligament force values in various degrees of DF and PF using isolated cadaveric specimens has been conducted.⁴² The anterior talofibular ligament comes under tension as the ankle moves into PF, while the calcaneofibular ligament comes under maximal tension in 15° of PF and DF. The deltoid ligament displays a pattern similar to that of the anterior talofibular ligament, coming under increased tension as the ankle moves into plantar flexion. The relevance of these results in resisting passive motion in vivo, with other articular structures intact, remains unknown. With respect to individual uniarticular muscle contributions, Gareis et al,⁴³ in establishing the active length-tension curves and passive force characteristics of 9 lower extremity muscles, demonstrated that the passive tension provided throughout the full range of elongation by the soleus muscle was more pronounced than that of the peroneus longus, tibialis posterior, and flexor digitorum longus muscles.

Our results of a significant condition-by-position interaction and a significant main effect for condition demonstrate that the gastrocnemius has substantial influence on passive ankle JCS. These results are in contrast with those reported by Chesworth

and Vandervoort.^{25,37} We find this discrepancy quite surprising and difficult to explain considering the identical methods in the 2 studies. The largest difference between the 2 studies involved the subjects. While we studied physically active men and women between 18 and 30 years old, Chesworth and Vandervoort^{25,37} studied only women between 21 and 80 years old.

Several studies considering the effects of ankle and knee joint angle on the gastrocnemius support our significant condition results. Grieve et al⁴⁴ established a technique to estimate the length of the gastrocnemius muscle from knee and ankle angular measurements. Using the equations provided by their report, the knee-straight condition would have increased the length of the gastrocnemius in our subjects by approximately 6.5% in comparison with the reference length (the length of the gastrocnemius with 90° ankle and 90° knee flexion) at each of the respective ankle angular positions (Table 5). Sale et al⁴¹ confirmed through a radiologic series that femoral condyle rotation accompanying knee extension causes considerable lengthening of the gastrocnemius independent of foot position. Although length and stiffness are not synonymous terms, temporarily increasing the length of the muscle-tendon unit with changes in joint position shifts the passive resistance-position curve to the left, resulting in increased stiffness at each ankle angle.

The post hoc analysis of the condition-by-position interaction revealed significant differences between conditions, beginning at the ankle neutral position. Further research should focus on further isolating the angular position where the significant difference begins. Given the differences in the passive length-force curves between the lateral and medial gastrocnemius⁴³ and differences in recruitment patterns,¹⁵ we recommend further research to consider the influence of each head independently.

In light of several other studies reporting sex differences in stiffness,^{7,26-28} our results of a significant sex-by-position interaction and significant main effect for sex are not surprising. The lack of significant sex-by-condition and sex-by-condition-by-position interactions, however, suggests that there was no difference in the stiffness of the gastrocnemius muscle between the sexes. The origin of the significant differences revealed could be related to dissimilarities in structural or physical characteristics. Potential structural characteristics include such factors as tissue elastic and collagen content variations. Komi and Karlsson^{45,46} suggested the lower rates of force development and elastic energy storage exhibited by women could be related to differences in elastic tissue content within the muscle. Resistance to passive motion in the absence of muscle activation has been attributed to the parallel elastic components of muscle.^{45,47,48} Different concentrations of elastic tissue between the sexes, as Komi and Karlsson^{45,46} sug-

gested, could therefore potentially explain the revealed sex differences in stiffness.

Pertinent physical characteristics include such factors as tissue cross-sectional areas, flexibility, and mechanical advantage differences. Several studies have demonstrated differences in muscle mass between the sexes.^{28,49,50} Additionally, relationships have been shown to exist between cross-sectional areas and stiffness.^{26,28,29} Thus, it could be that differences in muscle mass existed within our subjects between sexes, thereby accounting for the significant sex differences. However, within the limits of the relationships existing among muscle mass, body weight and ponderal index, this does not appear to be a factor in our study. The number of nonsignificant relationships revealed among stiffness, body weight, and ponderal index provides support for this statement. Interestingly, Chleboun et al²⁸ failed to reveal a significant relationship between muscle size and elbow stiffness at the end-range position, the location in the range where we revealed significant sex differences.

Much to our surprise were the significant, moderate-magnitude relationships revealed between the stiffness values and height. The results of the analysis of covariance, using height as the covariant, demonstrated that height could account for some of the previously revealed sex differences. This provides support for the ideas of structural or mechanical advantage (or both) differences existing between the sexes. Further research is needed to quantify the mechanism by which height influences stiffness.

We are not unique in finding significant position differences in stiffness at the ankle.^{11,13,14,16,25,34} Toft et al³⁴ suggested that as the joint reaches end range, more parallel tissue elements become loaded, giving rise to the exponential increases in resistive forces required to passively move the ankle into further DF. The curves presented by Gareis et al⁴³ illustrate the exponential increases in passive tension resulting from muscle lengthening. The technique provided by Grieve et al⁴⁴ provides us with a method of quantifying the approximate length changes the gastrocnemius undergoes as the ankle moves into DF under each condition (Table 4).

It is also interesting to note the negative mean stiffness values at the 10° PF position under the knee-bent condition (both sexes) and under the knee-straight condition (women). We are not the first authors to report negative stiffness values.⁵¹⁻⁵³ Such expression, however, does not comply with the traditional concept of the physical characteristic stiffness. As Latash and Zatsiorsky⁵¹ described, stiffness assessments reflect both features of the system and the method of testing. Because negative stiffness of biological tissues and structures is a clear impossibility, the negative stiffness values can be attributed to our particular assessment approach. Our method involved measuring the stiffness of the entire ankle joint complex under a passive state with respect to movement into dorsiflexion. During our measurements, throughout the entire range of motion, all components acting on the ankle joint, both anterior and posterior to the ankle joint axis, imposed their respective influences on our measurements. The negative stiffness value indicates that the net resistance measured while moving into DF from the 10° PF position was acting in the opposite direction (pulling the ankle into DF). In other words, in the plantar-flexed position, the posterior musculoskeletal structures did not impose sufficient passive torque to overcome the passive torque being imposed by the anterior musculoskeletal structures, so negative values resulted. These negative values indicate the opposite of stiffness, or compliance, in the direction

associated with the measurement (dorsiflexion). The anterior muscles (tibialis anterior, extensor digitorum longus, extensor hallucis longus) were most likely the major contributors to this phenomenon, with contributions also arising from the anterior joint capsule and anterior talofibular and deltoid ligaments.⁴² Thus, the negative stiffness is simply a result of conducting the passive torque measurements on both sides of the equilibrium point of the ankle joint.⁷ A similar phenomenon can be observed at the knee joint, as presented by Allison et al.²⁴

Clinically, our results suggest that the stiffness of the gastrocnemius represents an important consideration in the management of lower leg conditions. Examples include antagonistic muscle syndromes such as anterior compartment syndrome and insertional tendinitis. Activities of daily living, including normal gait,^{54,55} involve the ankle's moving repetitively into flexion positions greater than neutral by action of the anterior ankle muscles. If either of the previously mentioned antagonistic muscle syndromes is present, our results suggest that the work performed by the anterior muscles could be lessened through a reduction in the stiffness of the gastrocnemius. In other words, this investigation provides scientific rationale for addressing the gastrocnemius during clinical management strategies involving the lower limb. Further research should address the acute and chronic effectiveness of various intervention strategies on altering the passive stiffness of the ankle.

Most often, only flexibility (length) is taken into account during clinical assessments. Although stiffness and flexibility are interrelated,⁶ they are largely separate physical characteristics. Flexibility is best defined as the angle beyond which no further displacement is possible,³⁴ providing limited information regarding the behavior of muscle-tendon units in response to stretch,^{21,56} especially at muscle lengths used during daily activities.³⁴ In contrast, stiffness represents the amount of deformation proportional to the load applied.²¹ Thus, from a clinical perspective, measuring passive stiffness to imposed dorsiflexion may provide additional insight into the cause of such pathologic conditions.

CONCLUSIONS

The results of this investigation are applicable to all clinicians treating lower extremity conditions and provide a scientific basis for clinicians incorporating gastrocnemius-stretching regimens into rehabilitation programs. The cause and application of the significant sex-by-position interaction requires further study. Further research is also warranted regarding the influence of the gastrocnemius during various levels of muscle activation and knee positions. The effectiveness of short-term and long-term intervention strategies in altering passive ankle JCS remains largely unknown.

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COMMENTARY

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Musculoskeletal stiffness of the ankle joint is recognized as a significant factor contributing to postural stability, locomotion, and ankle joint stability. The clinical community is increasingly focusing on musculoskeletal joint stability for the prevention and treatment of injury. Although stiffness is not the lone determinant of stability, it is one of the major contributing factors. Hence, the authors provide much needed and timely data toward the understanding of ankle joint stiffness and stability. The purpose of this study was to identify the influences of ankle angle, knee angle (ie, gastrocnemius length), and sex on ankle joint complex stiffness (JCS). The results demonstrate a significant effect of all 3 factors on JCS.

Measurement and interpretation of musculoskeletal stiffness is a nontrivial effort. The authors provide a great service in pointing out that flexibility and stiffness are separate and distinct biomechanical concepts, each contributing in a unique manner to functional performance, injury, and pathology. An elegant method of computing stiffness was performed by computing the first spatial derivative of a best-fit analytic function to the measured force-by-angle data recorded during passive ankle dorsiflexion. Although the accuracy of the curve-fit was not reported, example data in Figure 2 suggest excellent performance, indicating reliability when computing the slope, or JCS. However, the results and discussion illustrate the need for improved characterization of stiffness. Stiffness is not merely the ratio of joint moment divided by joint angle. It is defined as the partial derivative of force by length, or equivalently, rotational stiffness is the partial derivative of moment by angle. The authors employed the ordinary derivative to compute stiffness, thereby making the assumption that angle is the sole factor contributing to joint moment. This ignores changes in muscle activation and gravitational, inertial, and viscous contributions to joint moment among other factors. Focusing specifically on passive behavior controlled muscle activation. Examining the behavior during slow, isokinetic movement reduced inertial and viscous components. Gravitational confounding remained in the data and was noted by the authors (eg, negative stiffness values were explained as a gravitational artifact in combination with anterior muscle behavior). It is agreed that these effects caused this negative relationship between moment and angle, so it should not be confused with stiffness. Instead, it is simply the passive moment-by-angle relationship influenced by many factors. In fact, in this paradigm, negative stiffness would represent a source of free energy generation, thereby violating multiple laws of physics. Anatomic and physiologic stiffness elements that are

part of the musculoskeletal system remain an excellent means for energy storage in the propagation of locomotion,^{1,2} but the stiffness elements cannot generate energy. Hence, the values of negative stiffness reported in the results illustrate a source of error and the need for rigorous definitions and characterization of biomechanical stiffness. In an otherwise well-controlled effort, the results demonstrate potential pitfalls when interpreting stiffness data and warrant caution when interpreting the results.

The study identifies dorsiflexion angle, gastrocnemius length, and sex as factors that influence ankle JCS. Results from this study support previously published data illustrating that passive ankle JCS increases as the joint is moved toward the end range of motion.³ Joint complex stiffness during the knee-flexed posture was reduced compared with the knee-straight condition, presumably identifying the contribution of the gastrocnemius stiffness characteristics. Results indicating sex differences in JCS are particularly interesting when considered from the clinical perspective of sex biases in musculoskeletal injury rates. Analyses suggest sex differences in JCS at the end range of motion may have been related to population differences in standing height and weight. These results are logical from a biomechanical perspective, as stiffness is related to material geometry, analogous to anthropometric trends in muscle cross-sectional area and length associated with height and weight.

The results describe passive characteristics of the ankle joint in unloaded conditions. One wonders to what extent these passive components contribute to functional performance or injury risk. During walking, a healthy adult readily exceeds 150 Nm of external dorsiflexion moment about the ankle.⁴ The joint moment measured in this study attributed merely 9 Nm of external dorsiflexion moment to the JCS. Moreover, the data in the current study were measured with the ankle joint unloaded (ie, the subject was either kneeling or lying prone). Recent evidence⁵ indicates that passive musculoskeletal stiffness may be influenced by the compressive load achieved in standing postures. Total JCS is also influenced by the contribution of active muscle stiffness⁶ generated during weight-bearing activity. Passive, unloaded stiffness data are necessary to understand ankle joint performance and injury prevention but require further research efforts to fully characterize ankle stability.

In general, this study presents some very insightful information when considered from a clinical perspective. I encourage future researchers, based upon these efforts, to improve the characterization of musculoskeletal stiffness and to advance the paradigm to permit insight into stiffness behavior during functionally loaded conditions, including passive and active muscle contributions to joint stability.

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AUTHORS' RESPONSE

We thank Dr Granata for his commentary concerning our manuscript and appreciate the opportunity to respond. Dr Granata provides some excellent insights and raises an important question regarding the measurement of musculoskeletal stiffness: How should musculoskeletal stiffness be assessed? In light of the recent increased interest in the potential role that stiffness may play in functional joint stability, it is important that researchers work under universally accepted operational definitions and methods to promote synthesis of information across investigations. To date, however, there has been little consistency throughout the literature in either the definitions or methods of stiffness used. Thus, we completely agree with Dr Granata's recommendation for future research to improve our understanding of musculoskeletal stiffness and its importance in joint stability.

However, in contrast to regarding stiffness from a joint stability perspective, our current research was aimed at beginning to evaluate stiffness as a potential contributor to pathologic musculoskeletal conditions. Specifically, our purpose was simply to provide athletic trainers with a scientific rationale regarding several intervention strategies involving the gastrocnemius. To this end, we chose a passive measurement approach, and to remain consistent with several previously published investigations (references 7,8,21,24,25,28–30), we opted for the approach used. As Dr Granata indicates, stiffness is the partial derivative of the force-length relationship (or moment-angle relationship in angular systems), and therefore, all other factors (ie, muscle activation, gravitational, inertial, viscosity, thixotropy, etc) confounding this relationship must be either controlled or reduced. Our methods involved a slow, isokinetic displacement in the absence of muscle activation, thereby controlling the issues associated with inertia, viscosity, and muscle activity. The confounding effects of gravity were reduced during data reduction by correcting for the mass of the footplate and foot and lever arm length, while thixotropy and stress relaxation effects were reduced through use of familiarization trials.

The apprehension that Dr Granata expresses concerning the negative stiffness values revealed in the 10° plantar-flexed position is valid. His interpretation of this phenomenon appears to be that residual confounding stiffness influences resulted from limitations in the measurement technique selected. The point concerning negative stiffness representing free energy generation is well taken; however, we propose that energy was not generated but rather released from storage as a result of forcing the ankle into a 20° plantar-flexed position. In this position, the anterior ankle musculature was stretched, and, as the ankle displaced into dorsiflexion, the anterior ankle musculature strain energy was recovered. Accompanying the changing ankle position would be alterations in the moment arms of the anterior ankle musculature, thus potentially explaining the negative stiffness values. Regardless, we agree with Dr Granata that caution must be exercised in interpreting stiffness data.

While we agree that our measurement method was not perfect and may account for the above discrepancies, we maintain our position that it was appropriate to answer our specific clinical research question: What is the role of the gastrocnemius in passive ankle joint complex stiffness? It is important to recognize that all structures about the ankle joint influenced our measurements, and thus, the final resistance measured by the load cell was the net torque acting at the ankle joint to the imposed dorsiflexion displacement. The fact that the negative stiffness values were revealed only under the knee-bent position for the men further supports our conclusion that the gastrocnemius potentially influences passive ankle joint complex stiffness. In other words, under reduced gastrocnemius influence, the ankle joint demonstrated compliance to the imposed dorsiflexion displacement in the 10° plantar-flexed position.

We want to emphasize that our results are valid only under specific circumstances: an unloaded joint without muscle activity. As Dr Granata indicates, these results cannot be directly applied to many acute injury mechanisms that involve the joint's being loaded with varying degrees of muscle activation. The results are applicable, however, to specific circumstances within activities of daily living, such as the anterior musculature working to dorsiflex the ankle during the swing phase of gait.

Again, we thank Dr Granata for his excellent commentary. We also thank the anonymous reviewers and editorial staff for their assistance in helping to improve the quality of this manuscript.