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## QUANTIFYING INTRINSIC ANKLE STIFFNESS IN QUIET STANDING: A SYSTEMATIC REVIEW

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### Abstract:

**Introduction:** Ankle stiffness is a factor that contributes to maintaining balance and counteracts the forces of gravity as the body sways. Stability while standing quietly depends on its value. Intrinsic ankle stiffness refers to (i) the passive resistance provided by the structural components of the ankle joint itself, such as the ligaments and joint capsules, (ii) a property of the joint that remains constant regardless of the external forces applied, (iii) a characteristic that affects the stability and control of the ankle joint. The aim is to present compilation of the results published in the researchers' work makes it possible to analyze changes in the value of ankle stiffness for perturbations of different amplitude.

**Material and Methods:** The review is based on articles identified through searches of Pubmed, Web of Science, and Scopus. There were no restrictions on the publication date. The Boolean search strategy was used.

**Results:** Ankle stiffness changes with increasing sway amplitude. Its values are greater for perturbations of smaller amplitudes and, therefore, cannot be represented by a single value. Passive stiffness reduces sway and allows time for the active system to respond. The interaction of these two mechanisms ensures the stability of an upright posture.

**Conclusions:** Ankle stiffness is a parameter that can be applied in clinical practice. The exact determination of the range of stiffness values is a useful tool to define a motor organization/reorganization.

## 1. Introduction

Estimating ankle stiffness during quiet standing is a key issue for understanding the basic mechanisms of motor control. Stabilization of the upright body posture is an example of the many complex tasks aimed at maintaining balance [1]. These tasks are found in everyday and professional life, as well as in more demanding sports situations characterized by repulsive forces that tend to tilt the body away from its intended position. The asymptotic stability of this position would be achieved if the destabilizing moment, which is usually proportional to the displacement, was compensated by a stronger recovery moment generated by the intrinsic stiffness of the muscles and other load-bearing tissues [1]. In the quiet standing position, the body has a slight tendency to lean forward, and instability is mainly caused by gravity [2]. Previous research has shown that people can have very different intrinsic ankle stiffness [1,3-4].

The rate of increase of the rollover moment (i.e. rollover moment per unit of angle factor) determines the critical level of stiffness [1]. Stability is maintained without control if stiffness exceeds a critical level. Below this level, insufficient stiffness requires active stabilization.

Two systems generate ankle torque. The first passive mechanism uses viscoelastic forces from stretching the ankle's muscles, tendons, and ligaments to work instantly as an ankle intrinsic stiffness [3]. The second is nervous system-controlled muscle activity modulation. In a standing position, anticipatory mechanisms of the nervous system reduce the delay in the response to unexpected perturbations. "Separating the reflexive and intrinsic contributions to the overall ankle stiffness during standing might give insight into neuromuscular disorders and help in clinical balance control assessment" [4]. The intrinsic ankle stiffness cannot stabilize the body in a standing position, according to research. However, it introduces a passive, immediate response to fall risk, allowing neuronal interventions when balance is lost [5]. Acting as springs, the Achilles tendon and triceps surae cause ankle stiffness [3].

Quiet standing reduces Achilles tendon and triceps surae stretching and ankle joint torque. A long, prone tendon and almost stationary nearby muscles with short fibers determine ankle stiffness [5]. The maximum ankle stiffness is determined by the tendon, which is 15 times weaker than the muscles in quiet standing [6]. In high-torque conditions like walking, jumping, and running, muscle stiffness only matters for ankle stiffness if tendon stiffness increases. Active twitch or passive resistance, stretching and changing the recent history of muscle activity (e.g., leaving the muscle motionless for 10-15 s), can cause thixotropy and stiffen muscles [7].

The Achilles tendon connects the triceps surae to the heel bone and can be extended by an active contraction or passive ankle dorsal flexion. Both types of tendon elongation lead to its stiffening. Torque increases ankle stiffness in standing. Tilting forward or dorsiflexing the foot increases ankle torque [3].

Understanding the foot and ankle joint's mechanisms requires understanding the intrinsic ankle stiffness. The present review's measurements can estimate intrinsic ankle stiffness during standing posture stabilization. This review covers the latest methods for measuring intrinsic ankle stiffness in the anterior-posterior plane.

Literature extensively discusses intrinsic ankle stiffness. There are many ways to estimate this parameter and interpret its mechanism [1,3-4,8,10] Although a unified methodology would be ideal, each author analyzes this phenomenon differently. Ankle stiffness research is evolving due to measurement technology and interdisciplinary approaches. This review presents current research results and phenomena that significantly affect such measurements. This systematic review also investigates how ankle stiffness changes with perturbation amplitude. We aim to find a trend or pattern in ankle stiffness measurements from selected publications to draw general conclusions about ankle stiffness variability under sagittal plane perturbations.

The paper describes measurement methods and ankle stiffness values from tests. This will enable parameter and trend presentation.

## 2. Material and Methods

### 2.1. Search strategy

The current review is based on articles identified through searches of Pubmed, Web of Science, and Scopus. There were no restrictions on the publication date. Based on the Boolean search strategy, database searches were conducted using the following keywords: "foot training" AND "postural control" AND "training" AND "balance" AND "postural sway" AND "functional ankle instability." In the subsequent stage of the search, the terms "ankle stiffness" AND "standing" OR "balance" OR "gait" OR "postural control" OR "sway" were utilized to refine the subject and search terms. The search was restricted to English articles in their entirety in their original language.

### 2.2. Inclusion criteria

Only studies meeting the following criteria were considered for inclusion in the review: 1) included a protocol for measuring ankle stiffness, 2) described the measuring device, 3) described the method for estimating ankle stiffness, and 4) presented the results of ankle stiffness measurements.

Studies that lacked a detailed description of the method, design, and results of measuring ankle stiffness were excluded. The review of the literature was based on inclusion and exclusion criteria.

### 3. Results

#### 3.1. Search results

An electronic search of the previously mentioned online databases identified 4,755 potentially relevant manuscripts. On the basis of the title, 4,435 articles were excluded from the research. Based on the abstracts, 18 articles were retained for full-text analysis. In the next stage of the search, lists of the 12 selected full-text articles were added. A total of 6 studies were eligible for review (see Figure 1).

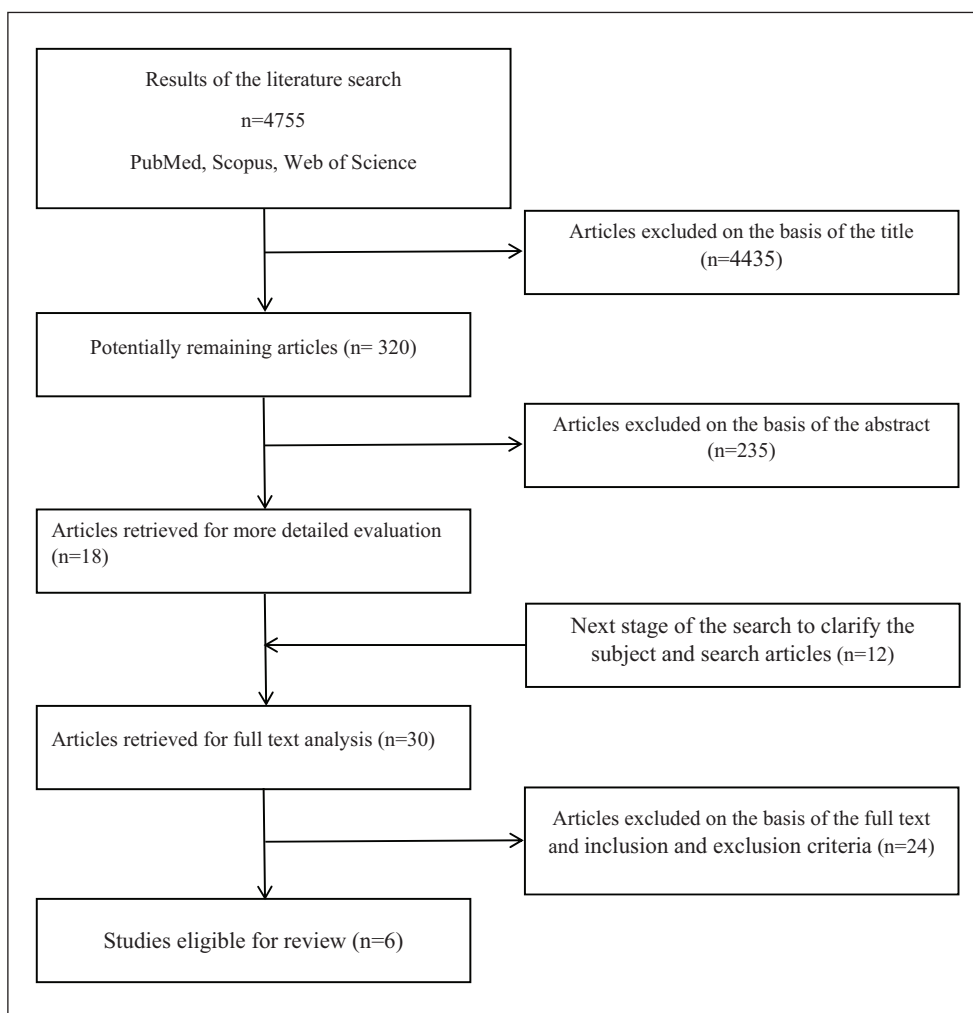


Figure 1. Flowchart

#### 3.2. Measurement results of ankle stiffness

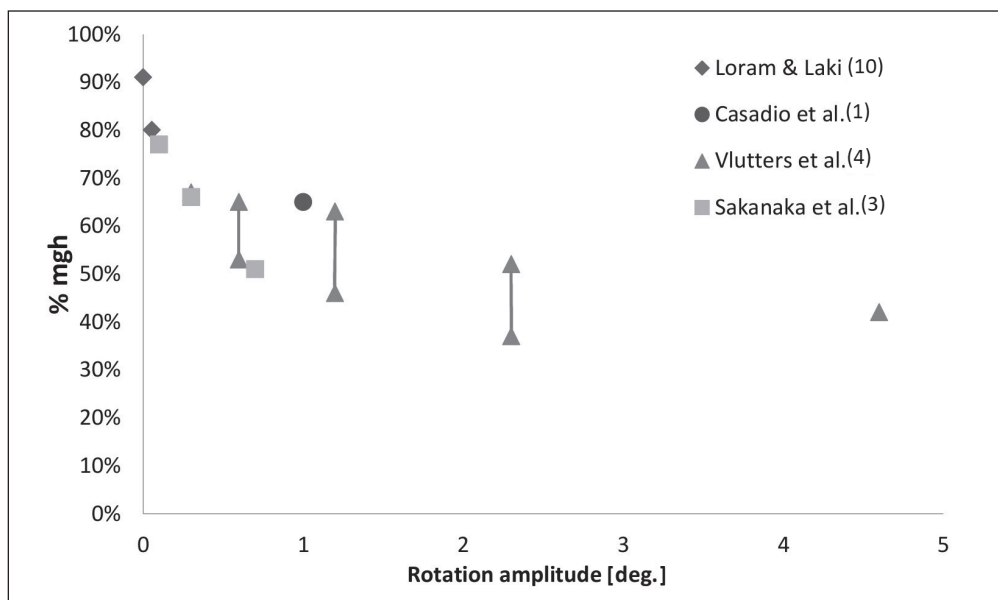
Table 1 shows the results obtained from the normal condition, making a comparison between experiments possible. Lang et al. (2004) presented average values for all conditions in total.

As the disorder amplitude increases, ankle stiffness decreases (Figure 2). Loram and Laki et al. [10] found the highest percentage of mgh for quiet standing and perturbations with the smallest amplitude of 0.055 degrees. This threshold flattens the percentage mgh curve. Loram and Lakie [10] found that smaller and slower ankle movements increase ankle stiffness. Quiet standing's upper limit may be 0.055-amplitude perturbations. Casadio et al. [1] note that Loram and Laki et al. [10] found stiffness close to the critical level and independent of torque due to the measurement method. We can better understand the ankle joint and foot mechanism by studying the foot mechanism during freestanding perturbations. The graph shows point determination mean values, but their spread is much larger. Vertical lines ending in points define the perturbation's percentage mgh range. Vlutters et al. [4] confirmed the graph's

**Table 1.** Intrinsic ankle stiffness during quiet standing

References	N/Sex/Age	Height/Weight	Time window	Platform position	Perturbation amplitude	P/N	Average relative pseudo intrinsic ankle stiffness (%).
Loram & Laki (2002)	15 subjects between 20-68 years	–	140 ms	H	0.055 deg.	30	91% <i>mgh</i> for free standing. 80% <i>mgh</i> for 0.055 deg perturbation. Direction: DF
Casadio et al. (2005)	9 female 9 male, between 21-31 years	1.55 - 1.88 m./ Between 47 kg and 88 kg.	150 ms	H	1 deg.	20	Female: 0.608 <i>mgh</i> . Male: 0.672 <i>mgh</i> For DF: 0.649, PF: 0.63
Lang et al. (2014)	6 male 3 female Between 23-29 years	–	500ms	H	0.03 rad.	2	For one leg: 26-45% <i>mgh</i> For two legs: 52-80% <i>mgh</i> . Positive position pedals indicate DF.
Vlutters et al. (2015)	1 female 7 male 23±1years.	1.85±0.07 m. / 75±8 kg	40 ms	H	0.08, 0.04, 0.02, 0.01 and 0.005 rad.	8	67% <i>mgh</i> for 0.005 rad., 65-53% <i>mgh</i> for 0.01 rad., 63-46% <i>mgh</i> for 0.02 rad., 52- 37% <i>mgh</i> for 0.04 rad., 42% <i>mgh</i> for 0.08 rad. Pooled values for PF and DF.
Amiri et al. (2017)	3 subjects, between 18-40 years.	–	Lasting until 40 milliseconds after the peak velocity, before reflex response.	H	0.02 rad.	–	106 Nm/rad at background torque of -5 and increase to Nm 253 Nm/rad in the background torque of -22 Nm for DF perturbation.
Sakanaka et al. (2018)	6 female 4 male, 28.1±4.4 years (mean±SD)	1.68±0.1m./ 65.9±8.3 kg.	70 ms	H	0.1 deg., 0.3 deg. and 0.7 deg.	30	For normal position: 77% for 0.1 66.3% <i>mgh</i> for 0.3 and 51% <i>mgh</i> for 0.7 deg. Combined values of both directions.

– Lack of information, % *mgh* – percentage of toppling torque per unit angle. DF – dorsiflexion. PF – plantar flexion, H – horizontal. P/N – perturbation number.



**Figure 2.** Ankle stiffness value data for a specific rotation amplitude value

steep descent, but their higher amplitude measurements showed a less steep decline. For the 2.3 degree perturbation, values were 52–38 percent mgh (data from the chart in the paper). The mean value was 42% mgh for a 4.6-degree perturbation. The Casadio et al. [1] study's 1 degree amplitude of perturbation outlier may be due to the measurement method. The intrinsic ankle stiffness was divided by the torque difference by the lower leg-platform angle. This method yields higher ankle stiffness values than the [pseudo] ankle stiffness estimated by dividing the difference in torque by the platform encoder angle. Loram and Lakie [1] and Vlutters et al. [1]. The latter estimated ankle stiffness using the two methods above, proving the measurement difference theorem.

Figure 2 shows the ankle stiffness data for a specific rotation amplitude value. The signs from the legend represent the average value for publication. In the case of the publication by Vlutters et al. [4], the authors presented the value range from the lowest to the highest recorded for the perturbation. In the chart, this data is presented by the range of values.

Stiffness was calculated as the difference in torque divided by the difference in the angle between the lower leg and the platform. This method shows higher values for ankle stiffness compared to the (pseudo) ankle stiffness estimated by dividing the difference in torque by the difference in the platform encoder angle and compared to the measurements obtained by Loram and Lakie [10] and Vlutters et al. [4]. The latter estimated the value of ankle stiffness with the two methods mentioned above, proving the theorem about the difference in measurement value.

#### 4. Measuring devices

For estimating ankle stiffness, the authors created their own machines. These machines were designed to apply specific types of perturbations. We believed it would be more comprehensible to present the components of the measuring devices as a table.

**Table 2.** The components of the measuring devices

References	Platform	EMG- measured muscles	Measuring devices and simplified procedures
Loram & Lakie (2002)	Piezo-electric translator that drives two footrests.	M. soleus, tibialis anterior, m. gastrocnemius medial and lateral head.	Two areas of ankle stiffness (foot, ankle) were estimated by 15-degree Hall's precision potentiometer and a fixed gain amplifier. An inclinometer measured the absolute angular position. A piezoelectric vibrating gyroscope measured the angular velocity of the pendulum. Load cells recorded torque. Variable reluctance displacement sensor estimated rotation of the left footrest relative to the platform. The piezoelectric gyroscope estimated the speed of the platform relative to the ground. The rotation of the footplate and the data sampling of the left footrest torque was 1000 Hz. Other sensors were sampled at 25 Hz. A laser rangefinder was used to measure foot deformity, ankle rotation, and Achilles length.
Casadio et al. (2005)	Force platform and motorized footplate. Torque motor rotated the platform	Scientists didn't measure EMG activity.	A differential transformer (LVDT Schaevitz mod. E 1000) measured footplate rotation. The PID module controled the motor position with the Pulse Width modulation output. Control loop calculations (including LVDT signal acquisition) were performed at 2 kHz. The load cell was used to estimate the COP shift in the sagittal plane and was sampled at 200 Hz.
Lang et al. (2014)	Foot pedals and bilateral hydraulic actuator used to apply position perturbation to the ankle.	M. soleus, tibialis anterior, m. gastrocnemius medial and lateral head.	Potentiometers were used to measure angular position of the foot pedals. Torque transducers were used to collect torque. Redundant safety mechanisms ensure safety.
Vlutters et al. (2015)	Foot pedals and bilateral hydraulic actuator used to apply position perturbation to the ankle.	M. soleus, tibialis anterior, m. gastrocnemius medial and lateral head.	Potentiometers were used to measure angular position of the foot pedals. Torque transducers were used to collect torque. Redundant safety mechanisms ensure safety.

Amiri et al. (2017)	Two independent foot pedals driven by electro-hydraulic rotatory actuators.	M. soleus, tibialis anterior and lateral head of m. gastrocnemius.	Transducers measured torque generated by each actuator and the angle of pedals. Two laser range finders measured shank angles. A range finder recorded linear displacement of the back. All signals were filtered at 400 Hz and sampled at 1KHz.
Sakanaka et al. (2018)	Motorized platform supporting two footplates. Liner motor rotated the platform.	M. soleus, tibialis anterior and gastrocnemius EMG activity but only from medial head.	Load cells measured torque for each ankle. A potentiometer measured the anteroposterior rotation of the footplate. The accelerometer measured left footplate acceleration. Two laser-reflex sensors tracked the level of the anteroposterior shin and body tilt.

## 5. Discussion

The objective of this systematic review of the measurement of intrinsic ankle stiffness was to synthesize the research findings in order to draw broad conclusions.

The highest values were reported for free standing and perturbations with an amplitude of 0.055 degrees. As the amplitude of the perturbation increases, the stiffness of the ankle joint decreases significantly. This trend continues up to a value of 0.6 degrees. After exceeding this value, one can observe a flattening tendency. This indicates a smaller decrease in ankle stiffness. In the work of Vlutters et al. [4], the maximum amplitude of platform rotation was 0.08 rad. Its value falls within the range obtained for perturbations of 0.02 rad amplitude. We hypothesize that the decrease in intrinsic ankle stiffness is smaller for perturbations with larger amplitudes, specifically for perturbations with amplitudes greater than 0.6 degrees. The research findings indicate that ankle stiffness is greater for movements that are slower and shorter.

In Chapter 3, results from four publications were used. This is because Amiri et al. [8] reported the values from their work using Nm/rad when background torque was considered. Due to the lack of anthropometric data on the subjects, we are unable to present these findings in the form of a frequency distribution (percentage mgh). A paper by Lang et al. [9] is the second most widely discussed publication in this review due to its innovative measurement method. The authors separated the procedure into three parts (Intrinsic Stiffness, Reflex Stiffness, and Background Muscle Activation). Two trials of each type were presented: normal, backward leaning, and forward leaning. According to the results provided by the authors, the overall value for stiffness was as follows: the maximum contribution of each leg to the critical value varied between 26 and 45 percent across subjects. When the maximum observed stiffness values for each leg were added together, the sum of the intrinsic stiffness values ranged from 52 to 80 percent of the critical value. However, this prevents us from comparing the results with the remaining measurements.

### 5.2. Effect of background torque on the estimation of ankle stiffness

The section on measurement procedures describes the measurement methods that account for the subject's initial position on the platform. There were substantial differences in the research procedures of Casadio et al. [1] and Vlutters et al. [4]. Estimates are affected by the standardization of the starting position and its effect on the background torque. Due to individual differences in height and weight, it was not possible to establish a single target torque that would cause all participants to behave like a rigid body (Sakanaka et al. [3]). The authors then described the variations in ankle and body angle for each experiment. In the first experiment, as participants leaned forward, their ankle and body angles increased. As reported, these angles did not increase by the same amount, indicating that the subjects did not act as rigid bodies. Therefore, the authors decided to calculate the average torques for each subject and plot them as separate traces. Typically, tests of ankle stiffness assume that intrinsic stiffness is constant when standing; this value was estimated in the study by averaging the responses. Loram and Lakie [10] found little change in intrinsic stiffness with an increase in active torque, with an average increase of 13.9 Nm in ankle torque from the normal to the leaning condition. Casadio et al. [1] discovered a 33.5% increase in intrinsic stiffness mgh. Consequently, with a marginally smaller increase in ankle torque, they discovered a greater increase in intrinsic stiffness. On the other hand, Loram & Lakie et al. [10] and Sakanaka et al. [3] described the observed situation, which highlights the need for the precise determination of background torque. They anticipated a smaller difference in stiffness for the smallest perturbations as the active ankle torque increased; however, this interaction was not observed.

The increased ankle stiffness during measurement is caused by the ankle angle rather than ankle torque [3]. As a hypothetical explanation, the authors refer to the consequence of either the coactivation of antagonist muscles, increase of calf muscle moment arm, unmeasured change in the knee angle acting on the biarticular gastrocnemius muscle, or the increased resistance of other tissues [3].

The correct separation of background torque will allow for a more precise evaluation of ankle stiffness. Amiri and Kearney [8] presented an innovative solution to the problem of background torque. The development of an IBK model demonstrated, in the opinion of the authors, that the ankle exhibits a range of stiffness values during stance and is, therefore, a time-varying system. The torque recorded during the intrinsic phase is a combination of background torque (i.e., the torque due to postural sway) and the response of the intrinsic ankle stiffness. In the pre-response phase, a linear regression was performed with COM angle as the independent variable and total torque as the output, assuming a linear relationship, in order to estimate the background torque in each response. After extrapolating the estimated line to the intrinsic response phase, the total torque was subtracted to obtain the intrinsic torque. The average joint torque during the pre-response was used as a measure of the response's background torque [8]. Separating background torque, according to the authors, is essential for accurately assessing ankle stiffness. The results indicate that there is a wide range of ankle background torque due to postural sway, which is accompanied by wide variations in the elastic component of intrinsic stiffness (K) [8].

Lang and Kearney [9] also emphasized the significance of separating the background torque for each individual measurement. In their publication, the answer was separated into four periods. The first served to model the background torque trend by fitting a linear model to the first 25 milliseconds of torque data. Reflex responses were separated by at least 500 ms and lasted less than 300 ms after a perturbation, according to the authors. Therefore, they did not influence the estimation of background torque. This pattern was extrapolated to 100 milliseconds and subtracted from the torque response [9]. For all subjects, intrinsic rigidity increased as background torque decreased (i.e., as the COP moved toward the toes).

### 5.3. EMG activity during perturbation and in the free-standing position

Muscle activation during the intrinsic ankle foot stiffness measurement affects the measurement values. To study this parameter, researchers use EMG to estimate the response time of the muscles in the lower leg and foot. Casadio et al. [1] used 150 ms perturbations for their research, with the perturbation evoking the short-latency reflex activation of the muscle. The authors stated that the effect of muscle activation on the final stiffness estimate is minimal. This conclusion was supported by the absence of statistically significant differences in three estimation methods that attributed a different weight to the final part of the response, as well as by the fact that the estimates were symmetric in relation to the perturbation direction [1]. This claim was not corroborated by Stein and Kearney [11] as they reported that for an ankle stretch reflex, the raw gastrocnemius EMG would have a latency of  $\sim 40$  ms and the torque response would have an onset latency of  $\sim 75$  ms reaching a peak value after  $\sim 170$  ms. Information on the earlier activation of the muscles of the lower leg was also provided by Grey et al. [12]. The authors found that a short-latency reflex activity in the human soleus muscle occurs approximately 40 ms after the onset of the stretch.

In the work of Loram and Lakie [10], the duration of the disturbance was 140 ms, thereby also exceeding the above-described limit of the activation of the calcaneus muscles. However, the authors reported that the EMG used in the study did not show any evidence of muscle activity during the application of perturbations. On the other hand, when the examined subjects stood in quiet standing, they recorded the slight activity of the soleus and gastrocnemius medialis and the possible activity of the tibialis anterior. In the integrated EMG record, this reaction started approximately 100 ms after the start of the dorsiflexion and reached a peak approximately 200 ms after the start of the dorsiflexion. For the standing activity, there was a corresponding torque reaction in both the right and left leg. As the reaction occurred in both legs, it was not a stretch reflex. However, when the subjects were standing or balancing the inverted pendulum, but not when strapped and maintaining constant levels of torque, there was a very interesting longer latency reaction in the triceps surae and tibialis anterior [10]. Vlutters et al. [4] presented in their study the view that until now, reflex activity has not been ruled out from the ankle stiffness estimates by applying sufficiently fast rotations during the stance. The authors used a 40 ms perturbation in the experiment. When examining the activity of the GM muscle, they showed a short latency reflex activity starting 45 ms after the perturbation onset. The authors explained that muscle reflex activity could not influence the torques used to estimate stiffness. Although the onset of the stretch reflex started within the time windows used for the estimates, it did not bias the torque in those windows due to the muscle's electromechanical delay [4]. This delay was shown to be in the order of 30 ms for the knee extensor muscle [13]. The authors noted the need to shorten the disturbance time in future measurements of ankle stiffness.

### 5.4. Unmeasured change in knee angle

Sakanaka et al. [3] noted that when leaning forward, there may have been a slight, unintended, and unmeasured extension of the knee joint, which could have caused a pull on the biarticular gastrocnemius muscle and contributed to increased ankle stiffness. Nevertheless, the contribution of the knee is probably minimal. Herbert et al. [14] dem-

onstrated that the correlation between the lengthening of the gastrocnemius muscle-tendon unit and joint rotation was significantly weaker for the knee than for the ankle [3]. Ankle rotation stretches the gastrocnemius by an average of 0.83 mm per degree, while knee rotation stretches it by only 0.23 mm per degree [14]. Sakanaka et al. [3] reported that the gastrocnemius would be lengthened by significantly less than 3 mm under the perturbations used in calculating the possible ranges of knee extension. The authors also stated that it is very small compared to the intended lengthening produced by ankle dorsiflexion, which would be nearly 17 mm. The effect of knee deflection on the estimation of ankle stiffness should be considered in future studies.

### 5.5. Participation of the foot in the estimation of ankle stiffness

In an article published by Loram and Lakie [10], the authors measured the stiffness of the foot. They reported that foot stiffness tended to decrease and become more consistent with increasing torque and that this partly offsets an increase in true ankle stiffness. The authors suggested that in using perturbations smaller than a certain critical size, the increasing compliance of the foot and soft tissues conceals the rise in stiffness associated with torque-induced tendon stiffening, setting a rather high and constant level of stiffness. The authors stated that this may be particularly relevant to quiet standing where many of the spontaneous sways tend to be very small in size. In this regimen, ankle stiffness may be effectively independent of torque level (muscle activity) and perhaps for very tiny perturbations or sways, stiffness maintains a constant level because the compliance of the foot and soft tissues acts as a relatively constant stiffness buffer [3]. In a study published by de Vlugt et al. [15], a model fitted to the human wrist was developed. The authors showed that the short-range stiffness of the entire musculotendinous complex did not change at angular velocities ranging from 1 to 4 rad/s. Their data showed that regardless of the stretching speed, the short-range stiffness was maintained for 30 ms. The authors suggested that the short-range stiffness is also independent of the rotational amplitude within this velocity range. Assuming this property holds in our velocity range (0.125–2 rad/s), the observed decrease might be attributed to changes in muscle stiffness occurring after short-range stiffness effects. Initially, the entire muscle-tendon complex might behave as a spring with constant stiffness, after which the muscle stiffness decreases due to detaching cross-bridges [4]. Loram and Lakie [10] suggested that the compliance of the foot means that the axis of rotation of the body COM is not a fixed center through the ankle joint. The visual observation seems to confirm that the axis of rotation moves forward as the body sways forward and more torque is transmitted through the foot. The authors also stated that this may mean that for small sways close to vertical, the toppling torque per unit angle is less than it would be if the center of rotation did not move. Thus, for such sways, the intrinsic mechanical stiffness could confer more stability than their calculations show. The authors further suggested that this mechanism should be measured and included for further estimation.

### 5.6. Plantarflexion and dorsiflexion in the measurement of intrinsic ankle stiffness

The influence of the direction of the perturbation on ankle stiffness has been investigated. The authors emphasized that they did not note the influence of plantarflexion and dorsiflexion direction on the estimation values. Therefore, they combined perturbations of both directions for the final calculation [3]. Vlutters et al. [4] agreed with this statement. However, the authors noted an exception for perturbations with an amplitude of 0.005 rad. For the remaining conditions, there were no differences between plantarflexion and dorsiflexion. Casadio et al. [1] also confirmed the lack of a significant correlation between gender and ankle stiffness. They found no significant effects for gender and the direction of the perturbation.

## 6. Conclusion

The primary goal of estimating ankle stiffness was to evaluate and understand the effect of this parameter on balance loss and recovery. Establishing what determines the intrinsic ankle stiffness required a thorough analysis of the tissue properties combined with the concept of stabilizing the inverted pendulum in the ankle joint. Joint stiffness defines the relationship between the position of a joint and the torque acting on it. According to Sakanaka et al. [3], ankle stiffness is the sum of the compliances of the structures that are deformed when the ankle is rotated. Loram and Lakie [10] consider intrinsic ankle stiffness to be the instantaneous mechanical stiffness provided by the combination of active muscle, tendon, connective tissue, and foot, and when one is measuring the stiffness combination of springs in a series, such as the muscle fibers and the tendon, the value of stiffness is limited by the weakest spring stiffness. Sakanaka et al. [3] claimed that stiffness is associated with ankle angle rather than ankle torque, thus an alternative explanation for the rise in stiffness is necessary. The authors concluded that this



could be a consequence of either the coactivation of antagonist muscles, increase of calf muscle moment arm, unmeasured change in knee angle acting on the biarticular gastrocnemius muscle, or the increased resistance of other tissues crossing the ankle. The entire muscle-tendon complex can act as a spring until the value at which the myosin bridges detach is exceeded.

In rehabilitation diagnostics, ankle stiffness has become a crucial variable. Its contribution to equilibrium is evident. Further research into passive stiffness and the correlation between the passive and active systems may bring us closer to comprehending the intricate mechanism underlying the operation of the entire system. Ankle stiffness is important in numerous areas of research, and the parameter can be utilized effectively in physiotherapy. Utilizing a training intervention and studying its effect on ankle stiffness across age groups is an intriguing concept (e.g. in seniors). Additional research is required to implement the stiffness parameter of the ankle joint.

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