Journal of Biomechanics 48 (2015) 1258–1263

Contents lists available at ScienceDirect

Journal of Biomechanics

journal homepage: www.elsevier.com/locate/jbiomech
www.JBiomech.com

Direct measurement of the intrinsic ankle stiffness during standing

M. Vlutters [a,*], T.A. Boonstra [a], A.C. Schouten [a,b], H. van der Kooij [a,b]

[a] Department of Biomechanical Engineering, University of Twente, The Netherlands
[b] Department of Biomechanical Engineering, Delft University of Technology, The Netherlands

ARTICLE INFO

Article history:
Accepted 7 March 2015

Keywords:
Human balance
Ankle
Joint rotations
Muscle–tendon stiffness
Perturbed standing

ABSTRACT

Ankle stiffness contributes to standing balance, counteracting the destabilizing effect of gravity. The ankle stiffness together with the compliance between the foot and the support surface make up the ankle-foot stiffness, which is relevant to quiet standing. The contribution of the intrinsic ankle-foot stiffness to balance, and the ankle-foot stiffness amplitude dependency remain a topic of debate in the literature. We therefore developed an experimental protocol to directly measure the bilateral intrinsic ankle-foot stiffness during standing balance, and determine its amplitude dependency. By applying fast (40 ms) ramp-and-hold support surface rotations (0.005–0.08 rad) during standing, reflexive contributions could be excluded, and the amplitude dependency of the intrinsic ankle-foot stiffness was investigated. Results showed that reflexive activity could not have biased the torque used for estimating the intrinsic stiffness. Furthermore, subjects required less recovery action to restore balance after bilateral rotations in opposite directions compared to rotations in the same direction. The intrinsic ankle-foot stiffness appears insufficient to ensure balance, ranging from 0.93 ± 0.09 to 0.44 ± 0.06 (normalized to critical stiffness 'mgh'). This implies that changes in muscle activation are required to maintain balance. The non-linear stiffness decrease with increasing rotation amplitude supports the previous published research. With the proposed method reflexive effects can be ruled out from the measured torque without any model assumptions, allowing direct estimation of intrinsic stiffness during standing.

1. Introduction

Human standing balance is continuously challenged by gravity, which imposes a negative stiffness on the upright equilibrium posture. This "critical stiffness" must be compensated to maintain upright stance. The ankles' muscle–tendon structures provide stiffness at multiple levels. First, stretch reflexes can lead to changes in muscle activation levels and affect joint stiffness (Sinkjaer et al., 1988). Second, the muscle–tendon complex provides a direct mechanical torque response to stretch (Rack and Westbury, 1974). This intrinsic stiffness depends on the tonic activation level, which influences the muscle's mechanical properties through cross-bridge formation. Cross-bridges are thought to cause a high short-range stiffness due to elastic stretch (Morgan, 1977; Rack and Westbury, 1974), and a lower long-range stiffness by detaching and sliding muscle filaments (Campbell and Lakie, 1998). Separating the reflexive and intrinsic contributions to the overall ankle stiffness during standing might give insight into neuromuscular disorders, and could help in the assessment of balance control in a clinical setting.

Ankle stiffness can be estimated by applying a rotation to the foot and measuring the torque response. In the current literature there are various definitions of ankle stiffness, which are here distinguished as: (1) The actual ankle stiffness, which can be estimated using the rotation between the lower leg and the foot. (2) The ankle-foot stiffness, which can be estimated using the rotation between the lower leg and the contact surface of the device used to apply a rotation to the foot. This includes both the ankle stiffness and possible foot compliance. (3) The pseudo ankle-foot stiffness, which can be estimated using only the rotation angle of the foot contact surface, assuming no lower leg movement.

Ankle stiffness, in general, has been investigated using a wide variety of conditions. Stiffness varies with muscle contraction level (Hunter and Kearney, 1982), mean joint angle (Gottlieb and Agarwal, 1978; Weiss et al., 1986) and rotation amplitude (Kearney and Hunter, 1982). In the latter study, pseudo-random binary sequence rotations varying from 0.01 to 0.25 rad were applied to the left foot in supine subjects. The pseudo ankle-foot stiffness decreased with increasing rotation amplitude, and both intrinsic and reflexive mechanisms contributed to the results. Later, in Kearney et al. (1997) system identification methods were applied to separate intrinsic and reflexive components. It was

* Correspondence to: De Horst 2, W215, 7522LW Enschede, The Netherlands. Tel.: +31 534892896.
E-mail address: m.vlutters@utwente.nl (M. Vlutters).

http://dx.doi.org/10.1016/j.jbiomech.2015.03.004
0021-9290/

concluded that reflexive contributions depend strongly on the conditions, and that the generated reflexive torques can be of the same magnitude as those from intrinsic mechanisms. Similar proportions were reported in hemiparetic patients (Sinkjær and Magnussen, 1994), where nerve stimulation was used to suppress reflexive activity.

In a limited number of studies ankle stiffness was estimated in upright stance, where stiffness is often expressed as “relative stiffness”, i.e. normalized to the critical stiffness. In Peterka (2002), subjects mimicked a single-link inverted pendulum during backboard supported stance. A pseudo-random ternary rotation sequence of 0.009–0.14 rad was applied to the support surface. Parametric estimates resulted in a relative ankle-foot stiffness of approximately 0.15. The work of Loram and Lakie (2002) described the use of a piëzo-electric element to apply 0.001 rad rotations to the left foot during both free and backboard supported stance. Cosine waves with a rise time of 70 ms were used. A relative pseudo ankle-foot stiffness of 0.91 was found by using parametric estimates. In a subsequent study, values of 0.67 and 0.54 were found for slow (>1 s) 0.003 and 0.007 rad rotations respectively, using a similar setup (Loram et al., 2007a, 2007b). Transient rotations of 0.02 rad and a rise time of 150 ms were used in Casadio et al. (2005). Subjects were freely standing on a footplate capable of perturbing both feet simultaneously. Various estimation methods were attempted to minimize potential effects of short latency reflex activity and lower leg movement, leading to a relative ankle-foot stiffness of 0.64.

Short latency reflex activity in human soleus muscle occurs approximately 40 ms after stretch onset (Grey et al., 2001). Until now, reflex activity has not been ruled out from the ankle stiffness estimates by applying sufficiently fast rotations during stance. Although several previous studies suggest that the relative (pseudo) ankle-foot stiffness is lower than 1, the intrinsic contribution remains uncertain. Here the goal is to directly quantify the rotational amplitude dependency of the intrinsic (pseudo) ankle-foot stiffness in healthy subjects during stance. By applying 40 ms ramp-and-hold plantar- and dorsiflexion rotations to both ankle joints simultaneously, reflex activity will be removed from the stiffness estimates. What remains is the intrinsic stiffness that can be estimated directly, without model assumptions. Furthermore, simultaneously applying a plantar flexion to one ankle and a dorsiflexion to the other might prevent disturbing the subject's balance during the experiment.

2. Methods

2.1. Participants

Eight healthy volunteers with no known history of neurological or muscular disorders participated in the study (7 men, age 23 ± 1 years, weight 75 ± 8 kg, height 1.85 ± 0.07 m, mean $\pm$ sd). All subjects gave prior written informed consent in agreement with the guidelines of the local ethical committee, and in accordance with the Declaration of Helsinki.

2.2. Apparatus

Rotations were applied to both ankle joints using the bilateral ankle perturbator (BAP) as shown in Fig. 1. A detailed description of the apparatus can be found in Schouten et al. (2011). The device consists of two lightweight platforms, each connected to an electromotor (HIWIN, IL; type TMS3C) via a lever arm. The lever arms can be adjusted to align the subject's ankle joints with the rotational axis of the motors. As a safety measure, an ultrasound sensor was incorporated in each platform to check for heel contact. Rotations could not be applied if the subject did not make heel contact with the sensor. Furthermore, a safety harness connected to the ceiling with a belt and locking retractor was worn around the chest to prevent injury in case of a fall. The harness did not provide any support while standing on the BAP.

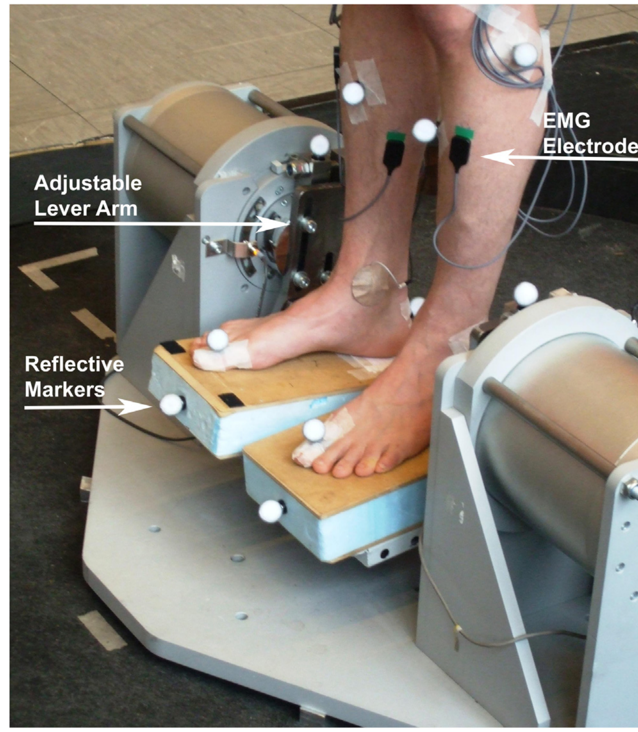

Fig. 1. The Bilateral Ankle Perturbator (BAP). Rapid 40 ms plantar- and dorsiflexions were simultaneously applied to both feet using the BAP. The lever arms can be adjusted to align the subject's ankle joints with the motor axis.

Force transducers (Revere Transducers Inc, CA; type ALC-C2) between each lever arm and motor were used to measure the torque exerted on each platform. Platform angular displacement and velocity were measured using rotary encoders (2.5×10^{-4} rad accuracy). All BAP data was captured at 10 kHz using a DAQ-card (HUMUSOFT, Czech Republic, MF624) running xPC-target (The Mathworks, Natick, US). Kinematic data was captured at 120 Hz using a 6-camera VICON system (Oxford Metrics, Oxford, UK) and 20 reflective markers. Markers were placed at the acromion, femur head, lateral epicondyle, tibia, lateral malleolus, calcaneus and metatarsal 1 head on the left and right side of the body, as well as on top of each lever arm, and on the front and back of each platform. Activity patterns of the tibialis anterior (TA), soleus (SO), gastrocnemius medialis (GM) and gastrocnemius lateralis (GL) muscles were recorded using surface EMG electrodes (Delsys Inc, Natic, MA). EMG data was amplified (1000x) and captured at 1560 Hz using the AD converter of the Vicon.

2.3. Experimental protocol

Subjects stood on the BAP platforms, and were instructed to keep knees and hip in an extended position. Arms were held over the chest. Subjects leaned slightly forward to reduce effects of natural sway and achieve a consistent ankle angle at perturbation onset. A screen in front of the subject gave visual feedback on a target ankle torque and the exerted ankle torques. The target torque for each ankle was derived from a simple linearized inverted pendulum equation

$$T_{\text{target}} = (m \times g \times h \times \varphi)/2$$

where m is the subject's mass (kg), g the earth's gravitational constant (m/s^2), h the subject's estimated center of mass (COM) height (m) and φ the desired subject lean angle from the vertical (rad). The term $m \times g \times h$ is equal to the critical stiffness (Casadio et al., 2005), being the minimum intrinsic ankle-foot stiffness required for stabilization without changes in muscle activity. The COM height was estimated using a weighted average of body segments (Winter, 2009). For all subjects φ was set to 0.07 rad (4°). This led to a subject average target torque of 26 ± 3 N m per ankle.

A ramp-and-hold rotation was applied when the torque exerted on each platform was held within 10% of the target torque for a random time interval of 2–4 s. To allow non-parametric intrinsic stiffness estimation, 40 ms minimum-jerk profiles (Burdet et al., 2000) were used. These ensure (near) zero velocity and acceleration at the start and end of the perturbation, such that damping and inertial effects of the platforms and feet are minimized. Both plantar- and dorsiflexion rotations of 0.08, 0.04, 0.02, 0.01 and 0.005 rad were applied. Either unidirectional rotations (UR) or bidirectional rotations (BR) were used, for which the absolute amplitude within one condition was equal for both platforms.

Each condition was repeated eight times, resulting in 160 trials per subject. All rotations were applied in a randomized order to prevent anticipation or learning effects. Two seconds after rotation onset the platforms returned to their neutral position in one second. The subject could subsequently try to reach the target torque again for the next perturbation. To prevent muscle fatigue, subjects were instructed to sit down and rest for a short period after every 6 min of measurements. Baseline trials were collected by applying the rotations without a subject standing on the BAP.

2.4. Data analysis

All data were processed using Matlab (R2012b, The Mathworks, Natick, US). Subject trials for which the average torque before onset deviated more than 10% from the target torque were disregarded. The torque responses of the baseline trials were averaged over the eight repetitions. These were subtracted from all corresponding subject torque responses to correct for gravitational effects of the platforms. Baseline corrected torques were subsequently filtered with a 2nd order 150 Hz zero-phase Butterworth filter.

Raw marker data were used to get an indication of the platform, feet and lower leg segment rotations. All markers that moved more than 10 cm in any direction during a window of 100 ms before to 200 ms after perturbing were considered outliers. The corresponding trials were removed from further analysis. The toe, heel and malleolus data were interpolated between 30 and 70 ms after perturbation onset to deal with (skin) movement artefacts resulting from the perturbation. Segment rotations were calculated using the marker displacements averaged over the eight repetitions. Rotations of the feet could not be calculated accurately for the smaller perturbations (<0.04 rad), given the accuracy of the motion capture device. Consequently, the actual ankle stiffness was not further analyzed. Marker data were filtered with a 20 Hz 2nd order zero-phase Butterworth filter to estimate the COM position and its deflection angle from the vertical passing through the ankle joint markers.

The intrinsic pseudo ankle-foot stiffness for each ankle was estimated by dividing the difference in torque by the difference in platform encoder angle. The intrinsic ankle-foot stiffness for each ankle was estimated by dividing the difference in torque by the difference in the angle between lower leg and platform. For both estimates, differences were calculated for each trial using the average signal values before (-15 to -5 ms) and after (45–55 ms) rotation onset. The relative intrinsic (pseudo) ankle-foot stiffness was computed from the sum of the left and right ankle joint stiffness, divided by the subject's critical stiffness. For each subject and each condition an average relative intrinsic stiffness was calculated over the eight repetitions. These averages were used to compute a between-subjects standard deviation.

A statistical linear mixed model with fixed effects for amplitude, direction (plantar- and dorsiflexion) and similarity (UR and BR) was used to investigate their effects on the intrinsic (pseudo) ankle-foot stiffness. Subject effects were specified as random. Effects of rotation direction were only investigated for UR, since the total stiffness calculated for BR already consisted of both plantar- and dorsiflexions. Given the test results, plantar- and dorsiflexions were considered equal and were pooled for further analysis. Subsequently, differences in stiffness between UR and BR were investigated, as well as the effect of the rotation amplitude on the pooled UR and BR stiffness. A significance level of $\alpha=0.01$ was used and a Bonferroni correction was applied to correct for multiple comparisons. Finally, a least squares fit of the form $y=a\times{}^{10}\log(x)+b$ was made to the rotation amplitude and the pooled UR and BR intrinsic (pseudo) ankle-foot stiffness per amplitude.

EMG data were detrended and filtered with a 1st order 48–52 Hz bandstop Butterworth filter. These data were subsequently rectified, filtered with a 1st order 40 Hz low-pass Butterworth filter, and cut into sequences from 1 s before to 2 s after perturbation onset. EMG averages were calculated over the eight repetitions of each condition, as well as over all subjects for each condition.

3. Results

3.1. Subject responses

The 40 ms minimum-jerk ramp-and-hold rotations were completed before short latency reflex activity occurred. This can be seen in Fig. 2, where the subjects' average response to the UR and BR conditions is shown for left 0.08 rad dorsiflexions. The GM muscles showed short latency reflex activity, starting approximately 45 ms after perturbation onset. The amplitude of the reflex response is slightly lower for UR compared to BR. The GL and SO muscles showed similar effects as those of the GM. The TA muscles showed little to no short latency responses.

The difference in torque before and after perturbing was found slightly higher for all BR compared to UR. For all 0.08–0.005 rad BR the average torque differences ranged from 13.70 ± 2.87–1.23 ± 0.51 N m respectively, compared to 11.60 ± 2.78–1.12 ± 0.48 N m respectively for UR. Subjects required a larger balance recovery response for UR compared to BR. This is reflected in the torque and EMG signals, as well as in the COM angle with the vertical. After approximately 150 ms, an increase in TA EMG can be observed for UR but not for BR. This is followed by a dorsiflexion torque, which is higher for UR compared to BR. Furthermore, the COM shows more deflection from its initial position for UR compared to BR.

Finally, the segment rotation angles as calculated from the marker data revealed lower leg rotations, and foot rotations smaller than those of the platforms. The average rotation angle between platform and lower leg is 5 ± 2, 10 ± 2, 16 ± 6, 23 ± 6 and 26 ± 9% less than the applied platform rotations of 0.08–0.005 rad respectively. Consequently, the encoder angle is not an accurate representation of the angle between platform and lower leg over the full perturbation range.

3.2. Intrinsic ankle stiffness

Of the total of 1280 trials, 188 were disregarded because the average torque before perturbing deviated more than 10% from the target torque. For the intrinsic ankle-foot stiffness estimates, another 148 trials were disregarded for inaccurate marker data. Fig. 3 shows the amplitude dependency of the relative intrinsic (pseudo) ankle-foot stiffness for plantar- and dorsiflexions of the UR. Plantar- and dorsiflexions for UR tested significantly different for the 0.005 rad ($p<0.001$) rotations, with a mean difference of 0.13 and 0.37 for the pseudo ankle-foot stiffness and ankle-foot stiffness respectively. No significant differences between plantar- and dorsiflexions were found for the other amplitudes ($p=0.040$ for 0.01 rad in the pseudo ankle-foot stiffness, $p>0.116$ for all others).

Fig. 4A shows the relative intrinsic pseudo ankle-foot stiffness for the BR and the UR with pooled plantar- and dorsiflexions. Several values found in the literature are also included in the figure. Loram et al. (2002, 2007a, 2007b) mainly applied smaller rotations than in this study. Their results are in proximity of the fit, but suggest a steeper descending slope. On average, the relative intrinsic pseudo ankle-foot stiffness ranged from 0.67 ± 0.05 for 0.005 rad rotations to 0.42 ± 0.06 for 0.08 rad rotations. Parameter values of the fit $y=a\times{}^{10}\log(x)+b$ are $a=-0.21$ and $b=0.21$. The fit has a coefficient of determination (R^2) of 0.97 and a root mean square error of 0.02. The fit extrapolates to 1 for approximately 10^{-4} rad rotations (not shown).

Fig. 4B shows the relative intrinsic ankle-foot stiffness calculated using the marker segment rotation angles. Especially for the smaller rotations the stiffness is higher compared to those found using the encoder angle, leading to a steeper decrease in stiffness with increasing rotation amplitude. The result at 0.02 rad found by Casadio et al. (2005) is in accordance with the fit. On average, the intrinsic ankle-foot stiffness ranged from 0.93 ± 0.09 for 0.005 rad rotations to 0.44 ± 0.06 for 0.08 rad rotations. Parameter values of the fit are $a=-0.38$ and $b=0.01$. The fit has an R^2 of 0.99 and a root mean square error of 0.01. The fit extrapolates to 1 for 2.5×10^{-3} rad rotations.

For both estimates, the intrinsic stiffness during standing decreased non-linearly with increasing rotation amplitude. With the exception of the ankle-foot stiffness for the 0.005 rad rotations, the UR yielded a consistently lower stiffness than the BR. These tested significantly different with a mean difference of 0.06 ($p<0.001$) and 0.05 ($p=0.007$) for the pseudo ankle-foot stiffness and ankle-foot stiffness respectively.

For pooled UR and BR, the relative intrinsic pseudo ankle-foot stiffness at each applied rotation amplitude tested significantly different from the stiffness at all other amplitudes ($p<0.003$). For

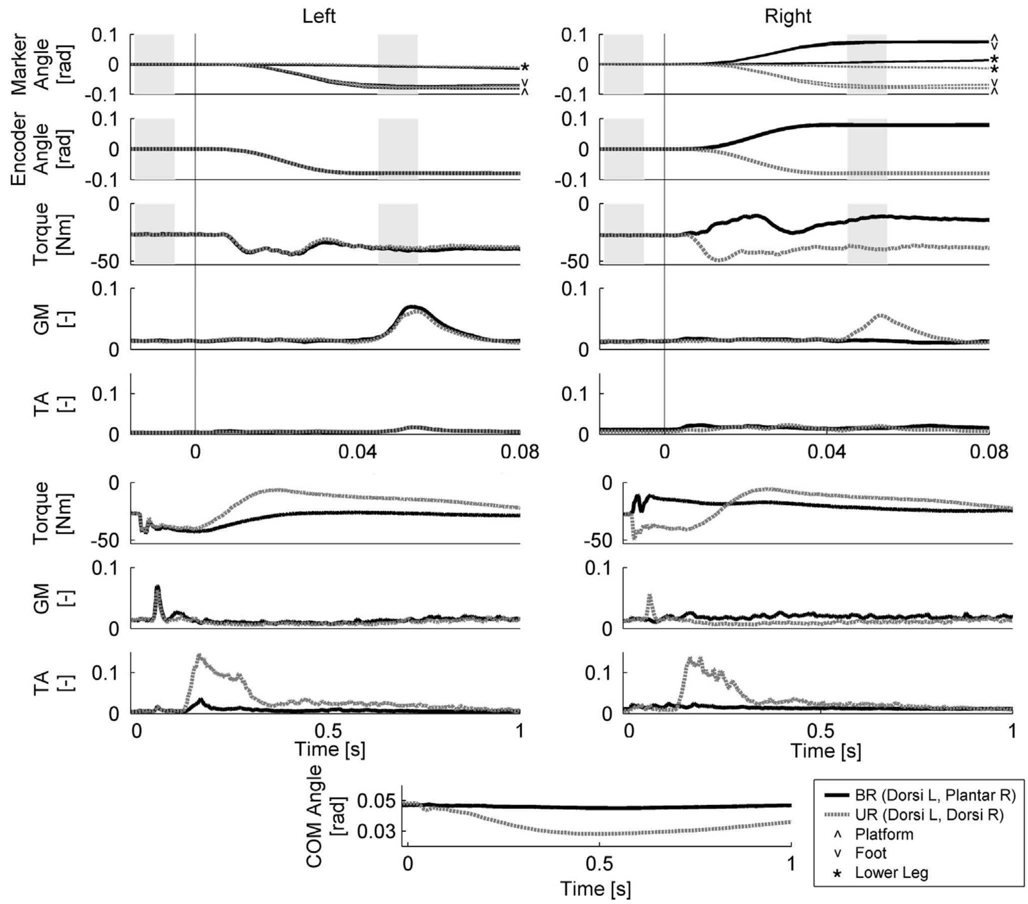

Fig. 2. Time series average over all subjects' bidirectional rotations (BR, solid black) and unidirectional rotations (UR, dashed gray) for left 0.08 rad dorsiflexion trials. The top row shows the rotation angles of the platforms (arrow up), feet (arrow down) and lower legs (star). The next four rows show the applied rotation angle, measured torque, and GM and TA EMG responses for the left and right ankle. The vertical line indicates perturbation onset. A negative torque corresponds to a plantar flexing torque exerted on the BAP platforms. The middle three rows show the same torque and EMG responses on a longer time scale. The bottom graph shows the COM deflection angle with the vertical axis from the ankle joints. For each trial the average signal values over the shaded areas were used for stiffness estimation.

the relative intrinsic ankle-foot stiffness these differences tested significant as well ($p<0.002$), with the exception of that between the 0.08 and 0.04 rad rotations ($p=0.024$).

4. Discussion

4.1. Reflexive activity

Rapid 40 ms minimum-jerk profiles of various amplitudes were applied to the separate ankle joints in order to directly estimate the intrinsic (pseudo) ankle-foot stiffness during standing. Muscle reflex activity could not influence the torques used in the stiffness estimation. Although stretch reflex onset starts within the time windows used for the estimates, it will not bias the torque in those windows due to the muscle's electromechanical delay. This delay was shown to be in the order of 30 ms for knee extensor muscle (Häkkinen and Komi, 1983).

4.2. Intrinsic ankle stiffness

On average, the relative intrinsic (pseudo) ankle-foot stiffness was found to be lower than 1 for pooled plantar- and dorsiflexions. Assuming that no velocity effects would occur, extrapolating the current findings suggests that the intrinsic (pseudo) ankle-foot stiffness could counteract rotations that are smaller than those occurring in the ankle during natural sway (Aramaki et al., 2001). Hence, a 40 ms externally applied disturbance of nearly any

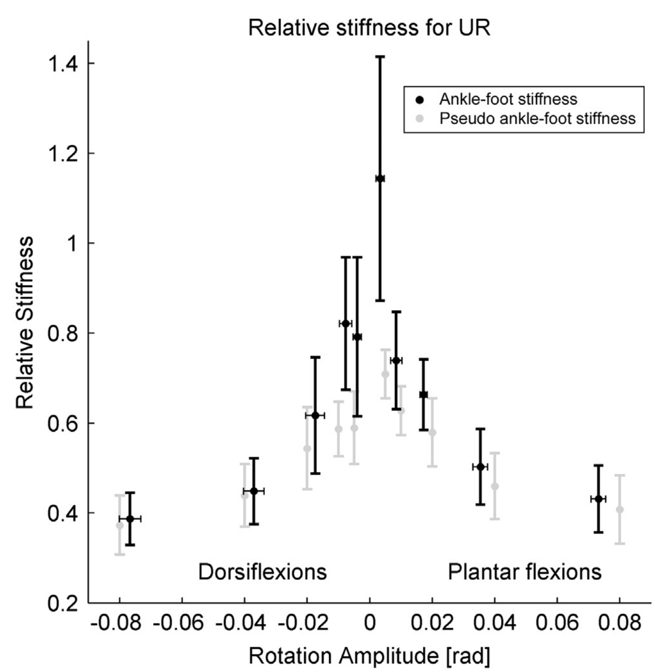

Fig. 3. Rotation amplitude dependency of the subject average relative intrinsic pseudo ankle-foot stiffness (gray) and ankle-foot stiffness (black) for UR, shown separately for plantar- and dorsiflexions. Errorbars indicate the between-subject standard deviation.

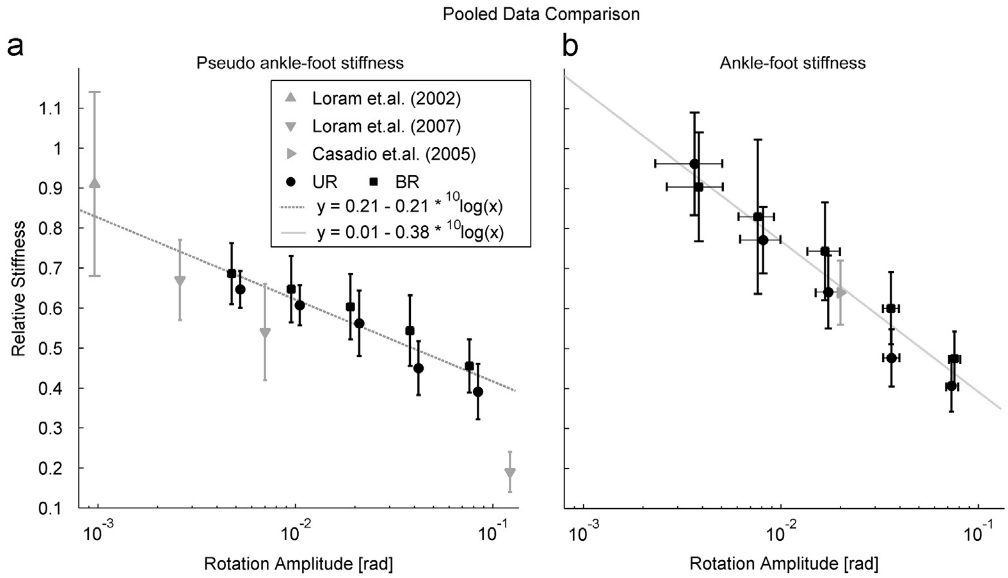

Fig. 4. Rotation amplitude dependency of the subject average relative intrinsic psuedo ankle-foot stiffness (A) and ankle-foot stiffness (B) for BR, and for UR with pooled plantar- and dorsiflexions. Other values from the literature are shown for comparison. Errorbars indicate the between-subject standard deviation. The *x*-scale is logarithmic. The UR and BR data in panel A were slightly shifted on the *x*-axis for clarity, to prevent overlap. The fits were established by a linear least squares estimate on the ^{10}log of the (average) rotation amplitude and the average of the UR and BR relative stiffness.

magnitude cannot be counteracted by the passive ankle structures under these tonic muscle activation levels. These results suggest that the intrinsic ankle-foot stiffness alone is insufficient to maintain upright balance. This is in accordance with previous findings (Loram et al., 2005; Morasso and Sanguineti, 2002). Consequently, changes in ankle muscle activation are required, or another method of control such as a hip strategy must be applied.

The differences with the findings in (Loram et al., 2007a, 2007b) might be explained by the different experimental conditions. In their work subjects were supported at the waist to ensure minimal muscle activity. Here, subjects maintained a small forward lean angle. The higher muscle contraction levels in this experiment can explain the higher stiffness.

The lean angle and corresponding muscle contraction levels are not expected to differ greatly from normal quiet standing. In Winter et al. (2001), an average sway angle during quiet standing of 0.064 rad (3.67°) is reported. This angle was estimated using the whole body COM deflection from the ankle joint. The accompanying average total exerted torque was 55 N m. This is in the same range as our average onset torque (26 N m per ankle).

The actual ankle stiffness could not be accurately estimated for the smaller rotations due to the limited spatial and temporal resolution of the motion capture device. Nevertheless, it is the ankle-foot stiffness that is relevant to standing balance. The higher ankle-foot stiffness found for the 0.005 rad plantar flexion compared to the dorsiflexion is mainly caused by the marker data, as the difference is less pronounced in the pseudo ankle-foot stiffness. A similar high stiffness is not found for the BR. Given the inter-distance of the markers, these rotations are near the accuracy of the motion capture system, possibly causing these effects.

4.3. Rotational amplitude dependency

The observed decrease in stiffness with increasing rotation amplitude is likely caused by the contractile tissue. A Hill-type muscle model would suggests mainly tendon stretch at high velocities due to the viscous properties of the muscle. However, in our protocol the different rotations had a fixed duration, hence the largest are also the fastest. A Hill-type model does therefore not apply to the data, because it cannot predict a decrease in stiffness for increasing stretch velocity. Furthermore, tendon stiffness does not decrease with increasing stretch (Maganaris, 2002), hence cannot explain the observed decrease.

In Loram et al. (2007a, 2007b) it was shown that the contractile tissue is stiffer for smaller rotations, while the stiffness of the series elastic element remained largely constant over various rotation amplitudes. This suggests that the observed decrease relates to the muscle. In de Vlugt et al. (2011), model fits to the human wrist revealed that the short-range stiffness of the total muscle-tendon complex did not vary with angular velocities between 1and 4 rad/s. Their data also showed that short-range stiffness lasted for 30 ms regardless of the stretch velocity. Therefore, 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.

4.4. Effects of rotational velocity

The rotational velocity decreased with decreasing rotation amplitude, possibly introducing a velocity effect. In cat soleus muscle the stiffness was shown to be independent of stretch velocity for stretches faster than 8 mm/s, and decreased for slower movements (Rack and Westbury, 1974). Here, velocity effects would lead to a lower stiffness for smaller rotations. Consequently, applying all perturbations with the same angular velocity could lead to an even steeper decrease in stiffness with increasing rotation amplitude.

4.5. Uni- and bidirectional rotations

The systematic differences in stiffness found for UR and BR might be caused by body movement in response to the rotation. The mechanical coupling of both legs through the pelvis might explain the observed differences in torque response between UR and BR. The UR could lead to slight COM movement in the same direction as the perturbation, whereas BR could lead to rotation about the body's longitudinal axis. Hence, effects of the COM on the measured torque might not be completely disregarded.

4.6. Recommendations

Our results stress the need for accurate measurement of the differences between applied and actual ankle(-foot) angle. This might be improved by using dedicated equipment, such as a laser as in Loram and Lakie (2002). Unwanted effects caused by the rotations, such as a loss of balance, might be reduced by using BR.

Although not the purpose of this study, a further reduction in rotation duration might be required to solely estimate short-range stiffness effects using the current method. We believe applying sufficiently fast rotations to rule out unwanted effects can be a valid method to investigate muscle–tendon properties and their contributions to standing balance. With the proposed method reflexive effects can be ruled out from the torque response, without any model assumptions.

Conflict of interest statement

The authors declare that there is no conflict of interest that could influence the content of the presented work.

Acknowledgments

This work was supported by the BALANCE (Balance Augmentation in Locomotion, through Anticipative, Natural and Cooperative control of Exoskeletons) project, partially funded under grant 601003 of the Seventh Framework Program (FP7) of the European Commission (Information and Communication Technologies, ICT-2011.2.1). This work was supported by the BrainGain Smart Mix Programme of Netherlands Ministry of Economic Affairs and Netherlands Ministry of Education, Culture and Science, funded under grant SSM06011. The funding parties had no role in study design, data collection and analysis, decision to publish, or preparation of the manuscript.

References

Aramaki, Y., et al., 2001. Reciprocal angular acceleration of the ankle and hip joints during quiet standing in humans. Exp. Brain Res. 1, 463–473.

Burdet, E., et al., 2000. A method for measuring endpoint stiffness during multi-joint arm movements. J. Biomech. 12, 1705–1709.

Campbell, K.S., Lakie, M., 1998. A cross-bridge mechanism can explain the thixotropic short-range elastic component of relaxed frog skeletal muscle. J. Physiol. 3, 941–962.

Casadio, M., et al., 2005. Direct measurement of ankle stiffness during quiet standing: implications for control modelling and clinical application. Gait Posture 4, 410–424.

de Vlugt, E., et al., 2011. Short range stiffness elastic limit depends on joint velocity. J. Biomech. 11, 2106–2112.

Gottlieb, G.L., Agarwal, G.C., 1978. Dependence of human ankle compliance on joint angle. J. Biomech. 4, 177–181.

Grey, M.J., et al., 2001. Group II muscle afferents probably contribute to the medium latency soleus stretch reflex during walking in humans. J. Physiol. 3, 925–933.

Häkkinen, K., Komi, P.V., 1983. Electromyographic and mechanical characteristics of human skeletal muscle during fatigue under voluntary and reflex conditions. Electroencephalogr. Clin. Neurophysiol. 4, 436–444.

Hunter, I., Kearney, R., 1982. Dynamics of human ankle stiffness: variation with mean ankle torque. J. Biomech. 10, 747–752.

Kearney, R., Hunter, I., 1982. Dynamics of human ankle stiffness: variation with displacement amplitude. J. Biomech. 10, 753–756.

Kearney, R.E., et al., 1997. Identification of intrinsic and reflex contributions to human ankle stiffness dynamics. IEEE Trans. Biomed. Eng. 6, 493–504.

Loram, I.D., Lakie, M., 2002. Direct measurement of human ankle stiffness during quiet standing: the intrinsic mechanical stiffness is insufficient for stability. J. Physiol. 3, 1041–1053.

Loram, I.D., et al., 2005. Human postural sway results from frequent, ballistic bias impulses by soleus and gastrocnemius. J. Physiol. 1, 295–311.

Loram, I.D., et al., 2007a. The passive, human calf muscles in relation to standing: the non-linear decrease from short range to long range stiffness. J. Physiol. 2, 661–675.

Loram, I.D., et al., 2007b. The passive, human calf muscles in relation to standing: the short range stiffness lies in the contractile component. J. Physiol. 2, 677–692.

Maganaris, C.N., 2002. Tensile properties of in vivo human tendinous tissue. J. Biomech. 8, 1019–1027.

Morasso, P.G., Sanguineti, V., 2002. Ankle muscle stiffness alone cannot stabilize balance during quiet standing. J. Neurophysiol. 4, 2157–2162.

Morgan, D., 1977. Separation of active and passive components of short-range stiffness of muscle. Am. J. Physiol.-Cell Physiol. 1, 45–49.

Peterka, R., 2002. Sensorimotor integration in human postural control. J. Neurophysiol. 3, 1097–1118.

Rack, P.M., Westbury, D., 1974. The short range stiffness of active mammalian muscle and its effect on mechanical properties. J. Physiol. 2, 331–350.

Schouten, A.C., et al., 2011. A bilateral ankle manipulator to investigate human balance control. IEEE Trans. Neural Syst. Rehabil. Eng. 6, 660–669.

Sinkjær, T., Magnussen, I., 1994. Passive, intrinsic and reflex-mediated stiffness in the ankle extensors of hemiparetic patients. Brain 2, 355–363.

Sinkjaer, T., et al., 1988. Muscle stiffness in human ankle dorsiflexors: intrinsic and reflex components. J. Neurophysiol. 3, 1110–1121.

Weiss, P., et al., 1986. Position dependence of ankle joint dynamics—I. Passive mechanics. J. Biomech. 9, 727–735.

Winter, D.A., 2009. Biomechanics and Motor Control of Human Movement. John Wiley & Sons, New Jersey.

Winter, D.A., et al., 2001. Ankle muscle stiffness in the control of balance during quiet standing. J. Neurophysiol. 6, 2630–2633.