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Direct measurement of the intrinsic ankle stiffness during standing

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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

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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.

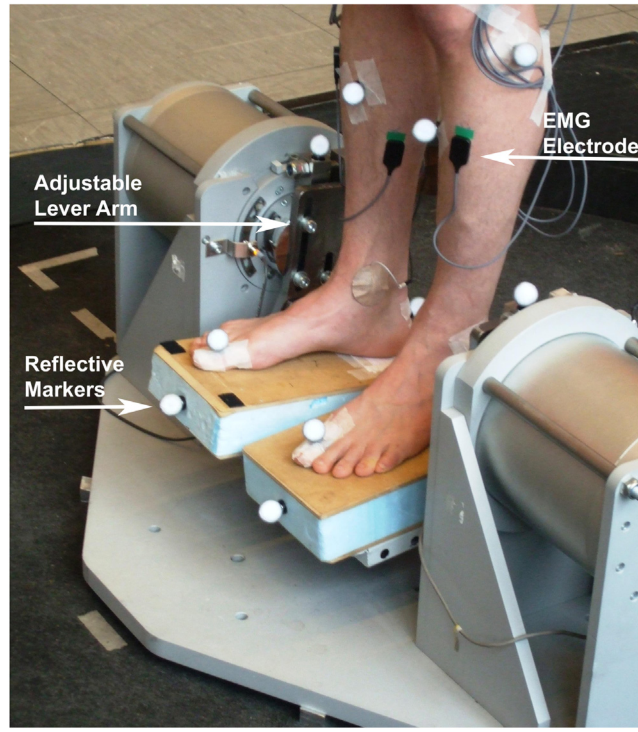

Fig. 1. The Bilateral Ankle Perturbator (BAP). Rapid 40 ms plantar- and dorsiflexions were simultaneously applied to both feet using the BAP. The lever arms can be adjusted to align the subject's ankle joints with the motor axis.

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.