Assessment of a portable device for the quantitative measurement of ankle joint stiffness in spastic individuals

Jakob Lorentzen a,b,c, Michael J. Grey c,d, Svend S. Geertsen c,d, Fin Biering-Sørensen e, Kelly Brunton f, Monica Gorassini f, Jens Bo Nielsen c,d,*

a Department of Physiotherapy, Hvidovre Hospital Kettegård Allé, 2950 Hvidovre, Denmark
b Department of Neurorehabilitation TBI Unit, Copenhagen University Hospital, Glostrup, Denmark
c Department of Exercise and Sport Science, Nørre Alle 51, 2200 Copenhagen N, Denmark
d Department of Neuroscience and Pharmacology, University of Copenhagen, Blegdamsvej 3, 2200 Copenhagen N, Denmark
e Clinic for Spinal Cord Injury, Rigshospitalet, Blegdamsvej 9, 2100 Copenhagen Ø, Faculty of Health Sciences, University of Copenhagen, Denmark
f Department of Biomedical Engineering, University of Alberta, Edmonton, AB, Canada

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HIGHLIGHTS

• The portable device correlated well with measures of joint stiffness obtained by a torque motor in both controls and neurological patients with high intra- and inter-rater reliability.
• The device could easily distinguish between stiff and control ankle joints.
• A portable device can be a useful diagnostic tool to obtain reliable information of stiffness for the ankle joint.

ABSTRACT

Objective: Spasticity is a common complication with neurological diseases and CNS lesions. Instrumented systems to evaluate spasticity often cannot provide an immediate result, thus limiting their clinical usefulness. In this study we investigated the accuracy and reliability of the portable Neurokinetics RA1 Rigidity Analyzer to measure stiffness of the ankle joint in 46 controls, 14 spinal cord injured (SCI) and 23 multiple sclerosis (MS) participants.

Methods: Ankle stiffness measures were made twice by two raters, at speeds above and below the expected stretch reflex threshold. Ankle torque was measured with the portable device and a stationary torque motor. Inter- and intra-rater reliability was assessed with the intra-class correlation coefficient (ICC).

Results: Stiffness measures with the portable and stationary devices were significantly correlated for controls and MS participants (p < 0.01). Intra-rater reliability for the portable device ranged from 0.60–0.89 (SCI) and 0.63–0.67 (control) and inter-rater reliability ranged from 0.70–0.73 (SCI) and 0.61–0.77 (control). Ankle stiffness measures in SCI and MS participants were significantly larger than in controls for both slow (p < 0.05) and fast movements (p < 0.01), with stiffness being larger for fast compared to slow movements in SCI and MS participants (p < 0.05), but not in controls (p = 0.5).

Conclusion: The portable device correlated well with measures obtained by a torque motor in both controls and MS participants, showed high intra- and inter-rater reliability for the SCI participants, and could easily distinguish between stiff and control ankle joints. However, the device, in its current form, may be less accurate during rapid movements when inertia contributes to stiffness and the shape of the air-filled pads did not provide a good interface with the foot.

Significance: This study demonstrates that a portable device can potentially be a useful diagnostic tool to obtain reliable information of stiffness for the ankle joint.

1. Introduction

Spasticity is a common complication to many neurological diseases or lesions in the central nervous system (CNS) (Simpson et al., 2009; Kirshblum, 1999; Mayer, 1997). The broad definition...
of spasticity as “disordered sensorimotor-control, resulting from upper neuron lesion, presenting as intermittent or sustained involuntary activation of muscles” (Pandy et al., 2005) contains many of the different clinical features related to increased muscle activity caused by a lesion in the CNS. The more narrow definition by Lance as “a velocity-dependent increase in tonic stretch reflexes to phasic stretch, in the absence of voluntary activity” (Lance, 1980) describes one specific reflex mechanism that causes increased muscle activation and subsequently, increased stiffness to imposed movements. However, within the conceptual understanding of both definitions there is an aim of making a separation and quantification of the neural and non-neural contributions to stiffness. This is important in order to initiate and maintain adequate treatment. This quantification of spasticity is often based on clinical evaluation systems such as the Ashworth Score (AS) (Ashworth, 1964) or Modified Ashworth Score (MAS) (Bohannon and Smith, 1987). These methods are based on judgement of the resistance of the limb from manually imposed movements. One problem with these methods is that no clinically relevant information can be extracted from these judgements unless the evaluators are able to distinguish between stiffness caused by neuronal activity or by passive elastic properties in the muscle, tendon and joints (passive and active stiffness). This has been demonstrated to be a very difficult task with a questionable reliability (O’Dwyer et al., 1996; Dietz and Sinkjaer, 2007; Galiana et al., 2005; Malhotra et al., 2008; Biering-Sørensen et al., 2006; Lorentzen et al., 2010). The Tardieu scale emphasises the use of different test velocities in accordance with Lance’s definition, but this is problematic because of the difficulty in “judging” different velocities and joint angles in a clinical test situation (Biering-Sørensen et al., 2006; Malhotra et al., 2008; Lorentzen et al., 2010).

Elements such as velocity, measurement of reflex onset and resistance against a passive movement can be controlled very precisely in a laboratory setting and measures such as torque and EMG can provide a reliable, objective and quantifiable separation of passive and active contributions to muscle stiffness (Knutsson and Martensson, 1980; Sinkjaer et al., 1993; Sinkjaer and Magnusson, 1994; Lorentzen et al., 2010). However, these methods are not easily applied in a clinical setting because they are time-consuming, expensive, require space and are not always well tolerated by patients. Therefore, there is a need for a portable device that can provide a quantification of muscle stiffness in a clinical setting with the qualities of the stationary biomechanical devices.

Quantification of joint stiffness by portable hand-held devices has been reported for the ankle in children with cerebral palsy (CP) (Boiteau and Malouin, 1995; Malouin et al., 1989) and in SCI participants (Lamontagne et al., 1998), in the knee in different neurological pathologies (Lebiedowska and Fisk, 2009; Stein et al., 1996) and in the elbow joint in individuals with stroke (Lee et al., 2002, 2004; Chen et al., 2005) and Parkinson’s Disease (Prochazka et al., 1997). For the ankle, there appears to be no direct reliable correlation between the measured stiffness from the hand-held dynamometers and from a quantitative stationary device (Malouin et al., 1989; Lamontagne et al., 1998). Different resistive torques and a large variability of the results obtained with the two methods were found (Malouin et al., 1989; Lamontagne et al., 1998). Also no information about the sensitivity of the portable devices was provided from these studies since only the group mean values were reported.

Test–retest reproducibility of the resistive torque measured with hand-held devices has generally been demonstrated to be high (Boiteau and Malouin, 1995; Malouin et al., 1989; Lamontagne et al., 1998), but inter-rater reliability has only been investigated in a few studies (Malouin et al., 1989; Dvir et al., 1991). Malouin et al. (1989) found a large variability between raters for stiffness measured during slow and fast ankle movements in spastic CP children with CP (ICC 0.62; 0.59), whereas Dvir et al. (1991) found a high inter-rater reliability (ICC > 0.89) when measuring the resistive force of plantar flexors in CP children.

These variable results are one reason why portable devices that measure stiffness have not been adopted in the clinical setting. In addition, most of the hand-held devices consist of either a myometer or a hand-held strain gauge with electrogoniometry as two separate devices that require offline analysis in order to obtain a measure, which is not practical in a clinical setting. Here we investigate the accuracy and reproducibility of measuring ankle joint stiffness in controls, SCI and MS participants using a portable device (Prochazka et al., 1997) that provides both force and length data to immediately obtain stiffness measures. This device was originally developed to test the rigidity of arm muscles in individuals with Parkinson’s disease (Prochazka et al., 1997), but its design also makes it potentially suitable to evaluate ankle stiffness in individuals with spasticity.

2. Methods

The data in this study are based on three sets of experiments in which mechanical impedance was measured in the ankle. Mechanical impedance, or resistance to movement, contains inertial, viscous and elastic components, the later components commonly referred to as viscous and elastic stiffness in biological literature. Therefore, we refer to the amount of resistance applied to the ankle joint over a given distance as ankle stiffness throughout this study. In the first set of experiments we correlated measures of ankle stiffness obtained with the portable device to quantitative biomechanical measurements of ankle stiffness (passive and reflex torque) obtained with an instrumented torque motor in control and MS participants. The second set of experiments was conducted in control and SCI participants to determine the reliability of the stiffness measurements obtained with the portable device on different experimental sessions for the same examiner (intra-rater reliability) and between two examiners in the same experimental session (inter-rater reliability). Finally, in the third set of experiments we tested the ability of the portable device to distinguish stiffness between controls, spastic SCI and MS participants.

Forty-six (age 32 ± 7 years; 16 female) volunteers with no history of neurological disease or CNS trauma participated in the study. Fourteen SCI participants (age 48 ± 12 years; 1 female), and 23 MS participants (age 53.3 ± 12 years; 14 females) were recruited from rehabilitation centres. The specific number and age of the participants for each set of experiments is described below (see details in Table 1).

All subjects gave written and oral consent to the study, which was approved by the local ethics committee (Videnskabsetisk komite for Region Hovedstaden), and performed in accordance with the Declaration of Helsinki.

2.1. Design of study

2.1.1. Test Series 1: Comparison between portable and static device

In each participant, the ankle joint stiffness measured by one rater with the portable device for the fast and slow movements was correlated to the reflex and passive torque obtained from objective biomechanical and electrophysiological parameters. The test with the stationary device was conducted immediately after the test with the portable device in 12 controls aged 32.4 years ± 7.2 (3 females) and in 23 MS participants aged 53.3 years ± 12 (14 females). In order to elicit the reflex and measure the maximal reflex response it was necessary to apply these dorsiflexions at very fast speeds (200 °/s). The different velocities were chosen to determine the threshold for the stretch reflex. The total torque was measured as the averaged maximal torque produced during
the fastest stretch velocity (200/s), whereas the passive torque was the average torque measured during the slowest stretch velocity (8/s) where no stretch reflex response was produced as identified with EMG.

2.1.2. Test Series 2: Reliability

The reliability of the ankle stiffness measures from the portable device was investigated by an inter-rater test and intra-rater test. The inter-rater test in SCI participants was performed on the same day in 9 participants with SCI clinically identified with spasticity based on tendon reflex, Modified Ashworth and Ankle Clonus scores (age 44±13 years; 1 female) (Table 1). Two experienced physiotherapists measured each participant, separated by 20 min between each test, with the order: 1st rater, 2nd rater, 1st rater and then 2nd rater.

The intra-rater tests in controls were performed at the same time of day over two consecutive days in 11 controls (age 30±5 years; 5 female) by the two raters. The inter-rater test for the ankle in 9 controls (age 44±10 years; 2 female) were done on the same day, separated by 20-min between each test by the raters.

Table 1

<table>
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<tr>
<th>Participant No.</th>
<th>Neurological level of SCI</th>
<th>Age (years)</th>
<th>Reflex (Achilles)</th>
<th>Reflex (patella)</th>
<th>MAS (ankle)</th>
<th>MAS (knee Flex.)</th>
<th>Ankle clonus</th>
<th>Year of SCI/MS</th>
<th>Examined side</th>
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2.1.3. Test Series 3: Comparison of stiffness between controls, SCI and MS participants with the portable device

Fourteen participants with SCI (age: 48 ± 12 years; 1 female) having increased ankle stiffness (MAS ≥ 1) and twenty-three participants with MS (age: 53.3 ± 14 female) were enrolled in this study (Table 1). Twelve of the SCI participants (age 51 ± 10 years; 1 female) had MAS scores ≥ 1+ and participated in the comparison of ankle stiffness. The clinical evaluations, as well as the portable stiffness tests for the comparison between participants with SCI and controls, were conducted by the same investigator. We waited 5 min after the clinical examination before initiating the test with the portable device. Fourteen uninjured, age-matched (48 ± 10 years; 4 female) individuals were used as the control group.

Stiffness was measured at two velocities for the ankle (fast: 150.5 ± 75/s; slow 48.1 ± 75/s) in the SCI, MS and control group. A pause of 5 min was held between the fast and slow test sets. For each velocity, stiffness measured at 20 movement cycles was averaged together for each person in the study. The average stiffness (Nm/°) values for all participants were averaged and compared across fast and slow movements and between groups.

2.2. Apparatus

2.2.1. Stiffness test with the portable device

The test apparatus (The Neurokinetics RA1 Rigidity Analyzer, Edmonton, Canada) uses two force sensors (air-filled pads), connected to a pressure-sensitive diaphragm and positioned on the anterior and posterior surfaces of the foot. The rater gripped the sensors as shown in Fig. 1 with equal force applied to each pad. When the user applied a similar grip force to both pads, the pressure on each side of the diaphragm was similar, and the net force signal remained close to zero. However, when the foot was moved, the difference between the forces on each pad was registered by the pressure-sensitive diaphragm. Joint torque was computed by multiplying the force measured at the foot by the moment arm to the point of force application. The device also measured joint displacement and velocity during movements via a gyroscope attached to the dorsal force pad.

The operator imposed sinusoidal movements to the ankle as described below for 50 s. Passive joint stiffness was then estimated from the torque and displacement data by fitting the torque and displacement data to a second-order model of the equation of motion: T = Kx + Bv + C. Where: T is torque; K is stiffness; x is displacement; B is viscosity; v is velocity and c is constant offset of the sensors. The device software estimated stiffness with a standard linear least-squares algorithm to fit the three parameters K, B, and C. In addition to providing an immediate output of stiffness, the data were sent to a PC for storage and off-line analysis. A typical data set is shown in Fig. 2. For further information about the device see Prochazka et al. (1997).

Displacement and velocity measures were calculated off-line by analysis of amplitudes and slopes, respectively, of the movement curves (Fig. 2). Stiffness calculations were made in a window when the oscillations were stable, with the first two movements excluded in all cases.

2.3. Test method

All participants were asked not to take any medication or drink any coffee or alcohol on the day when the test was conducted. The participants were also asked to refrain from any unusual physical exercise on the day prior to and of the test session. The SCI participants were asked not to make any change in their medication on the day prior to and of the test session. It was not possible for the MS participants to refrain from taking medication prior to the experiments.

All tests took place in a quiet room at approximately 20 °C. The participants were positioned supine with their knees stretched and their feet just free from touching the bed during the ankle test (Fig. 1). The force pads were positioned on the dorsal and plantar sides of the foot during the test with the centre of the gyroscopes placed 13 cm distal to the centre of the medial malleolus on the dorsal side of the foot. One hand of the rater was used to stabilize the knee and the other to move the foot. The hand moving the foot was positioned with the fingers and the thumb on each of the two force pads, respectively. The raters were instructed to apply the force exclusively to the force pad during the movements. The movements imposed by the raters were identical to a normal clinical examination of spasticity in the ankle except that the movements were made through the force measuring device (Fig. 1).

The stiffness measurements were made at two velocities with approximately 20 alternating movements per set. Movement velocity was chosen based on a previous observation that the stretch reflex appears in the electromyogram at approximately 75 °/s (Lorentzen et al., 2010). Therefore, the fast movements in the present study were performed faster than 75 °/s and the slow movements slower than 75 °/s, as described below. The operators were instructed to complete each oscillation in approximately 1 s or less for the fast movements (~120 °/s) and 3 s for the slow movements (~20 °/s). Each oscillation was followed by a 1 s pause and each set of 20 movements was followed by a 5 min break.

The average velocity of the ankle dorsiflexion was 150.5 ± 9.1 °/s (fast movements) and 48.1 ± 7.2 °/s (slow movements). The pauses between each movement ranged from 0.79–0.85 s. The average displacement for the ankle movements during the stiffness tests was 36.8 ± 2.5 ° for fast movements and 35.7 ± 4.0 ° for slow movements (p > 0.3). A typical example of the force, displacement and velocity profiles for both the fast and slow ankle movements are shown in Fig. 2.

The participants were asked to relax as much as possible during the examination. The resting position (approximately 110° plantar flexion) for the ankle joint was the start position for the stretches and the end position was the position where no further dorsiflexion was possible (approximately 80° plantar flexion). The displacement size at approximately 30° was chosen to make it possible for the
rater to reach the desired speed of the fast displacements of $\geq 120^\circ/s$.

The raters practiced the test procedure on a control participant prior to the tests. The importance of following the test procedures very accurately was emphasised to the raters prior to each test session.

2.3.1. Quantitative biomechanical measurement

The quantitative biomechanical method to quantify reflexive and passive torque is detailed in Lorentzen et al., 2010. Briefly, a computer-controlled torque motor applied stretches to the ankle plantar flexor muscles at different velocities (8–200 $^\circ/s$, amplitude 6$^\circ$). During the tests, the participants were seated in a chair with their foot strapped to a pedal with the ankle and knee extended to an angle of 100$^\circ$. The participants were asked to relax throughout the test. The contribution of passive and reflex elements to the torque response of the ankle was determined by comparing torque during slow movements (below reflex threshold) to torque during fast movements that were above reflex threshold. Recordings of torque were used to calculate total torque during a fast stretch and soleus EMG was used to ensure the absence of muscle activity for the passive torque measurements. To measure reflex torque, we subtracted the measured torque where no reflex activity was present, as determined with EMG, from the total torque from the fast stretches with maximal reflex response (reflex component of torque = fast (reflex + passive torque) – slow (passive torque)). We have shown previously (Lorentzen et al., 2010) that passive stiffness remains constant for the investigated stretch velocities. The difference between the fast and slow stretches should therefore equate with the reflex mediated torque.

2.4. Statistics

Matlab version R2006b was used for off-line analysis of the quantitative torque measurements (for details see Lorentzen et al., 2010).
et al., 2010) and displacement/velocity measurements from the portable stiffness measurements.

SigmaStat statistical software version 11.0 was used for all statistical analysis except for Intraclass correlation coefficient (ICC) analysis where SPSS (version 17.0) was used.

ICC could in general be called the ratio of two variances (variance = \( \sigma \) where:

\[
\text{ICC} = \frac{\sigma \text{ due to rats} + \sigma \text{ due to raters} + \sigma \text{ residual}}{\sigma \text{ due to rated participants}}
\]

ICC singles measures for absolute agreement (two ways mixed model) were used as measure for variability between raters (inter-rater reliability) and variability between the two tests sessions for each rater (intra-rater reliability). The ICC value represents excellent reliability when >0.75; moderate to good reliability when 0.4–0.75; poor reliability when below 0.4 (Fleiss, 1986).

Bland Altman plots with limits of agreement were used to illustrate systematic differences in measurements between the raters as well as between the repeated tests. The limits of agreement are the range in which the difference in stiffness values between the raters or the repeated tests should lie for 95% of the participants (Bland and Altman, 1999).

Differences in mean values for stiffness were tested by independent sample t-test unless in situations where data were not normally distributed. In those cases the Kruskal–Wallis one way analysis of variance was used.

Altman limits of agreement were also similar between raters and residual measures for the slow movements for the MS participants (0.156 ± 0.05 Nm/s) and slow (0.192 ± 0.05 Nm/s) movements were found (p = 0.9).

3. Results

3.1. Test Series 1: Correlation between portable stiffness measurements and torque measured in the static device

Fig. 3 shows the relationship between various torque measurements from the ankle obtained using the static device and the stiffness measured with the portable device for the fast and slow movements in 12 controls and 23 MS participants. As expected, a significant relationship was found between the total ankle torque measured during fast movements with the static device and the stiffness measured with the portable device in controls \((r^2 = 0.580; p = 0.004, \text{Fig. 3A})\) and MS participants \((r^2 = 0.46; p = 0.001, \text{Fig. 3E})\). The relationship between the ankle stiffness produced by the reflex activity alone (fast torque–slow torque) also correlated significantly with the stiffness measured during the fast movements using the portable device in controls \((r^2 = 0.459; p = 0.02, \text{Fig. 3B})\) and in MS participants \((r^2 = 0.32; p = 0.005, \text{Fig. 3F})\). No significant correlations were found for passive torque and the stiffness measured with the portable device for the slow or fast movements for either controls (Fig. 3C and D) or MS participants (Fig. 3G and H).

The average threshold to evoke a stretch reflex measured with the stationary device was 88.2 ± 33.3 /s for the controls and 24.0 ± 11.5 /s for the MS participants. The average velocity for the ankle movements with the portable device for all participants was 150.5 ± 91.1 /s (fast movements) and 48.1 ± 7.2 /s (slow movements). There were no statistically significant differences in fast and slow rates of movements between the groups. The average torque response to the fastest movements measured with the stationary device was significant larger for the MS participants (10.6 ± 4.7 Nm) compared to the controls (6.2 ± 1.7 Nm; p = 0.001). For the slow movements without presence of any reflex activity no difference between MS (2.7 ± 1.9 Nm) and controls (3.2 ± 1.7 Nm; p = 0.12) was found.

The average response measured with the portable device to the fast movements \((0.192 ± 0.04 \text{ Nm/s})\) was significantly larger than for the slow movements for the MS participants \((0.156 ± 0.05 \text{ Nm/s}; p = 0.01)\). For the controls no significant difference between the average response measured with the fast \((0.196 ± 0.06 \text{ Nm/s})\) and slow \((0.192 ± 0.05 \text{ Nm/s})\) movements were found (p = 0.9).

3.2. Test Series 2: Reliability of the portable device

The reliability of the portable device was assessed with intra-rater and inter-rater measurements in 9 SCI participants and 20 controls (Fig. 4). The intra-rater reliability for the SCI participants, measured as ICC-values, ranged from 0.60–0.89 and the inter-rater reliability ranged from 0.70–0.73 (Table 2).

Differences across assessments for measures of the controls are presented in Fig. 4 as either intra-rater reliability or inter-rater reliability by Bland–Altman plots including limits of agreements. In Table 2, all reliability tests showed ICC values between 0.53–0.98. The inter-rater reliability was similar for both fast and slow movements (0.61; 0.77) to the intra-rater reliability (0.63; 0.67). The potential differences in stiffness indicated by the Bland–Altman limits of agreement were also similar between raters and between the two test sessions for the same rater (Table 2). As seen in Fig. 4, all differences in stiffness values between raters and between tests in the same rater were within the limits of agreements as marked by the horizontal black and grey lines indicating mean ± 1.96SD. The Bland–Altman plots were examined for trends in difference between the raters or between tests in the same rater. This was not the case in any of the comparisons (Fig. 4).

3.3. Test Series 3: Stiffness measures for controls, SCI and MS participants

The difference between the stiffness measured with the portable device for controls (solid symbols), SCI (open symbols) and MS participants (grey symbols) is illustrated in Fig. 5A. The average stiffness measured for the fast ankle movements was significantly larger for the SCI and MS participants compared to the controls (SCI: difference in mean = 0.214 Nm/s; p < 0.001; MS: difference in mean = 0.111 Nm/s; p < 0.001). Compared to controls, stiffness was also larger in SCI and MS participants when measured during slow movements (SCI: difference in mean = 0.082 Nm/s; p = 0.001; MS: difference in mean = 0.083 Nm/s; p < 0.001) (Fig. 5A). A significant difference between the average stiffness for fast and slow ankle movements was seen for the SCI and the MS participants (SCI: fast, mean = 0.295 Nm/s; slow, mean = 0.155 Nm/s; difference in mean = 0.140 Nm/s; p < 0.05; MS: fast, mean = 0.193 Nm/s; slow, mean = 0.156 Nm/s; difference in mean = 0.037 Nm/s; p = 0.013), but not for the controls (fast, mean = 0.081 Nm/s; slow, mean = 0.073 Nm/s; difference in mean = 0.044 Nm/s; p = 0.5; Fig. 5A).

The average stiffness values for the controls and SCI participants are summarised in Table 3.

No significant correlation was found between MAS and the ankle stiffness measured with the portable device for MS participants \((r^2 = 0.05–0.08; p = 0.2–0.3)\) or SCI participants \((r^2 = 0.004–0.2; p = 0.2–0.8)\). There was also no significant correlation between duration since injury and the ankle stiffness for the SCI participants \((r^2 = 0.0675; p = 0.574 \text{ for slow movement and } r^2 = 0.0194; p = 0.322 \text{ for fast movement})\). Finally, there was no correlation between the ankle stiffness and the neurological level in the SCI participants.

4. Discussion

In this study, we used a commercially available portable device with integrated myometer (force recordings) and gyroscope to...
demonstrate that ankle stiffness measures from the portable device were well correlated to those measured with a stationary device. Reproducible measures were obtained for stiffness of the ankle with high intra-rater reliability and high inter-rater reliability for the SCI participants and moderate to good reliability for the controls, with all ICC values within the limits of agreement.

Fig. 3. The relation between ankle stiffness measured with the portable device and the torque measured with the stationary device. The stiffness for the fast movements can be predicted by the total torque and reflex torque in controls (n = 12) (A and B) and MS (n = 23) (E and F). The passive torque (no reflex activity), however, does not predict the stiffness for the slow or fast movements in controls (C and D) and MS participants (G and H).

Moreover, the stiffness measured with the portable device was significantly larger in participants with SCI and MS compared to the controls.

4.1. Comparison of ankle stiffness measurements obtained with the portable and stationary devices

In our previous study (Lorentzen et al., 2010), we found that the average velocity at which dorsiflexion evoked a stretch reflex in the soleus muscle was 75/s for controls and spastic participants. We therefore consequently used fast and slow movements with the portable device which were much faster (150/s) and slower (48/s) than this in order to obtain measurements with and without a reflex contribution. The reflex threshold of 88.2 ± 33.3/s obtained in the 12 controls in this study agrees with our earlier findings, but the threshold of 24.0 ± 11.5/s in the MS participants is lower than expected, and the reflex may therefore have been a contributor to the stiffness measured with the portable device for both the fast and the slow movements.

It is not surprising that we found a significant correlation between the stiffness for the fast movements made with the portable device and the total and reflex torque measured with the stationary device. In these cases, the evoked stretch reflex activity is likely to be a major determinant of the measured stiffness. The significant relationship for the fast movements between the two methods was found despite the fact that different amplitudes of movement were used. Approximately 35° of dorsiflexion was used with the portable device and 6° of dorsiflexion was used with the stationary device with approximately the same start test angle. This suggests that the amplitude of movement is not of major importance for the determination of stiffness measured with fast movements with either method. However, it may be argued that

Table 2

<table>
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<th>Ankle</th>
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<th>ICC value (95%CI)</th>
<th>Mean difference and limits of agreements (1.96 × SD of mean diff)</th>
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<td>0.63 (0.09–0.89)</td>
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<td>0.89 (0.57–0.97)</td>
<td>0.01 (−0.06–0.08)</td>
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<td>0.67 (0.16–0.90)</td>
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<td>0.60 (−0.11–0.84)</td>
<td>0.001 (−0.07–0.08)</td>
</tr>
<tr>
<td>2. rater fast movements</td>
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<td>–</td>
<td>0.89 (0.62–0.98)</td>
<td>−0.01 (−0.07–0.04)</td>
</tr>
<tr>
<td>2. rater slow movements</td>
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<td>–</td>
<td>0.78 (0.33–0.95)</td>
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<tr>
<td>Inter-rater reliability</td>
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<tr>
<td>Fast movements</td>
<td>0.61 (0.03–0.89)</td>
<td>0.02 (−0.04–0.08)</td>
<td>0.70 (0.10–0.30)</td>
<td>0.02 (−0.12–0.17)</td>
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<td>Slow movements</td>
<td>0.77 (0.25–0.94)</td>
<td>0.00 (−0.07–0.06)</td>
<td>0.73 (0.20–0.90)</td>
<td>−0.01 (−0.09–0.07)</td>
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</table>

Fig. 4. Reliability of ankle stiffness measures for the portable device. The Bland–Altman plots including the mean difference ± 1.96 × SD between the raters for controls (n = 9; black circles and black lines) and SCI participants (n = 9; open circles and grey lines). Inter-rater reliability of the stiffness measures for the fast and slow ankle movements are shown in (A and B), respectively. The Intra-rater reliability for the 1st rater for fast and slow ankle movements are shown in (C and D), respectively.

Moreover, the stiffness measured with the portable device was significantly larger in participants with SCI and MS compared to the controls.

4.1. Comparison of ankle stiffness measurements obtained with the portable and stationary devices

In our previous study (Lorentzen et al., 2010), we found that the average velocity at which dorsiflexion evoked a stretch reflex in the soleus muscle was 75/s for controls and spastic participants. We therefore consequently used fast and slow movements with the portable device which were much faster (150/s) and slower (48/s) than this in order to obtain measurements with and without a reflex contribution. The reflex threshold of 88.2 ± 33.3/s obtained in the 12 controls in this study agrees with our earlier findings, but the threshold of 24.0 ± 11.5/s in the MS participants is lower than expected, and the reflex may therefore have been a contributor to the stiffness measured with the portable device for both the fast and the slow movements.

It is not surprising that we found a significant correlation between the stiffness for the fast movements made with the portable device and the total and reflex torque measured with the stationary device. In these cases, the evoked stretch reflex activity is likely to be a major determinant of the measured stiffness. The significant relationship for the fast movements between the two methods was found despite the fact that different amplitudes of movement were used. Approximately 35° of dorsiflexion was used with the portable device and 6° of dorsiflexion was used with the stationary device with approximately the same start test angle. This suggests that the amplitude of movement is not of major importance for the determination of stiffness measured with fast movements with either method. However, it may be argued that
much longer stretches were used with the portable device. It is a drawback of the study that we used a different range of variation in the measurements. However, since the velocities and amplitudes of movement should have been used with both the stationary and portable devices. However, since the increase in ankle torque is reportedly linear with manual perturbations over a large range of motion (−35 to 5° of dorsiflexion at >300 °/s) (Lamontagne et al., 1998), there is no reason to believe that a qualitatively different result would have been found for other joint angles and movement amplitudes. Regarding the velocities, especially the difference in the slow test velocities between the portable (48 °/s) and stationary (8 °/s) devices, the large discrepancy in movement velocities may partly explain the poor correlation found for the ‘passive torque’ (Fig. 3).

In summary, no other study has, to our knowledge, reported a direct correlation between portable and stationary device measurements of stiffness to ankle movements in controls and neurological participants.

4.2. Reproducibility

The intra-rater reliability for the measurements with the portable device, as indicated by the ICC was 0.78–0.89 for SCI participants and 0.63–0.67 for controls. Similar, intra-rater reliability results with the use of hand-held dynamometers were found for the ankle in studies in children with CP (Boiteau and Malouin, 1995) and adults with SCI (Lamontagne et al., 1998). The difference in stiffness measures between the raters in this study (inter-rater reliability) for the ankle was demonstrated by an ICC of 0.70–0.73 (SCI) and 0.61–0.77 (controls). Larger inter-rater reliability for ankle measurements than in our study was found in one previous study (ICC: 0.89; Dvir et al., 1991), whereas another study found lower reliability compared to our study (ICC: 0.59–0.62; Malouin et al., 1989). In the study by Dvir et al. (1991), each limb movement included a hold period of 5 s at the end point of movement. This is a potential problem since the stiffness is naturally quite large at the end point of movement of the limb, but does not reflect the actual muscular stiffness during the movement itself. This may also explain the large reliability in that study, since the measurements were likely to be heavily influenced by the amount of force that the rater applied at the end of the movement. In our study the raters, similar to the study by Malouin et al. (1989), were just short of the end point position of the ankle (which was approximately 0° for most participants), and from this position, moved back to the initial position (approximately 35° plantar flexion). This method was chosen to ensure a range that was large enough to make it possible for the rater to reach a velocity for the dorsal flexions that was fast enough to elicit the stretch reflex, while at the same time avoiding application of too large a force at the end point of movement where the stiffness is naturally expected to be large.

Finally, the fact that inertia is not added as a factor in the calculation of stiffness used here may possibly subtract some of the variation in the measurements. However, since the velocities and

![Fig. 5. Mean stiffness for controls, SCI, and MS participants. (A) Mean stiffness in controls (n = 14) (c fast, c slow), SCI (n = 12) (s fast, s slow) and MS participants (n = 23) (MS fast, MS slow). * indicates significant difference (p < 0.05) between mean value for the SCI, MS patients and controls. ‡ indicates significant difference between the mean values for fast and slow movements within the control, SCI or MS group. (B) Relationship (including regression lines) between the stiffness measured at the first and second round for rater 1 is illustrated for fast (black circles and lines) and slow (grey triangles and lines) movements. (C) Relationship between stiffness measured by the 1st rater and the 2nd rater for the fast and slow movements.]
amplitudes of the movements were very similar across participants, raters and different testing days in this study, this factor should not have a large influence on the difference between the measurements – rather, it would more likely have an influence on the scaling of the absolute values.

4.3. SCI and MS participants vs. controls

In the controls, only a very minor difference in ankle stiffness was found between the fast and slow movements. This result is a little surprising since the device showed ability to distinguish different levels of ankle stiffness in both spastic and controls. One explanation could be that the hand-held device may not be able to identify relatively small velocity dependent contribution to the stiffness in controls. For the SCI and MS participants, significantly larger stiffness was found for the fast movements compared to the slow movements of the ankle. Due to the position of the participants during the test, the inertia was not influencing the stiffness values for the ankle, and since the velocity for the fast movements was above the average threshold velocity for the stretch reflex, these results suggest a significant reflex contribution to the stiffness measured with the hand-held device during fast ankle movements in the neurological participants. This is supported by the EMG measurements that documented a larger reflex response for the MS participants compared to the healthy controls. This result supports previous findings showing increased reflex activity for fast plantar flexions with stationary devices (Lorentzen et al., 2010; Mirbagheri et al., 2001) and hand-held dynamometers (Lamontagne et al., 1998; Boiteau and Malouin, 1995) and fits nicely with the definition of spasticity by Lance with “velocity-dependent increase in tonic stretch reflexes to phasic stretch, in the absence of voluntary activity” (Lance, 1980).

The ankle stiffness for the slow movements in SCI and MS participants was also significantly larger than the stiffness for the controls. This result is probably caused by an increase in the passive stiffness parameters of the muscles and tendons, which has been found in other studies measured with stationary devices (Mirbagheri et al., 2001; Dietz et al., 1981, 1991; Dietz and Berger, 1983; Lehmann et al., 1989; Toft et al., 1993; Sinkjaer et al., 1993; Sinkjaer and Magnussen, 1994). Another possibility could be that spastic participants had lowered threshold for stretch reflex as found for MS participants (Mirbagheri et al., 2001; Sinkjaer et al., 1993).

4.4. Methodological considerations

The intra-rater and inter-rater variability of stiffness in the control group (lower ICC values) measured with the portable device was considerably larger than the variability measured in the SCI group. We believe that the primary reason for the lower ICC in controls is because the way we performed the stiffness measures changed slightly over time. As part of the study, we gradually became better at avoiding the endpoint stiffness so that the later measurements (e.g., control measurements that were compared to the SCI participants in Fig. 5) may have given smaller values than earlier measurements (control inter-rater and intra-rater measurements (Fig. 4). We tested this in control participants by comparing the earlier inter-rater and intra-rater measurements with the later measurements, and we found significantly larger stiffness values for the earlier measurements ($p < 0.05$). This underscores the necessity for a very structured testing protocol and several testing sessions prior to the measurements.

The cuff that was positioned on the ankle during the measurement was originally designed to fit the wrist, and the raters generally found it difficult to find a stable position on the foot. This could cause small movements of the cuff and gyroscope that were unrelated to the movement of the limb, which could have had an influence on the measured stiffness. Even though the distance between the rotating point for the joint and the position of the cuff was controlled very carefully, small variations between the positioning of the cuff by the raters was still unavoidable and may be part of the reason for the variability observed in the test. To test the speculation that a more stable fit could lower the variability between raters would require a re-development of the system. Until then, a very exact procedure and several practise sessions for each user of the device are considered important prior to testing.

Electrophysiological information in combination with the existing stiffness measurements would improve the ability of the device to make distinctions between reflex stiffness and passive stiffness. EMG measurements in combination with the stiffness measurements should therefore be considered for future hand-held devices.

The calculation of stiffness used in this study did not take the inertial factor into account. According to Procházka et al., 1997 this factor will not have a significant impact on the stiffness measure for movements made with slow velocities (<1 Hz), which were the case for the slow velocities used in this study. However, the fast movements in this study could possibly be influenced by the inertial factor resulting in lower stiffness. This may possibly have caused an underestimation of stiffness in all fast movements where inertia due to the position of the test participant could have been influenced by the weight of the foot when moved against gravity. Special attention to the test procedure regarding velocity and position should be made in future studies.

4.5. Measurement of stiffness by portable device in routine clinical examination?

The significant correlation between ankle stiffness at velocities above reflex threshold measured with the portable and the stationary device suggests that it is possible to obtain measures with the portable device, which are quite comparable to what may be obtained with much more complex and elaborate stationary devices. In contrast, no significant correlation between the clinical measure of spasticity given by the Modified Ashworth scale and the stiffness measures obtained with the portable device (or for that matter the stationary device; cf. Lorentzen et al., 2010) was found for any of the two participating groups. This illustrates the difficulty in obtaining valid measures of muscle stiffness clinically without the}

| Table 3 |
| The mean stiffness values including SD for the spastic SCI, MS and control group for the ankle movements. |

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the help of objective biomechanical techniques. In line with this, the Ashworth scale has been found to have low inter- and intra-rater reliability and to suffer from unequally distributed rating (see Biering-Sørensen et al., 2006 for a review). Thus, there is a definite need to find more optimal evaluation techniques such as the hand-held device that we have tested here.

The relatively high intra-rater reliability (ICC: 0.60–0.89, SCI: 0.63–0.67, controls), inter-rater reliability (ICC: 0.70–0.73, SCI: 0.61–0.77, controls) and its ability to distinguish increased muscle tone in neurological participants also suggests that the hand-held device may prove useful for consecutive measures of muscle stiffness and spasticity by the same person. In regards to increasing the reliability of the device, we would like, in particular, to emphasize the need of developing a device which is less sensitive to the exact placement of the hand, the size of the hand and the way that the actual movements are made. Also, there is a need for addition of the inertia in the calculation of stiffness, which would make great improvements to the measurement, especially for patients that need to be positioned where the limb must be moved against gravity. The rather rigorous evaluation regime, which was developed for the present study, is unlikely to be implemented in the clinic. A more ‘user-friendly’ device is therefore clearly necessary. Moreover, it would also be an advantage if future devices would include identifying movement velocities at which reflexes are evoked in order to distinguish passive and active contributions to the measured stiffness.

Participants with SCI and MS showed no significant differences in stiffness for the fast movements. It would also be surprising if this had been the case, since the mechanical manifestation of increased reflex excitability must be expected to be the same in these two populations. Similar neuronal mechanisms have also been suggested to be involved in the development of spasticity in SCI and SCI participants. In both populations, reduced reciprocal inhibition, presynaptic inhibition and post-activation depression have been reported (see Nielsen et al., 2007 for a review). Moreover, reduced post-activation depression has been found in these two populations to be correlated to increased muscle stiffness that was measured biomechanically (Grey et al., 2007), which is consistent with the larger resistance documented with the hand-held device in the present study.

5. Conclusion

We have demonstrated that a portable device, designed specifically to measure elbow rigidity, can be used to measure ankle stiffness. Measures of stiffness at high velocity correlated well with similar stiffness measures obtained by a torque motor, and the device can also distinguish between stiffness measured at low (presumably reflecting passive stiffness) and high (presumably reflecting active stiffness) movement velocity. The portable device has high intra- and inter-rater reliability when used with the ankle and is sufficiently sensitive to distinguish neurological participants from controls. However, in its present form, the device does not account for the inertial component of impedance and the shape of the air-filled pads was designed to couple well with the wrist. This coupling is less ideal for the foot, making the device somewhat difficult to use.

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References


