A Clinical Measurement to Quantify Spasticity in Children with Cerebral Palsy by Integration of Multidimensional Signals

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1. Introduction

Spasticity is a major cause of secondary problems such as muscle contractures and bony deformities in children with cerebral palsy (CP). However, its assessment remains a controversial topic in clinical settings [1,2]. The most accepted definitions of spasticity refer to a velocity-dependent increase in stretch-reflexes [3], causing an increase in resistance during an externally applied passive stretch [4].

Spasticity is generally assessed by passively moving a joint to grade muscular resistance, using e.g. the Modified Ashworth Scale (MAS [5]) or the Modified Tardieu Scale (MTS [6]). These tests have been criticized for incorrectly measuring spasticity due to their inability to differentiate neural from non-neural components [7] and for their poor reliability when assessing lower limb spasticity [1,8]. Objective, quantitative and robust measurements are thus crucial for the accurate evaluation of spasticity and treatment efficacy [2].

Several instrumented approaches have been suggested which can be categorized as biomechanical, neurophysiological and integrated. Biomechanical approaches capture the behavior of muscles by measuring joint position, angular velocity and torque (reactive-resistance) during well-defined movements. Isokinetic devices, displace a limb at a controlled velocity and have been used to measure limb resistance to passive movement [9,10]. However, these are often bulky and difficult to apply for children. Smaller, manually-controlled dynamometers have similar reproducibility and are clinically more feasible [11]. Neurophysiological approaches investigate the muscles’ electrical activity in reaction to e.g. passive and active movements [12,13], generally using surface electromyography (sEMG). These are considered more valid than the clinical scales and sEMG has been used to identify different spasticity patterns [14,15]. However, when used in isolation, the electrophysiological approaches provide no information about reactive-resistance.
Although some biomechanical approaches additionally use sEMG to determine whether reflex activity was present during the measurements [9,13], few studies interpreted muscle activity together with resistance and velocity measurements. Such integrated approach considers both neurophysiological and biomechanical methods [1,2]. So far, hardly any measurement that fully integrates multidimensional signals is clinically feasible and few have been assessed for reliability [16,17].

Therefore, the first aim of this study was to create an instrumented clinical spasticity measurement and determine a set of quantitative spasticity-sensitive parameters based on multidimensional signals. Second, the clinimetric properties of the measurement, i.e. reliability, discriminative validity and the relation with clinical spasticity scales were assessed. The current study reports on spasticity measurements in the medial hamstrings (MEH) and gastrocnemius (GAS), as both muscles are often treated in children with spastic CP.

2. Method

2.1. Participants

Children with spastic CP were recruited from the clinical motion analysis laboratory (University Hospital xxx) and included if they met following criteria: (1) diagnosis of spastic CP; (2) 5–18 years; (3) ability to understand and perform the test procedure. Children were excluded if they had received botulinum toxin type-A injections within 6 months prior to the assessment; intrathecal baclofen; lower limb orthopedic surgery; had bony deformities/contractures hindering neutral alignment; severe weakness and lack of control of the tested muscle (i.e. manual muscle tests < 2+ [6]). Minimal strength production was required because a representative voluntary contraction was used as an individual reference to evaluate sEMG signals during the spasticity assessment. Ethical approval was granted by the
University Hospitals’ Ethics Committee. All children with CP received physical therapy and day and night orthotic management as routine treatment. Age-matched children of coworkers were recruited as typically-developing (TD) controls.

2.2. Measurement protocol

All children were tested by the same trained assessor. They underwent a full clinical lower limb assessment on the day of testing, including passive range of motion (ROM), spasticity (MAS [5] and MTS [6]), muscle strength and selectivity [6]. Body-weight, height and leg-lengths were recorded.

In children with unilateral CP, the affected side was tested. In children with bilateral involvement, the most affected side (highest average MAS-score for GAS and MEH or highest MTS in case of symmetrical MAS-scores) was evaluated. In TD children one side was randomly chosen for evaluation. Participants in the reliability study were tested twice within 4 weeks, during which they received no treatment except the usual physical therapy.

During the entire instrumented measurement procedure sEMG data was collected from the GAS and MEH as well as from antagonist muscles - tibialis anterior and rectus femoris, respectively - using Zerowire (Cometa, Milan, IT) at a sample rate of 2000Hz. Antagonist activation indicated other tone problems (e.g. dystonia) or active assistance of the child during passive stretches. Circular Ag/AgCl electrodes (diameter of 2cm) were placed on the muscle bellies, with an inter-electrode distance of 2cm (SENIAM guidelines [18]). sEMG recordings were collected during rest periods between test trials to define baseline noise. Three repetitions of agonist and antagonist MVIC were carried out with the subject supine.

To define joint position, angular velocity and acceleration, two inertial measurement units (IMUs: Analog Devices, ADIS16354) were used to track the movement of the distal limb segment with respect to the proximal. To compute the anatomical joint angles from IMU
measurements, calibration trials with predefined motions in the sagittal plane (Supplementary material 1) were performed prior to the passive stretch trials. Torque was measured using a six degrees-of-freedom (DoF) force-sensor load-cell (ATI mini45: Industrial Automation). The load-cell was attached to the segment using a light-weight foot and shank orthosis for the GAS and the MEH, respectively (Figure 1). Motion and torque were sampled at 200Hz.

Segment lengths (lower leg and foot length) and the perpendicular distances (moment arms) between the joint axes and the z-axis of the load-cell (Figure 1) were measured by the assessor. During the passive stretch trials, muscles were tested by manually moving the joints from a predefined starting position (Figure 1) through the full ROM, at low velocity (LV) during 5 seconds, at medium velocity (not included in current data analysis) and finally at high velocity (HV), performed as fast as possible. Each stretch trial was repeated four times with an interval of 7 seconds rest between repetitions.

2.3. Data analysis

Data analysis was carried out using custom-made software (MATLAB 7.6.0 R2008a: MathWorks). Raw sEMG signals were filtered with a 6th order zero-phase Butterworth bandpass filter from 20-500Hz. The root mean square envelope of the sEMG (RMS-EMG) was extracted by applying a low-pass 30Hz 6th order zero-phase Butterworth filter on the squared signal.

To estimate joint position, angular velocity and acceleration, a Kalman smoother [19] was applied on the IMU-data. Using the segment lengths and moment arms, the net internal joint torque was calculated from the external forces, the exerted moments, and the forces caused by gravity and inertia [20] (Figure 1B and Supplementary material 2).

2.3.1. Outcome parameters
Performance-related parameters were used to evaluate the quality of the performance of the measurements. These included ROM and maximum angular velocity during the passive stretch ($V_{\text{MAX}}$), and peak RMS-EMG during MVIC.

Spasticity-related parameters highlighted velocity-dependent components and were based on RMS-EMG and torque. Average RMS-EMG was calculated as the area underneath the RMS-EMG-time curve starting 200ms prior to $V_{\text{MAX}}$ up to 90% of the full ROM of the stretch. This was expressed as a percentage of the peak RMS-EMG value of three MVIC repetitions. EMG-onset was defined as the time of the first muscle activity according to the method of Staude [21]. Torque was analyzed at $V_{\text{MAX}}$. Work was calculated as the area underneath the torque-position graph starting at the time of $V_{\text{MAX}}$ up to 90% of the full ROM.

To investigate how RMS-EMG, torque at $V_{\text{MAX}}$, and work varied with stretch velocity, the absolute change of the parameters between the average of the repetitions at HV and the average of the repetitions at LV was calculated. These spasticity-related parameters are referred to as ‘change’ parameters: ‘EMG-change’ (%), ‘Torque-change’ (Nm), and ‘Work-change’ (J).

2.3.2. Data reduction

All data were visually scanned for consistency and quality of performance. Repetitions were discarded in case of poor sEMG signal quality (low signal-to-noise ratio or clear artifacts), or evidence of antagonist activation.

For the reliability study, three repetitions were compared for within-session analysis. These were subsequently averaged per session for analysis of between-session reliability. Change parameters were only investigated for between-session reliability as these were based on data that was already averaged. For the validity study, only averaged data was used.
2.3.2. Statistical analysis

Statistical analysis was performed using MATLAB 7.6.0 R2008a (MathWorks), and Statistica 10 (StatSoft).

Within- and between-session reliability was assessed at LV and HV using intraclass correlation coefficients (ICC$_{1,1}$) with 95% confidence intervals and the standard error of measurement (SEM). The SEM was calculated from the square root of the mean square error from one-way ANOVA [22]. ICC-values $\geq 0.80$ indicated high; $\geq 0.60$ moderately high; and $\geq 0.40$ moderate reliability [23]. Bland-Altman plots were created to determine systematic bias. For the validity study, medians and interquartile ranges (IQR) of ROM, $V_{\text{MAX}}$, EMG-change, Torque-change, and Work-change were calculated for four groups: (1) children with CP; (2) TD children; (3) children with CP with low spasticity (MAS 1 or 1+); and (4) children with CP with high spasticity (MAS 2 or 3). Mann Whitney-U (MWU) tests were used to compare group 1 vs. 2, and group 3 vs. 4. Subsequently, the correlations of spasticity-related parameters with MAS-scores and MTS-scores were explored using spearman rank correlations. Correlations $>0.8$ were considered very good; 0.61–0.80 good; 0.41–0.60 moderate; 0.21–0.40 fair; and $<0.20$ poor [24]. Significance for all analyses was set at $p < 0.05$.

3. Results

Twenty-eight children with CP and 10 TD children participated in the study (Table 1). Six of the children with CP were retested for the reliability study, with an average interval of $13\pm8$ days. For the GAS, 12 children had low MAS-scores (three MAS 1; nine MAS 1+) and 16 had high MAS-scores (12 MAS 2; four MAS 3). For the MEH, 12 children had low MAS-scores (five MAS 1; seven MAS 1+) and 16 had high MAS-scores (13 MAS 2; three MAS 3). No differences were found between the groups for age, weight, or gender.
3.1. Reliability

Parameters of the GAS and MEH had moderately high reliability with lower within-
than between-session SEM values for all parameters, except \( V_{\text{MAX}} \) of the GAS at LV (Table 
2). Bland-Altman plots showed no evidence of systematic bias (Supplementary material 3). 
The average between-session difference for the GAS was 0.57% (EMG-change), 0.09Nm 
(Torque-change) and 0.06J (Work-change); and -0.81%, 0.31Nm, and -0.92J for the MEH, 
respectively.

3.2. Validity

Generally, performance-related parameters were similar and spasticity-related 
parameters different between the CP and TD groups (Table 3). In the CP group at LV, EMG-
onset was detected in three of 28 GAS muscles and in 14 of 28 MEH muscles. At HV, EMG-
onset was detected in 24 of 28 GAS muscles and in 27 of 28 MEH muscles.

In children with CP, moderate correlations were found for the GAS between EMG-
change, Torque-change, Work-change and the individual MAS-scores \( (r=0.463, r=0.474, 
r=0.492 \) respectively) and moderate to good correlations for the MEH \( (r=0.638, r=0.444, 
r=0.594 \) respectively). The MTS of the GAS correlated poorly with EMG-change \( (r=0.098) 
\) and Work-change \( (r=0.116) \) and fairly with Torque-change \( (r=0.219) \). For MEH, MTS had a 
very good correlation to EMG-change \( (r=-0.809) \), a moderate correlation to Work-change 
\( (r=-0.586) \) and a fair correlation to Torque-change \( (r=-0.275) \).

4. Discussion

This study proposed a clinical measurement that integrates multidimensional signals to 
quantify spasticity in children with spastic CP. The MAS and MTS were only moderately
correlated to spasticity-parameters showing that the clinical scales do not sufficiently detect or investigate spasticity. Quantitative parameters however, extracted from torque and sEMG signals, explored around \( V_{\text{MAX}} \), and compared between velocity conditions, were found to be sensitive to measure spasticity. The proposed method could thus potentially be used to reliably and thoroughly classify muscles and patients, which ensures an adequate evaluation of treatment efficacy. Moreover, the tool provides crucial scientific information that will stimulate the progress of research on the nature of spasticity and its assessment.

4.1. Reliability

While previous studies have shown good reliability of sEMG- and torque-related parameters during manually-applied, passive stretches in healthy subjects and post-stroke adults [16,17], this is the first study to investigate similar parameters in children with spastic CP. Although the current method is based on manual stretching, unlike the MAS, it additionally provides quantitative information on the performance and repeatability of the stretch. Given the difficulties associated with standardizing manual stretches, such information is vital to accurately interpret spasticity. Rabita et al. (2005) showed that isokinetic devices cannot simulate the transient acceleration applied by manual stretching and as such are not suited to elicit stretch reflexes [25]. Moreover, similar torque parameter SEM-values as those in the current study have been reported for stretches of the biceps brachii using a motor-driven dynamometer in stroke patients [26]. Hence, a similar level of accuracy can be achieved with manual stretching.

4.2. Validity

The proposed measurement distinguished spastic from typical muscles, thereby supporting the methods’ discriminative validity. While quantitative spasticity assessments
have been previously introduced, few manually-driven, integrated applications for the lower limbs exist [7,15,27]. In the current study, EMG-change was over 100 times higher in children with CP compared to TD children and Torque-change in the MEH twice that of TD children. These values are higher than those previously reported [15]. The difference may be attributed to the fact that sEMG-parameters in the current study were analyzed in a zone specifically highlighting velocity-dependency, rather than evaluating sEMG over the entire ROM [7,15].

For both muscles, spasticity-related parameters were only moderately correlated to the MAS and most parameters were poorly correlated to the MTS. Previous studies have shown that the MAS is associated with resistance measured during passive stretch, but not with RMS-EMG [7,17,27]. Comparing CP children with high MAS-scores to those with low MAS-scores for the GAS showed different Work-change parameters but similar EMG-change parameters. The MAS is thus less likely to be associated with electrophysiological parameters and therefore cannot be considered a valid measure of spasticity [3]. The discrepancy between clinical scales and objective measures was also highlighted by the lack of EMG-onset in some muscles during stretches at HV, despite the clinical diagnosis of spasticity in all patients. This again indicates the importance of measuring the electrophysiological response of muscles.

4.3 Study limitations

Although the instrumented spasticity assessment was found reliable and sensitive, some study limitations need to be considered. Firstly, the number of subjects was small, especially for the reliability study. Secondly, the MVIC may be difficult to collect in children with CP [28]. However, normalization allows for more accurate intra- and inter-subject comparisons and normalizing to the MVIC additionally allows interpretation of spasticity relative to muscle strength. Thirdly, stretch velocity was significantly different between
groups, which may have impacted on results. However, although TD children and children with low MAS-scores were stretched at higher velocities, their spasticity-related parameters remained lower than those in children with high MAS-scores. Furthermore, it was assumed that calibration and stretch trials were performed in the sagittal plane and that only one joint was moved. Out-of-plane-movements, or movement of the proximal joint, caused small errors (see Supplementary material 1). Therefore, subjects lacking neutral joint-alignment were excluded and out-of-plane movements were minimized by means of standardized reporting on the performance of each trial. Lastly, inertial influences on torque were estimated with anthropometric approximations whereby the foot and lower leg were considered one segment [20]. The error introduced by assuming the ankle fixed during knee movements was assumed to have only a limited effect on the resulting knee-joint torque (Supplementary material 2).

4.4. Clinical relevance and future developments

The current approach yields new opportunities to improve the standards of clinical practice as well as research by providing objective, continuous data. This allows clinicians to interpret the effects of spasticity treatment in a more detailed and sensitive way, compared to the ordinal and subjective rating provided by the MAS. Furthermore, the integrated method relates more closely to the definition of spasticity and could potentially enable the clinician to classify muscles and/or patients according to both severity and varying spasticity patterns.

EMG-onset was detected at LV in 50% of the spastic MEH muscles and in 11% of the spastic GAS muscles, although at very low intensity. Similar activation at LV in spastic muscles has been previously reported [8,15]. These results suggest that spastic muscles possess a degree of position dependency [14,15]. Identifying those muscles in which position has a higher impact than velocity can be clinically relevant and help refine treatment.
Finally, while many spasticity treatments aim to improve gait, weak correlations between MAS-scores and parameters of 3D gait analysis have been reported [29]. It is believed that quantitative spasticity-related parameters will help clarify the contribution of spasticity to certain gait deviations. This knowledge will assist in classifying spastic muscles and will undoubtedly facilitate treatment planning and outcome evaluation in children with CP.

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Conflict of interest

There were no conflicts of interest.

References


### Table 1. Subject characteristics

<table>
<thead>
<tr>
<th>Median (IQR)</th>
<th>Average age (years (std))</th>
<th>Average mass Kg (std)</th>
<th>Gender M/F</th>
<th>GMFCS level</th>
<th>Av. pop. angle (std)</th>
<th>Av. max. dorsiflex. angle (std)</th>
<th>Diagnosis unilateral/bilateral CP</th>
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<tr>
<td>Reliability study CP (n=6)</td>
<td>11.20 (4.8533.7 (20.3))</td>
<td>3 M, 3 F 1I, 4II, 1III</td>
<td>-49.2 (11.1)</td>
<td>4.2 (7.4)</td>
<td>4 unilateral (3L, 1R hemiplegia), 2 bilateral (2 diplegia)</td>
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<td>Validity study TD (n=10)</td>
<td>10.75 (6.4130.5 (23.2))</td>
<td>4 M, 6F NA</td>
<td>NA</td>
<td>NA</td>
<td>NA</td>
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<td>CP (n=28)</td>
<td>10.45 (4.8928.3 (17.6))</td>
<td>8 M, 20 10I, 12II, 5III, 1IV</td>
<td>-62 (14.7)</td>
<td>7.4 (7.2)</td>
<td>8 unilateral (5L, 3R hemiplegia), 20 bilateral (19 diplegia, 1 quadriplegia)</td>
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<td>GAS CP MAS 1,1+ (n=12)</td>
<td>10.17 (4.8827.7 (15.6))</td>
<td>3 M, 9 F 6I, 4II, 1III, 1IV</td>
<td>-56.7 (15.1)</td>
<td>7.5 (6.6)</td>
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<td>10.36 (6.6531.1 (16.1))</td>
<td>5 M, 11 F</td>
<td>-66.9 (13.0)</td>
<td>7.3 (8.1)</td>
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<td>10.36 (5.1227.7 (15.2))</td>
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CP: cerebral palsy; TD: typically developing; GAS: Gastrocnemius; MEH: Medial Hamstrings; MAS: Modified Ashworth Score; NA: not applicable; M: Male, F: Female; L: left-sided; R: right-sided; GMFCS: Gross Motor Function Classification Scale; Av. pop. angle: Average popliteal angle measured using a goniometer with the subject supine and the contralateral leg in full knee extension; Av. max. dorsiflex. angle: Average maximum ankle dorsiflexion angle measured using a goniometer with the subject supine and the knee in full extension.

### Table 2. Means and standard deviations (SD) of parameters at low (LV) and high (HV) velocity stretches of the gastrocnemius (GAS) and medial hamstrings (MEH) in both sessions (test, retest) in children with cerebral palsy. Intraclass correlation coefficients (ICC) and standard error of measure (SEM) for between- and within-session (ICC<sub>B</sub>/ICC<sub>W</sub>, SEM<sub>B</sub>/SEM<sub>W</sub>) reliability.
ROM: range of motion; MVIC: maximum voluntary isometric contraction; \( V_{\text{MAX}} \): maximum angular velocity; EMG change: change in average normalized RMS-EMG between LV and HV; Torque change: change in torque at \( V_{\text{MAX}} \) between LV and HV; Work change: change in work between LV and HV; NA: not applicable.

Table 3. Median and inter-quartile ranges (IQR) for performance- and spasticity-related parameters for gastrocnemius (GAS) and medial hamstrings (MEH) in children with cerebral palsy (CP) and typically developing (TD) children.
LV: low velocity stretches; HV: high velocity stretches; ROM: range of motion; V_{MAX}: maximum angular velocity; MVIC: maximum voluntary isometric contraction; EMG change: change in average normalized RMS-EMG between LV and HV; Torque change: change in torque at V_{MAX} between LV and HV; Work change: change in work between LV and HV. * p<0.05.

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* p<0.05.
FIGURES

Figure 1A. Test starting positions and direction of stretch (white arrows) for the medial hamstrings (MEH) -ensuring that the upper leg was kept still at 90° hip flexion; and the gastrocnemius (GAS) -with a predefined knee angle measured during the calibration trial (see Supplementary material 1). Muscle activity was measured with surface electromyography; joint-angle characteristics with inertial measurement units; and torque using a force-sensor (ATI) attached to either a shank orthosis on the posterior aspect of the lower leg (MEH) or a foot orthosis (GAS). B. Free body diagram of the medial hamstrings and gastrocnemius tests. The masses of the orthoses are considered negligible. $d_y$ and $d_z$ correspond to the distances measured by the therapist. The shank adapter is mounted such that $F_z$ works perpendicular to the main axis of the shank. The mass parameters (mass, inertia and center of gravity) of the shank contain both shank and foot and are determined by an anthropometric model [20]. The body segment is moved with an angular acceleration of $\alpha_{hor}$. This is not the acceleration of the
joint angle, but the acceleration of the angle with the horizontal. The moments for the joints are given by:

\[ M_{\text{knee}} = -F_z + F_y D_y + M_x - mg \cos(\alpha_{\text{hor}}) r_{\text{prox}} \ - l_{\text{axial}} \ \ddot{\alpha}_{\text{hor}} \]  
\[ M_{\text{ankle}} = -F_z d_z - F_y d_y - M_x \]  

(2)

(\text{with } l_{\text{axial}} = m k_{\text{prox}}^2)

Where \( M_{\text{joint}} \) is the net internal moment acting at the joint and is composed of different components: the perpendicular force \( (F_z) \), the non-perpendicular force components \( (F_y) \), the torque exerted on the handle \( (M_x) \), gravity \( (m.g) \) and inertia \( (I_{\text{axial}}) \). \( r_{\text{prox}} \) is the distance of the center of gravity to the center of the knee joint. The moment of inertia of the shank is given by its mass \( (m) \) and the radius of gyration \( (k) \) with respect to the proximal joint \( K_{\text{prox}} \). For the knee joint moment computation, the influence of the ankle movement on the knee torque was considered small and therefore not integrated into the computation (see Supplementary material 2). For the ankle joint moment computation, the moment of inertia for both foot and foot-piece are neglected, because the mass of both is relatively small and the radius of gyration is small. It is expected that \( F_y \) and \( M_x \) have the largest influence.
Figure 2. Example of position-time (A, E), velocity-time (B, F), RMS-EMG-time (C, G) and torque-time (D, H) graphs during low (continuous line) and high (dashed line) velocity stretches of the gastrocnemius (A-D) and medial hamstrings (E-H) of a child with spastic cerebral palsy. Zero seconds was expressed as the time that maximum velocity occurred (for evaluating the effect of maximum velocity).
SUPPLEMENTARY MATERIAL

Supplementary material 1

Calibration trials of the inertial measurement units (IMUs) were carried out prior to the passive stretch trials. To the authors’ knowledge, no references that apply this exact method of calibration are available. Two approaches to compute the anatomical joint angles from IMU measurements are possible. The first method requires placement of the IMUs on predefined bony landmarks and reconstructing the model after measurement. The second method places the IMUs arbitrarily and performs calibration procedures. The second method was applied in the current study because firstly, it was deemed more accurate and secondly, it was more convenient and less time consuming.

It was assumed that calibration and stretch trials were performed in the sagittal plane and that only one joint was moved. A pilot study, comparing the proposed measurement method with Vicon measurements (Vicon, Oxford Metrics Group, UK) reported that extreme movement out-of-the plane of calibration ( >15°) or movement of the proximal-joint, caused a 5-10° error. Smaller out-of-plane movements (<10°) caused only 1-5° error.

Calibration trials

1. Ankle sensor calibration trial

Sensor placement
Two IMU sensors were secured to the patients’ skin using double sided sticky tape and fixated with tape (two pieces, one at each end of the sensor). The cable of the sensors faced proximally and was taped, allowing for some slack, to the skin. The IMUs were placed arbitrarily on the lower limb so as not to interfere with the position of other sensors: the proximal IMU on the medial aspect of the tibia and the distal IMU on the dorsal aspect of the foot.

Calibration movement
With the lower leg resting on a support, the knee was stabilized with as much knee extension as possible and the ankle was moved from 0° dorsiflexion to maximal plantar flexion. If 0° dorsiflexion was not possible, the degree of limitation was measured using a goniometer and noted. The end position was kept fixed for a minimum of 5 seconds.
2. Knee sensor calibration trial

Sensor placement

Two IMU sensors were secured to the patients' skin using double sided sticky tape and fixated with tape (two pieces, one at each end of the sensor). The cable of the sensors faced proximally and was taped, allowing for some slack, to the skin. The IMUs were placed arbitrarily on the lower limb so as not to interfere with the position of other sensors: the proximal IMU on the lateral aspect of the thigh (at the level of the iliotibial tract) and the distal IMU on the medial aspect of the tibia.

Calibration movement

With the lower leg resting on a support, the knee was stabilized with as much knee extension as possible. The popliteal angle, as well as the angle of the lower leg with the horizontal was measured using a goniometer and the angles noted (Figure 1). Once this data had been registered, the leg was not moved for a minimum of 5 seconds. After 5 seconds in the starting position, the leg was moved from knee extension to maximum knee flexion ensuring that the hip angle did not exceed 90°. The leg was kept fixed in the end position (hip at 90° and knee in full flexion) for a minimum of 5 seconds.

![Diagram of popliteal angle](image1)

a. Popliteal angle

![Diagram of negative angle](image2)

b. Negative angle of the lower leg with the horizontal

![Diagram of positive angle](image3)

c. Positive angle of the lower leg with the horizontal

**Figure 1.** Measuring (a) the popliteal angle, (b) the position of the lower leg indicating a negative angle with respect to the horizontal and, (c) the position of the lower leg indicating a positive angle with respect to the horizontal.
During trials of passive stretch of the MEH, the knee joint was moved from full flexion to extension with the subject supine. The upper leg was kept still at 90° hip flexion and did not move during the measurement (Figure 1A of the manuscript). However, the motion of the ankle during the knee joint test and moment computation was not recorded and not integrated into the computation. Instead, the influence of the ankle movement on the knee torque was considered small. To confirm this assumption, a comparison, based on average age, mass, foot length and shank length of the subjects in the study (n=28) was conducted. The assumptions were verified in two ways: (1) The mass, the center of mass (cm), the gyration lengths (rg) of the shank, foot and combination shank and foot were estimated (MATLAB 7.6.0 R2008a: MathWorks) based on the anthropometric model of Jensen, 1986 [1]. The moment of inertia around the knee rotation axis of the combination of foot and shank for ankle angles -60° plantarflexion and -80° plantarflexion were calculated. The relative difference between the two moments of inertia was 2.41%. (2) The effect of the relative position of foot and ankle on the inertia of the combination shank and foot segment around the knee rotation axis was computed. The relative difference was 2%. These values were considered small. It could therefore be assumed that ankle movement of the magnitude of 20° towards plantarflexion would only have a minor influence on the overall inertia of the shank and foot combination segment.

References:

Figure 1. Bland-Altman plots of the range of differences between sessions against the mean of the two sessions for EMG-change (A, B); Torque-change (C, D); and Work-change (E, F).
for the gastrocnemius (A, C, E) and medial hamstrings (B, D, F) for subjects in the reliability study (n=6). The solid line represents the mean of the differences and the dashed lines the limits of agreement (mean of the differences ±1.96 SD).