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ABSTRACT

This study aimed to evaluate the validity and test–retest reliability of trunk muscle strength testing performed with a latest-generation isokinetic dynamometer. Eccentric, isometric, and concentric peak torque of the trunk flexor and extensor muscles was measured in 15 healthy subjects. Muscle cross sectional area (CSA) and surface electromyographic (EMG) activity were respectively correlated to peak torque and submaximal isometric torque for erector spinae and rectus abdominis muscles. Reliability of peak torque measurements was determined during test and retest sessions. Significant correlations were consistently observed between muscle CSA and peak torque for all contraction types ($r = 0.74–0.85$; $P < 0.001$) and between EMG activity and submaximal isometric torque ($r \geqslant 0.99$; $P < 0.05$), for both extensor and flexor muscles. Intraclass correlation coefficients were comprised between 0.87 and 0.95, and standard errors of measurement were lower than 9% for all contraction modes. The mean difference in peak torque between test and retest ranged from −3.7% to 3.7% with no significant mean directional bias. Overall, our findings establish the validity of torque measurements using the tested trunk module. Also considering the excellent test–retest reliability of peak torque measurements, we conclude that this latest-generation isokinetic dynamometer could be used with confidence to evaluate trunk muscle function for clinical or athletic purposes.

1. Introduction

Strength is an important quality for skeletal muscles of the trunk (particularly flexors and extensors), as it helps in maintaining an optimal posture and core stability (Andersson et al., 1988) and prevents excessive loading on passive structures, such as ligaments, connective tissues and tendons (El-Rich et al., 2004). Indeed, muscle strength of trunk extensors and flexors may contribute to spine stability (Lee et al., 1999), thereby preventing the occurrence of musculoskeletal disorders such as low back pain (Iwai et al., 2004; Yahia et al., 2011). Furthermore, when performed in adequate conditions (i.e., with sufficient learning) (Urzica et al., 2007), training programs aiming at improving trunk muscle strength could also reduce the disability level resulting from back pain (Keller et al., 2008), which is the most common pathology in the general population and constitutes a major public health issue (Schaafsma et al., 2013). In the same way, a strong and stable trunk facilitates the transfer of the forces generated by the upper and lower limb muscles (Kubo et al., 2011), thus contributing to athletic performance in different sports (Kubo et al., 2011; Tanaka et al., 2013).

Muscle force-generating capacity depends on the volume, fascicle length and pennation angle of the muscle (Lieber and Friden, 2000). Muscle strength is thus considered to be proportional to the physiological cross-sectional area (PCSA, muscular determinant) which corresponds to the volume divided by the pennated fiber length (Lieber and Friden, 2000; Stokes and Gardner-Morse, 1999). As the assessment of fascicle pennation angle is difficult to achieve in a valid way on trunk muscles *in vivo*, muscle volume has mainly been inferred from anatomical cross-sectional area (Kubo et al., 2011; Tanaka et al., 2013). The second main physiological determinants of trunk muscle strength is muscle activation (neural determinant), that is mainly regulated by spatial and temporal characteristics of motor unit recruitment during a voluntary contraction (Lippold, 1952).

Assessment of trunk muscle strength, though challenging (Grabiner et al., 1990; Newton and Waddell, 1993), is relevant to

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both clinical practice and research for discriminative, evaluative and predictive purposes (Iwai et al., 2004; Yahia et al., 2011). Besides non-dynamometric evaluation of static and dynamic strength, isokinetic dynamometry is one of the most widely used methods to test (and train) trunk muscle strength (Grabiner et al., 1990; Newton and Waddell, 1993) in an objective way. On the one hand, measurement of trunk flexion and extension strength with different isokinetic machines and at various angular speeds and contraction modes (isometric, concentric and eccentric) has been found to be safe (den Hartog et al., 2010), reliable (Hupli et al., 1997) and sensitive enough to detect muscle weakness (Langrana et al., 1984) and rehabilitation-induced improvements (Brady et al., 1994) in patients with low back pain. On the other hand, however, the reliability of isokinetic and isometric trunk strength testing has frequently been challenged, mainly due to the considerable contribution of hip muscles (Thorstensson and Nilsson, 1982), the migration of the instantaneous center of rotation of the vertebral column during dynamic assessments (Grabiner et al., 1990) and the torque "overshoot" artifacts provoked by impact forces at the end of the movement (Ayers and Pollock, 1999). These potential drawbacks may seriously affect the validity of sagittal plane trunk strength outcomes, as the recorded torque signal might not solely originate from the main prime movers. As a matter of fact, the general validity of trunk strength testing (i.e., the correlation with a reference value) has not been adequately demonstrated to date. Based on decades of experience with isokinetic devices, a dynamometer has recently been introduced, which allows testing in the standing position (more representative of daily-life tasks, Fig. 1), with a correction of gravity throughout the range motion. This device also ensures a comfortable and firm fixation of the subject that overall reduces the potential impact of the above-mentioned sources of artifacts on torque measurements.

Therefore, the main purpose of this study was to examine the construct validity of trunk flexors and extensors muscle strength testing realized with a latest-generation isokinetic dynamometer. Construct validity was evaluated by testing potential zero-correlations between muscle strength obtained in isometric, concentric and eccentric conditions and its two main physiological determinants, namely anatomical cross-sectional area (CSA, as determined with magnetic resonance imaging, MRI), and muscle activation (as determined with electromyography, EMG) of the *erector spinae* (prime mover for trunk extension) and *rectus abdominis* (prime mover for trunk flexion). A secondary aim was to establish the test–retest reliability of isometric, concentric and eccentric strength of trunk flexors and extensors.

2. Methods

2.1. Participants

Fifteen (7 men and 8 women) healthy volunteers (26 ± 4 years; 170 ± 10 cm; 58 ± 23 kg) with no previous history of trunk injury or major pathology participated in this study. All participants were informed regarding the nature, aims and risks associated with the experimental procedure before they gave their written consent to participate. The study was approved by the local ethical committee and was conducted in accordance with the Helsinki Declaration.

2.2. Protocol

The procedure included three test sessions separated by a week. Participants first attended a 90-min familiarization session dedicated to carefully accustom them to the dynamometer and to the testing procedures, and to assess *erector spinae* and *rectus abdominis* CSA by MRI. The subsequent test and retest sessions, lasting approximately 60 min, were identical and dedicated to the assessment of eccentric, isometric and concentric peak torque of trunk flexors and extensors, and of the associated EMG activity.

2.2.1. Dynamometry

Torque measurements were performed using a Con-Trex MJ isokinetic dynamometer (CMV AG, Dübendorf, Switzerland) (Maffiuletti et al., 2007) coupled with a specific trunk module moving on the sagittal plane (Con-Trex TP-1000) (Fig. 1). The dynamometer was designed to enable trunk flexion and extension movements in an upright position with the feet positioned in two horizontal plates and the knees in a slightly flexed position (~10–20°). Trunk flexion movements were performed from −10° to 50° (i.e., 60° range of motion; 0° = vertical position) and vice versa for trunk extension. For gravity correction purpose, the torque resulting from upper body mass was measured in passive mode while the subject was relaxed, throughout the whole range of motion, prior to testing. During each test session, maximal strength testing consisted of:

- 3 consecutive eccentric contractions at angular velocity of $-60^\circ\,s^{-1}$ that were realized separately for trunk flexors and extensors (1 min rest), with the return phase set in passive mode.
- 3 non-consecutive maximal isometric contractions (trunk position: 25°) with a 5-s duration and a progressive rate of force

Fig. 1. Experimental set-up. Frontal (left picture), three-quarter (middle picture) and lateral (right picture) view of the Con-Trex TP-1000 module (a) connected to the isokinetic motor (b). Surface electromyographic activity of *rectus abdominis* (c) and *erector spinae* was synchronously recorded with mechanical data provided by the dynamometer, which was driven by a dedicated computer (d). Trunk extensions and flexions were executed from −10° to 50° (0° = vertical position).

development. Trials were realized separately for trunk flexors and extensors, with a 1-min rest period in between.

- 3 consecutive concentric contractions at angular velocities of $60^{\circ}\,s^{-1}$ and $120^{\circ}\,s^{-1}$ that were realized reciprocally for trunk flexors and extensors. A rest period of 1 min was respected between the two tested velocities.

Eccentric, isometric and concentric trials were presented in a randomized order at the test session, and the same order was reproduced during the retest session. Then, participants were asked to perform three isometric ramp contractions of trunk extensors and flexors, in a randomized order. For trunk extensors, the total duration of these contractions was 16 s, and subjects were requested to progressively modulate torque from 0% to 80% (+10% per second) and then from 80% to 0% (−10% per second) of the previously determined isometric peak torque in extension. For trunk flexors, the procedure was comparable, except that the maximal target the participants could reach without discomfort was set at 70% of isometric peak torque; this resulted in a total duration of 14 s for the ramp contractions of trunk flexors. The characteristics of the ramp contractions (duration, torque modulation and torque range) were defined based on preliminary experiments with healthy volunteers. Isometric torque and the associated EMG activity were synchronously recorded during these ramp contractions. All mechanical signals provided by the dynamometer (i.e., angular position, torque and velocity) were digitized by a 12-bit analog to digital converter (DT 9804, Data Translation, Marlboro, USA) at a sampling frequency of 1000 Hz.

2.2.2. Magnetic resonance imaging (MRI)

During the familiarization session, transaxial T_1-weighted magnetic resonance images of the trunk musculature were obtained using a dedicated low-magnetic field (0.25 T) MRI system (ESAOTE, Genoa, Italy; Fig. 2a; (Guilhem et al., 2013)). Participants were asked to enter into a trunk-dedicated optimized coil and to lie comfortably in the supine position for approximately 20 min. They were placed so that the junction between their second and third lumbar vertebrae (L2–L3) was located in the center of the coil. Longitudinal scans were first performed to identify the portion of the lumbar vertebrae to investigate and to set the anatomical position and orientation of anatomical slices (Fig. 2b). Transverse scanning of T_1-weighted images (thickness: 4 mm) was performed at the mid-level of each of the following scans: L1–L2, L2–L3, L3–L4, L4–L5, and L5–S1. The transverse image located in the middle of the intervertebral level was selected for further analysis (Fig. 2c). The MRI sequence was as follows: spin echo technique; repetition time/echo time: 600 ms/26 ms; 256 × 192 matrix; two excitations: 300 × 300 mm field of view; gap between slices: 6.2 mm; 3 slices per sequence.

2.2.3. Electromyography (EMG)

Surface EMG activity of right and left *erector spinae* and *rectus abdominis* was recorded with silver/silver chloride electrodes. The skin was shaved, gently abraded and cleaned with a solution containing ether, acetone and alcohol to minimize inter-electrode impedance. Pairs of silver/silver chloride electrodes (Blue Sensor N-00-S, Ambu, Baltorpbakken, Danemark) were placed longitudinally with respect to the underlying muscle fiber arrangement according to standard recommendations (Fig. 1c). Wires and electrodes were well secured to the skin to avoid movement-induced artifacts. Raw EMG signals were pre-amplified (Mazet Electronique Model, Electronique du Mazet, Mazet Saint-Voy, France; input impedance: 10 GΩ; common mode-rejection ratio: 100 dB; gain: 600; bandwidth: 6–500 Hz) and sampled through the same digital converter used for mechanical data at 1000 Hz.

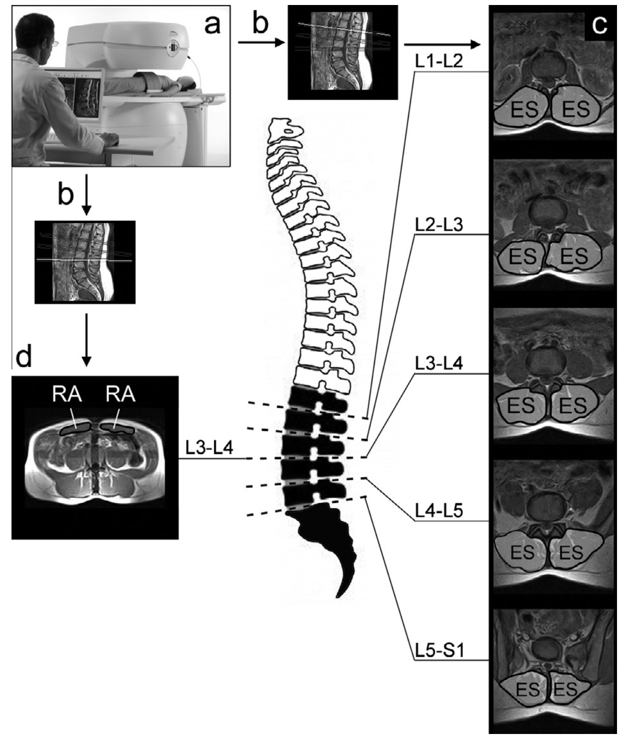

Fig. 2. Magnetic resonance imaging sequence. Low-magnetic field (0.25 T) MRI system (a) was used to determine location and orientation of transaxial T_1-weighted images of trunk musculature (b). Cross sectional area of *erector spinae* (ES) muscles was determined at L1–L2, L2–L3, L3–L4, L4–L5 and L5–S1 level (c) while cross sectional area of *rectus abdominis* (RA) muscles was determined at L2–L3 level (d).

2.3. Data processing

All mechanical and EMG data were analyzed with custom-written scripts (OriginPro 9.0, OriginLab Corporation, Northampton, MA, USA).

2.3.1. Mechanical data

Angular position, torque and velocity were low-pass filtered (6th order zero lag Butterworth filter with a cut-off frequency of 10 Hz) and torque data were consistently corrected for gravity using the upper body passive torque measured before the tests. For maximal contractions, only the trial with the highest eccentric, isometric and concentric peak torque was retained. For submaximal isometric ramp contractions, only the most accurate trial (i.e., where the actual torque was the closest to the target torque) were considered for further analysis. During these ramp contractions, torque was consistently expressed as a percentage of isometric peak torque for respective muscle groups.

2.3.2. MRI data

For every transverse image, a single experienced rater, who was blind to the subjects' characteristics, traced along the inner surface of the considered muscle using a public-domain image processing software (Image J, National Institute of Health, Bethesda, USA). The areas of *erector spinae* (Fig. 2c) and *rectus abdominis* (Fig. 2d) muscles were analyzed. As the field of view did not allow full visualization of the trunk flexors, the CSA of the *rectus abdominis* muscles was determined as the largest cross sectional area between L2 and L3 (Fig. 2d). The anatomical CSAs were calculated by summing the pixels within the outlines. For the *erector spinae* (including the *multifidus* muscle), the 5 images obtained were used for CSA determination. The sum of CSAs of the right and left sides was determined at each slice level. An average value of the 5 slice levels was then calculated to determine a representative value of muscle's CSA. This average value was used to examine its relationship with isometric, eccentric and concentric peak torque (i.e., CSA–torque relationship). The repeatability of the CSA measurements was tested in a preliminary study performed on 5 subjects (unpublished observations). As previously reported with similar methods (Kubo et al., 2011; Raty et al., 1999), our CSA measurements were reproducible with excellent intra-class correlation coefficient (ICC = 0.95), coefficient of variation (CV = 3.8%) and standard error of measurement (SEM = 0.21%) values.

2.3.3. EMG data

All EMG data collected during the maximal isometric contractions and submaximal ramp contractions (Fig. 3a) were first band-pass filtered (6th order zero lag Butterworth filter with a bandwidth frequency of 10–450 Hz). EMG signals were analyzed with a 500-ms root mean square (RMS) moving window then smoothed with a 20 Hz low-pass filter to produce an EMG RMS envelope (Fig. 3b). The submaximal EMG RMS values obtained during the ramp contractions were consistently normalized to the maximal EMG RMS (Fig. 3c) and expressed as a function of submaximal torque. To construct the EMG–torque relationship, the mean EMG RMS activity was calculated every 5% of the isometric peak torque throughout the entire ramp (e.g., from 2.5% to 7.5% for 5% of isometric peak torque; Fig. 3d).

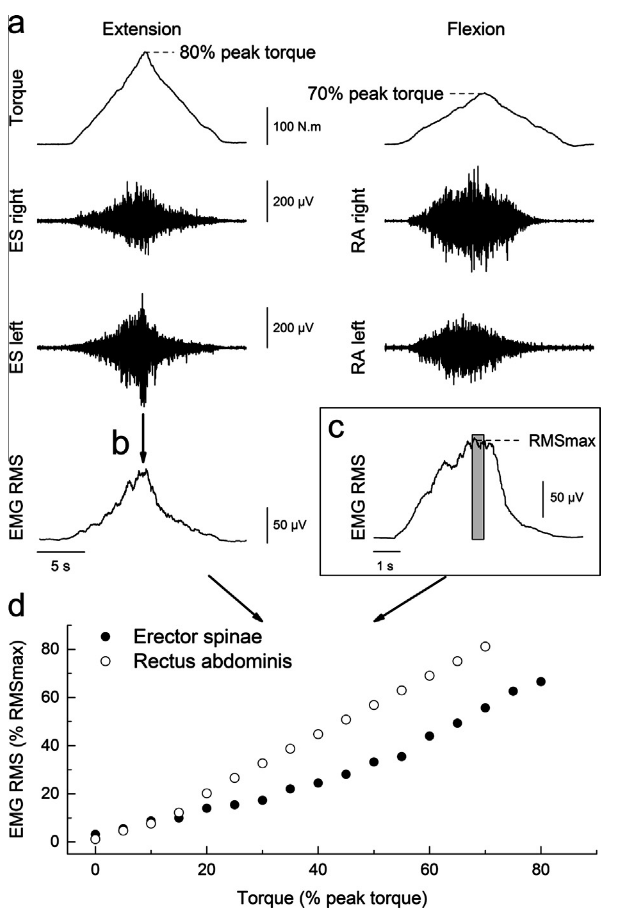

Fig. 3. Description of the EMG–torque relationship analysis. Isometric ramp contractions were performed until 80% (for trunk extension) and 70% (for trunk flexion) of the predetermined isometric peak torque in trunk extension and flexion. Torque measured by the dynamometer and electromyographic (EMG) activity of *erector spinae* (for extension ramp) and *rectus abdominis* (for flexion ramp) were recorded synchronously (a). EMG data were band-pass filtered (10–450 Hz), root mean squared with a time averaging period of 500 ms then smoothed with a 20 Hz low-pass filter to produce an RMS envelope (b), which was normalized to the maximal EMG activity recorded during a maximal isometric contraction (c). Normalized EMG RMS envelope was finally averaged every 5% of peak torque to construct the EMG–torque relationship for trunk extensor and flexor muscles (d).

2.4. Statistical analysis

All statistical analyses were conducted using the software Statistica version 7.1 (StatSoft, Tulsa, Oklahoma, USA). Data distribution was first checked by the Shapiro–Wilk normality test. Because all data were normally distributed, two-way ANOVAs (side × torque level) with repeated measures were performed on EMG activity. The significance level was set at $P < 0.05$. Data are expressed as mean ± standard deviation (SD).

2.4.1. Sample size

A non-inferiority sample size calculation was used to determine the sample size required for the validity analysis. Data for the sample size calculation were collected in a pilot study. Non-inferiority limits were set at 5% of the peak torque (i.e., maximal value obtained in eccentric, isometric and concentric contractions pooled) values obtained in the pilot study (i.e., 16 N m), with a standard deviation of 79 N m. With significance and power level set at 5% and 80% respectively, it was necessary to test 15 participants.

2.4.2. Validity

To determine the validity of isokinetic and isometric trunk extension and flexion torque measured by the dynamometer, linear regression analyses were performed between CSA and peak torque (eccentric, isometric and concentric), and between EMG RMS activity and submaximal isometric torque. The Bravais–Pearson correlation coefficient (r), slope and y-intercept of the linear regressions were calculated (Hopkins, 2000).

2.4.3. Reliability

Test–retest reliability was evaluated for the eccentric, isometric and concentric peak torque measurements using ICC (model 2,1) and SEM as a percentage of the mean values. Considering that the sample size ensured a statistical power above 0.8, the reliability was considered "excellent" for ICC above 0.8 and SEM below 10% (Hopkins, 2000). The mean difference between test and retest measurements (bias) was also calculated and verified with paired t-tests.

3. Results

3.1. Construct validity

No effect of side was observed for both CSA ($P = 0.15–0.36$) and EMG activity ($P = 0.17–0.42$) of *erector spinae* and *rectus abdominis* muscles. Consequently, left and right CSA (construct-1) were summed and EMG RMS values (construct-2) were averaged for all the analyses.

3.1.1. CSA–torque relationship

Fig. 4 depicts the relationship between *erector spinae* and *rectus abdominis* muscle CSA and respective peak torque (trunk extensors and flexors) produced during eccentric (Fig. 4a), isometric (Fig. 4b) and concentric (Fig. 4c and d) contractions. Significant correlations were consistently observed between muscle CSA and peak torque ($r = [0.70–0.85]$; $P < 0.001$).

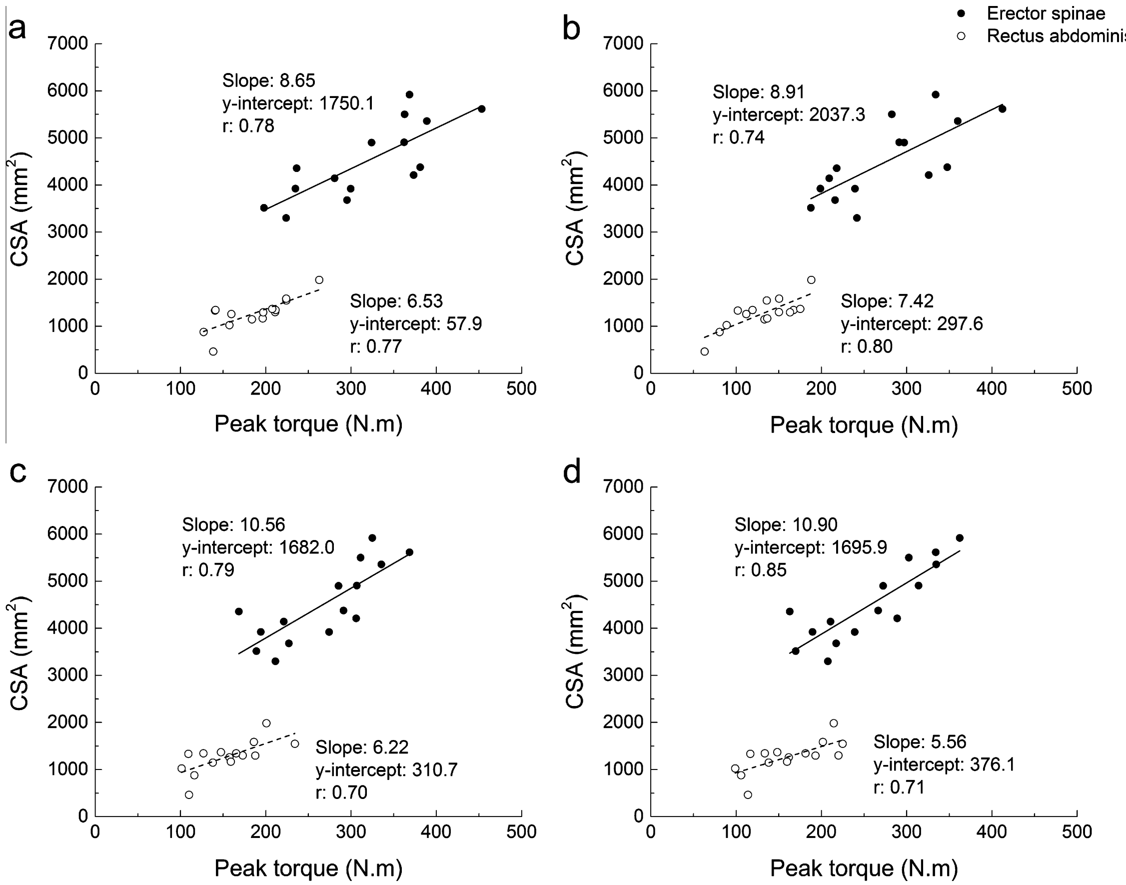

Fig. 4. CSA–torque relationships. Linear correlations between the cross sectional area of the *erector spinae* (black circles) and *rectus abdominis* (white circles) and maximal eccentric (a), isometric (b), 60° s^{-1} concentric (c) and 120° s^{-1} concentric (d) torque measured by the dynamometer. Each plot shows the slope, *y*-intercept and Bravais–Pearson *r* values.

3.1.2. EMG–torque relationship

The correlations between EMG and submaximal isometric torque were significant, with *r* values ⩾0.99 ($P < 0.0001$) for both trunk extensor and flexor muscles (Fig. 5). The slope of the linear regression was slightly higher for *rectus abdominis* (0.83; Fig. 5a) than for *erector spinae* (0.76; Fig. 5b), while *y*-intercept was closer to zero for the *rectus abdominis* (0.56) than for the *erector spinae* (−2.87).

3.2. Test–retest reliability

ICC, SEM and mean test–retest differences of eccentric, isometric and concentric peak torque of trunk flexors and extensors are shown in Table 1. Overall, reliability was excellent with ICC comprised between 0.87 and 0.95, and SEM lower than 9% for all contraction modes and angular velocities. The mean difference in peak torque between test and retest (bias) ranged from −3.7% to 3.7% and was not significant.

4. Discussion

Isokinetic testing has been widely used to measure trunk muscle strength and to study lumbar muscle function in healthy subjects and in patients with low back pain (den Hartog et al., 2010; Newton and Waddell, 1993). Although the reliability of isokinetic peak torque measurements has been often analyzed (Friedlander et al., 1991; Grabiner et al., 1990; Newton et al., 1993; Thorstensson and Nilsson, 1982), studies evaluating the validity component, particularly with respect to the physiological determinants of muscle strength, are scarce. Moreover, to our knowledge, the validity of the isokinetic trunk module tested in this study has not been demonstrated. Consequently, our study aimed to determine the construct validity and the test–retest reliability of isokinetic and isometric torque measurements. Based on CSA–torque (construct-1) and EMG–torque (construct-2) relationships, our findings demonstrated that both trunk extensor and flexor torque measurements were valid for maximal eccentric, isometric and concentric contractions. As a whole, these results establish, at least in part, the validity of trunk extensor and flexor torque measurements using this latest-generation isokinetic dynamometer.

Designed for testing sagittal plane trunk strength in the upright posture, the present dynamometer measured torque values ranging between 152 and 453 N m in trunk extension, and between 99 and 263 N m in trunk flexion, which is in accordance with the values reported for healthy subjects tested in similar conditions (Newton et al., 1993). Previous studies demonstrated a 30% increase of flexor torque from supine to standing position, which is closer to the functional configuration of daily or sportive tasks (McGill, 1996). Moreover, the upright configuration has been shown to reduce the non-willing contribution of muscles crossing the hip joint, thus leading to lower torque variations compared to the supine position (Thorstensson and Nilsson, 1982).

Muscle force-generating capacity has a well-known positive correlation with the amount of contractile tissue (muscle CSA) (Maughan et al., 1983). Our results showed good ($r = 0.74$) to very good ($r = 0.85$) correlations between the anatomical CSA of the trunk extensor muscles and the peak torque measured by the isokinetic dynamometer. Paraspinal muscle CSA has been previously associated with trunk extension peak torque and the former has been therefore recognized as an objective measurement

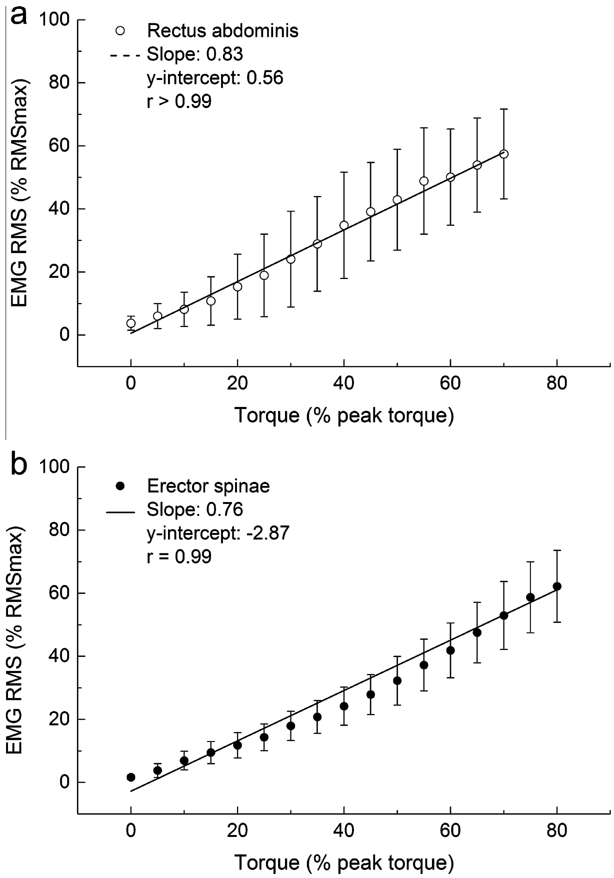

Fig. 5. EMG–torque relationships. Linear correlations between the EMG RMS of *rectus abdominis* (a, white circles) or *erector spinae* (b, black circles), and torque measured by the dynamometer during submaximal isometric ramp contraction performed from 0% to 70% (for trunk flexion) or 80% (for trunk extension) of isometric peak torque. Each plot displays the slope, *y*-intercept and Bravais–Pearson *r* values. Data are presented as mean ± standard deviation.

for back function (Bruce et al., 1997; Keller et al., 1999). Nonetheless, our findings explain between 49% and 72% of the variance in the relationship between CSA and peak torque, which could have been improved by considering PCSA as a more accurate structural determinant of muscle strength. However, given that the reliability of the assessment of fascicle pennation angle of trunk muscles *in vivo* is not established in the literature, anatomical CSA can be considered as a reasonable alternative of muscle volume evaluation (Kubo et al., 2011; Tanaka et al., 2013). Our correlations could also be affected by the numerous synergist muscles potentially contributing to trunk extensor and flexor torque output. Due to the anatomical complexity of trunk musculature, it is difficult to determine how each muscle can be related to the global external torque (Bogduk, 2012). Posterior lumbar spine musculature includes deep intersegmental muscles inserted between adjacent vertebrae, which are too small to be clearly differentiated from MRI, thus complicating the assessment of their CSA (Bogduk, 2012). These intricate muscles also present short lever arms that generate insufficient torque levels to extend the trunk, and therefore they mainly contribute to the postural stability of the spine. According to McGill et al. (1988), the multi-joint muscles *multifidus* and *erector spinae* (lumbar portions), which attach directly to the lumbar vertebrae, constitute the main contributors to trunk extension force-generating capacity, due to their important muscle mass and lever arm length (McGill et al., 1988). However, most of the studies interested in the influence of trunk muscle mass on muscle performance analyzed only one cross-sectional image (Raty et al., 1999; Ropponen et al., 2008), thus sometimes leading to a weak association between muscle CSA and isometric strength (Keller et al., 2004). Keeping these elements in mind, we assessed the CSA of the *erector spinae* (including *multifidus*) at 5 different levels of the lumbar spine, as a representative value of trunk extensor muscle mass. The consistent relationships we observed between CSA and eccentric, isometric and concentric peak torque demonstrated the validity of isokinetic assessment of trunk extensor strength.

In line with a recent methodological study (Asaka et al., 2010), we observed a significant correlation between the anatomical CSA of the *rectus abdominis* and isometric or dynamic (eccentric, concentric) trunk flexion torque (r = 0.70–0.80; Fig. 4). Although *rectus abdominis* was initially regarded as the main determinant of trunk flexion torque (McGill et al., 1988), all trunk flexor muscles actually contribute to external torque, depending on the external loading direction and magnitude, joint mechanical properties or the different recruitment patterns of the other trunk muscles (Cholewicki and VanVliet, 2002). Among the muscles acting on the abdominal wall, the *transverse abdominis* is not only thought to contribute to trunk flexion, but more likely to the reduction of lumbar disc compression (Bogduk, 2012; McGill et al., 1988). The *obliquus externus* and *obliquus internus* muscles participate to trunk flexion but also to trunk rotation when activated in isolation (Bogduk, 2012). Consequently, we chose to investigate the *rectus abdominis* muscle, which allows for reliable acquisition of EMG activity and CSA measurement. Our present findings also confirm the validity of trunk flexor torque measurements, as provided by this latest-generation isokinetic dynamometer.

Besides its structural determinant, the external force generated by a muscle is also the result of the individual force produced by each of the activated motor units (Lippold, 1952). In order to strengthen the construct validity analysis, EMG activity was thus expressed as a function of the concomitant torque exerted under submaximal isometric conditions. We observed a significant and strong correlation between EMG RMS of *rectus abdominis* muscles and trunk flexor torque (r > 0.99; Fig. 5a). Our results are in agreement with previous analysis conducted on *rectus abdominis* and *obliquus* muscles (Brown and McGill, 2008). In the same way, the progressive increase in isometric trunk extension strength was positively correlated to trunk extensor muscle activity (Fig. 5b). While a limited amount of work has been done on EMG–torque

Table 1
Test–retest reliability of peak torque measurements provided by the isokinetic dynamometer at different angular velocities. ICC: intraclass correlation coefficient, SEM: standard error of measurement.

Movement	Velocity ($^\circ$ s^{-1})	Test (N m)	Retest (N m)	Δ (N m)	P value	ICC (2,1)	SEM (%)
Extension	−60	318.8 ± 71.1	318.8 ± 78.6	0.1 ± 26.8	0.99	0.94	5.6
	0	277.3 ± 68.1	272.5 ± 72.5	−4.9 ± 24.3	0.45	0.94	6.4
	60	272.7 ± 63.3	262.5 ± 63.2	−10.2 ± 32.7	0.25	0.87	8.2
	120	253.4 ± 56.4	262.7 ± 75.4	9.4 ± 37.5	0.35	0.88	9.0
Flexion	−60	183.6 ± 37.9	187.3 ± 42.7	3.8 ± 14.6	0.34	0.94	5.9
	0	133.5 ± 36.0	128.4 ± 38.0	−5.1 ± 11.9	0.12	0.95	7.0
	60	153.8 ± 37.2	154.4 ± 41.0	0.6 ± 16.8	0.89	0.94	5.9
	120	161.8 ± 41.6	159.7 ± 46.3	−2.1 ± 17.6	0.65	0.93	8.0

relationships of the abdominal muscles, previous research has generally focused on lumbar *erector spinae* activity, as the highest levels of EMG activity during trunk extension tasks are recorded from these muscles. In accordance to our findings, previous studies reported a linear or curvilinear correlation between trunk extensor muscle activity and strength (Brown and McGill, 2008; Seroussi and Pope, 1987; Stokes et al., 1987), confirming the validity of submaximal static torque measurements provided by the dynamometer used in this study.

A secondary purpose of the present study was to appraise the reliability of isokinetic and isometric peak torque measurements for trunk flexors and extensors. Test–retest data exhibited very low mean differences ($\leqslant$10 N m), and excellent ICC and SEM values. Although trunk extensor concentric torque showed slightly lower ICC and higher SEM values than eccentric and isometric torque, reliability was comparable between $60^\circ\,s^{-1}$ and $120^\circ\,s^{-1}$ angular velocities. Test–retest reliability results were also excellent for trunk flexor muscles, with ICC above 0.90 and SEM values below 8% for all the experimental conditions, which are similar to or better than previous reliability analyses conducted with other dynamometers (Friedlander et al., 1991; Grabiner et al., 1990; Hupli et al., 1997; Newton et al., 1993). The firm position of the subject ensured by the present dynamometer in a position representative of daily-life could have reduced the potential migration of the anatomical rotation axis and the torque overshoot observed with previous ergometers (Thorstensson and Nilsson, 1982). Moreover, the slightly flexed knee position substantially decreased the potential contribution of hip extensor muscles to external torque (Grabiner et al., 1990), while ensuring a comfortable body position. Consequently, our results demonstrated that the latest-generation commercially-available system we used ensured reliable torque measurements during maximal eccentric, isometric and concentric contractions for both trunk flexor and extensor muscles.

The present findings suggest the present isokinetic dynamometer would be suitable to evaluate longitudinal changes in trunk muscle function induced by specific interventions in healthy subjects (e.g., strength training programs). On the one hand, trunk muscle strength has indeed been reported to be positively correlated to sport performance in several activities (Asaka et al., 2010; Kubo et al., 2011). Specifically-designed protocols based on isokinetic trunk testing would thus help coaches in the evaluation of athletes' progression throughout their season or career. On the other hand, assessment of trunk muscle strength could be relevant for monitoring changes in trunk muscle function induced by conservative or surgical interventions in patients with musculoskeletal disorders (e.g., low back pain). Therefore, the use of valid testing methodologies, which entail the measuring device itself (the dynamometer), the stability of the person being measured (i.e., firm fixation of each segment of the body), the procedure for conducting measurements, and the main outcome measure, should be promoted in a wide variety of clinical and research settings. Unfortunately, however, the general validity of trunk strength testing – which entails analyses of construct validity, reliability, normative data, responsiveness/sensitivity to change and interpretability (Terwee et al., 2006) – has not been adequately demonstrated to date. The present study represents the first step towards determining the measurement properties of trunk strength testing using isokinetic dynamometry, in an attempt to render it more relevant to both clinical practice and research for discriminative, evaluative and predictive purposes.

Considering the potential influence of back pain on maximal strength (Mannion et al., 1997), only healthy subjects were included in the present study. Although the transfer of our results to patients suffering from low back pain or injury could not be completely appropriate, we attempted to minimize unwilling sources of variability highlighted in previous reliability studies (e.g., subject position, alignment of the center of rotation, gravity correction (Friedlander et al., 1991; Grabiner et al., 1990; Hupli et al., 1997; Newton et al., 1993)). Due to methodological constraints, our experimental approach did not take into account the whole trunk musculature, including some synergist muscles which can partly contribute to external torque during trunk flexion or extension. In this context, we only considered the main prime movers for the evaluation of trunk extensor (*erector spinae*) and flexor (*rectus abdominis*) strength. Trunk movements executed in the sagittal plane also offer an important lever arm length, which is influenced by the subject's height (McGill et al., 1988). However, trunk muscle mass should be proportional to height to ensure a proper stabilization and an effective mobilization of the spine (Bogduk, 2012; Kubo et al., 2011). Furthermore, force predictions solely based on height or body mass do not appear satisfying (McGill et al., 1988). The linear relationship we obtained between EMG activity and submaximal torque is in line with previous research (Seroussi and Pope, 1987), whereas other studies observed a curvilinear relationship (Stokes et al., 1987). The shape of this curve depends indeed on motor unit recruitment range and hence on muscle fiber type composition, while EMG of each head separately could be non-linearly related to the torque output (Staudenmann et al., 2010). This suggests that part of the apparent non-linearity of the EMG–torque relationship may be due to load sharing with unequal contributions of the synergist muscles at different contraction levels. In this context, the linearity of the EMG–torque relationship has been found to be improved when accounting for the torque generated by the antagonist muscles (Brown and McGill, 2008). Although co-activation could further enhance the fit of the resulting curve, overall our results corroborate a strong positive correlation ($r \geqslant 0.99$) between trunk extensor and flexor muscle activation and submaximal static torque.

In conclusion, based on the strict correlation of sagittal plane trunk torque with its two main physiological determinants (muscle mass and activation), the present methodological study confirms the construct validity of trunk flexor and extensor torque measurements obtained in eccentric, isometric and concentric conditions using the present isokinetic dynamometer and trunk module. Also considering the excellent test–retest reliability of trunk torque measurements, we conclude that this latest-generation isokinetic dynamometer could be used with confidence to evaluate trunk muscle function for clinical or sportive purposes.

Conflict of interest

The authors have no conflict of interest to declare.

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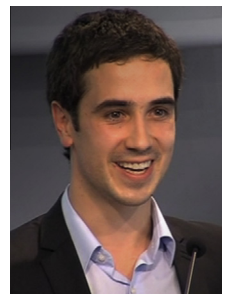

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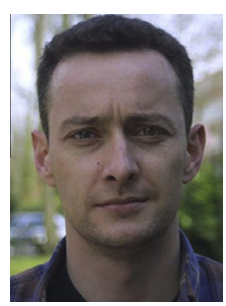

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