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ORIGINAL ARTICLE

# Muscle strength, size, and neuromuscular function before and during adolescence

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## Abstract

**Purpose** To compare measurements of muscle strength, size, and neuromuscular function among pre-adolescent and adolescent boys and girls with distinctly different strength capabilities.

**Methods** Fifteen boys (mean age ± confidence interval: 13.0 ± 1.0 years) and 13 girls (12.9 ± 1.1 years) were categorized as low strength (LS, $n = 14$) or high strength (HS, $n = 14$) based on isometric maximal voluntary contraction strength of the leg extensors. Height (HT), seated height, and weight (WT) determined maturity offset, while percent body fat and fat-free mass (FFM) were estimated from skinfold measurements. Quadriceps femoris muscle cross-sectional area (CSA) was assessed from ultrasound images. Isometric ramp contractions of the leg extensors were performed while surface electromyographic amplitude ($EMG_{RMS}$) and mechanomyographic amplitude ($MMG_{RMS}$) were recorded for the vastus lateralis (VL). Neuromuscular efficiency from the EMG and MMG signals ($NME_{EMG}$ and $NME_{MMG}$, respectively) and log-transformed EMG and MMG vs. torque relationships were also used to examine neuromuscular responses.

**Results** HS was 99–117% stronger, 2.3–2.8 years older, 14.0–15.7 cm taller, 20.9–22.3 kg heavier, 2.3–2.4 years more biologically mature, and exhibited 39–43% greater CSA than LS ($p \leq 0.001$). HS exhibited 74–81% higher $NME_{EMG}$ than LS ($p \leq 0.022$), while HS girls exhibited the highest $NME_{MMG}$ ($p \leq 0.045$). Even after scaling for HT, WT, CSA, and FFM, strength was still 36–90% greater for HS than LS ($p \leq 0.031$). The $MMG_{RMS}$ patterns in the LS group displayed more type I motor unit characteristics.

**Conclusions** Neuromuscular adaptations likely influence strength increases from pre-adolescence to adolescence, particularly when examining large, force-producing muscles and large strength differences explained by biological maturity, rather than simply age.

**Keywords** Electromyography · Mechanomyography · Isometric strength · Youth

## Abbreviations

| | |
|---|---|
| BF | Biceps femoris |
| BF% | Body fat percent |
| CSA | Cross-sectional area |
| EMG | Electromyography |
| FFM | Fat-free mass |
| HS | High strength |
| HT | Height |
| LS | Low strength |
| MMG | Mechanomyography |
| MVC | Maximum voluntary contraction |
| NME | Neuromuscular efficiency |
| VL | Vastus lateralis |
| WT | Weight |

## Introduction

Muscular strength is an important determinant of physical performance in children and adolescents (Grosset et al. 2008; Lambertz et al. 2003). Regardless of the muscle group examined, it has been demonstrated that muscle strength increases significantly during growth and development from childhood to adolescence (De Ste Croix 2007; Grosset

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et al. 2008; Housh et al. 1996; Kluka et al. 2015; Lambertz et al. 2003; Martin et al. 2015; O'Brien et al. 2010; Pitcher et al. 2012; Tonson et al. 2008). During this period, skeletal muscle also undergoes morphological changes, leading to increases in muscle size (Fukunaga et al. 2014; Pitcher et al. 2012; Tonson et al. 2008; Wood et al. 2004). Previous studies have suggested that morphological changes in skeletal muscle size can account for nearly all increases in muscle strength from childhood to adolescence (Fukunaga et al. 2014; Pitcher et al. 2012; Tonson et al. 2008; Wood et al. 2004). However, changes in neuromuscular control also occur during growth and development, and several other previous studies argue that neural adaptations also contribute to increases in muscle strength from childhood to adolescence (Bouchant et al. 2011; Grosset et al. 2008; Lambertz et al. 2003).

One method that has been used to examine the influences of muscle size on increases in skeletal muscle strength during growth and development is to simply express the strength measurement per unit of measured muscle size (i.e., cross-sectional area, volume, fat-free mass, etc.). For example, during a 6-month longitudinal study in prepubescent children, Pitcher et al. (2012) found that the strength:size ratio in the leg extensor and leg flexor muscles remains relatively consistent, suggesting that increases in muscle strength are dependent on increases in muscle size. Fukunaga et al. (2014) suggested that muscle size may be the best predictor of increases in strength from childhood to adolescence in both the leg extensor and plantar flexor muscles. Tonson et al. (2008) indicated that muscle size accounts for nearly all the differences in maximal grip strength (finger flexor muscles) between children and adolescents. Each of these previous studies (Fukunaga et al. 2014; Pitcher et al. 2012; Tonson et al. 2008) also concluded that neuromuscular adaptations could not be ruled out as influencing factors for strength increases. In contrast, many previous studies have demonstrated that muscle size alone is unable to account for increases in strength from childhood to adolescence (Grosset et al. 2008; Housh et al. 1996; Kanehisa et al. 1995; Kluka et al. 2015; Martin et al. 2015; Neu et al. 2002; O'Brien et al. 2010; Seger and Thorstensson 2000; Weir et al. 1999). In fact, these aforementioned studies suggest that neuromuscular control adaptations may account for a large proportion of the increases in skeletal muscle strength. To date, few authors have attempted to quantify differences in neuromuscular control across growth and development (Grosset et al. 2008; Lambertz et al. 2003). Furthermore, no previous studies have concurrently examined electromyographic (EMG) and mechanomyographic (MMG) signals during strength exercises in children.

Surface EMG represents a linear sum of the muscle fiber action potentials passing within the recording areas of the electrodes (Farina et al. 2004). Surface MMG records the low-frequency, lateral oscillations of the active skeletal muscle fibers (Barry and Cole 1990; Orizio 1993; Stokes and Dalton 1993). Together, the EMG and MMG signals, when recorded simultaneously, are complimentary noninvasive assessments of the electrical and mechanical components of neuromuscular control (Akataki et al. 2004; Beck et al. 2006; Coburn et al. 2005). The linear or quadratic increases in bipolar surface EMG amplitude with increasing muscle force (or torque) production during non-fatiguing contractions represent the activation of progressively larger motor units and progressively higher motor unit firing rates, but there is no way of methodologically distinguishing between the two (De Luca 1997; Herda et al. 2010), whereas the pattern of MMG amplitude across the force spectrum displays break points that may distinguish between the influences of motor unit recruitment versus motor unit firing rates for increasing force output (Beck et al. 2006; Coburn et al. 2005; Herda et al. 2010; Ryan et al. 2007). Therefore, examining the EMG and MMG amplitude values from these noninvasive assessments across the force spectrum in children may provide insight regarding muscle activation and motor unit recruitment strategies. Among the large, force-producing leg extensor muscles, the vastus lateralis is the largest and can be used to represent the neuromuscular responses of the other three quadriceps muscles (Häkkinen and Komi 1983; Jakobi and Cafarelli 1998).

Previous studies (Doix et al. 2013; Rose and McGill 2005; Xu et al. 2015) have reported lower EMG amplitude values in children with cerebral palsy, which demonstrates an inhibited muscle activation and subsequently lower skeletal muscle force production compared to healthy children. Compared to adults, previous studies have shown that children have lower muscle activation capabilities during maximal skeletal muscle contractions (Grosset et al. 2008; Kluka et al. 2015; Lambertz et al. 2003; Martin et al. 2015; O'Brien et al. 2010). Thus, in both diseased and healthy young subjects, neuromuscular efficiency (NME), which reflects muscle strength relative to the activation of that muscle (Deschenes et al. 2002), has been used to characterize neuromuscular adaptations that contribute to increases in strength from childhood to adulthood. However, few studies have carefully examined the EMG–force relationships in healthy children (Grosset et al. 2008; Lambertz et al. 2003), and no previous studies have examined the MMG–force relationships in healthy children. If NME and the patterns of responses for both the EMG- and MMG–force relationships can provide insight regarding potential adaptations to neuromuscular control, these noninvasive assessments may be useful for studying the neuromuscular aspects of growth and development.

We are unaware of any studies that have simultaneously recorded surface EMG and MMG to quantify and compare changes in neuromuscular function during growth and

development. Simultaneous recordings of surface EMG and MMG during maximal muscle actions can provide unique insight into the underlying mechanisms responsible for increases in maximal strength. Furthermore, based on Herda et al. (2009, 2010, 2011), surface EMG and MMG can be used during submaximal muscle actions to quantify motor unit recruitment strategies. Additionally, to examine growth and development related differences it is necessary to provide a quantification of biological maturity. This can be indirectly assessed via anthropometric measurements to estimate maturity offset based on Mirwald et al. (2002). This provides a quantification on how many years pre- or post-peak height velocity (PHV) (i.e., the growth spurt) children are. Therefore, the purpose of the present study was to compare measurements of muscle size and neuromuscular function between boys and girls with large differences in maximal isometric strength at distinctly different stages of growth and development. We hypothesize that the stronger, more mature children will have greater muscle size and neuromuscular function, but the differences in strength between groups will not be fully accounted for by muscle size.

## Methods

### Participants

Fifteen boys and 13 girls volunteered for this study. Participants were categorized as either low strength (LS) or high strength (HS) based on torque produced during maximum voluntary contraction (MVC) of the leg extensors. Specifically, subjects who achieved a peak torque ≥ 100 Nm were considered HS while those ≤ 80 Nm were considered LS. This led to HS having peak torque values ≥ 100.02 Nm, and LS having peak torque values ≤ 77.33 Nm. Subject demographics and categorizations are presented in Table 1. All participants reported participating in one or more sports for one to five hours per week during the year prior to this study. Sports included baseball, basketball, cheerleading, cross-country, football, gymnastics, lacrosse, rugby, soccer, softball, speed/power/agility training, swimming/diving, tennis, track and field, trap shooting, volleyball, weightlifting, and wrestling. Both the participants and their parent or legal guardian completed the PAR-Q + 2015 (Warburton et al. 2011) and were included in this study if questions 1–7 were answered "no" or all of the follow-up questions of the PAR-Q + 2015 were answered "no." This study was approved by the University of Nebraska-Lincoln Institutional Review Board for the protection of human subjects (IRB # 20171017495EP, title: *Changes in non-invasive, applied physiological laboratory measurements and field measurements of athletic performance in children and youth: Influences of growth and development*). Each subject signed the approved assent form if they were 7–18-years old, while 5- and 6-year olds verbally assented after being read an age-appropriate assent script, while one parent or legal guardian signed the approved consent form.

**Table 1** Means ± 95% confidence intervals for low strength (LS) and high strength (HS) boys and girls for anthropometrics and during the MVC assessment for MVC torque, EMG and MMG amplitude ($EMG_{RMS}$ and $MMG_{RMS}$, respectively), neuromuscular efficiency ($NME_{EMG}$ and $NME_{MMG}$), and coactivation of the biceps femoris

| | LS | | HS | |
|---|---|---|---|---|
| | Boys ($n=7$) | Girls ($n=7$) | Boys ($n=8$) | Girls ($n=6$) |
| Age (years) | 11.78 ± 1.55 | 11.72 ± 0.51 | 14.06 ± 1.13[a] | 14.47 ± 1.82[a] |
| Height (cm) | 149.13 ± 9.44 | 151.73 ± 6.41 | 164.79 ± 13.11[a] | 165.72 ± 4.89[a] |
| Weight (kg) | 44.50 ± 9.76 | 42.76 ± 7.05 | 66.75 ± 8.17[a] | 63.65 ± 15.62[a] |
| Maturity offset (years) | − 2.11 ± 1.09 | − 0.36 ± 0.64[b] | 0.23 ± 0.92[a] | 2.05 ± 1.16[a,b] |
| Body fat % | 20.39 ± 4.89 | 18.55 ± 6.17 | 19.16 ± 6.22 | 25.40 ± 10.74 |
| Subcutaneous fat | 1.81 ± 0.66 | 2.21 ± 0.60 | 2.11 ± 0.75 | 2.50 ± 0.74 |
| Fat-free mass (kg) | 35.19 ± 7.06 | 34.53 ± 4.41 | 53.88 ± 7.15[a] | 46.31 ± 5.84[a] |
| Quadriceps CSA ($cm^2$) | 36.07 ± 4.62 | 34.79 ± 6.56 | 51.72 ± 9.01[a] | 48.42 ± 12.94[a] |
| Echo intensity (AUs) | 107.27 ± 9.17[c] | 92.65 ± 12.20 | 87.55 ± 8.76 | 91.97 ± 15.53 |
| Torque (Nm) | 64.42 ± 10.10 | 60.84 ± 7.34 | 128.21 ± 26.93[a] | 131.93 ± 32.68[a] |
| $EMG_{RMS}$ (μV) | 196.21 ± 89.83 | 151.77 ± 32.39 | 204.43 ± 72.72 | 197.83 ± 72.45 |
| $MMG_{RMS}$ ($m\ s^{-2}$) | 0.27 ± 0.08 | 0.25 ± 0.05 | 0.43 ± 0.09[c] | 0.25 ± 0.07 |
| $NME_{EMG}$ ($Nm\ \mu V^{-1}$) | 0.40 ± 0.17 | 0.43 ± 0.14 | 0.71 ± 0.25[a] | 0.77 ± 0.44[a] |
| $NME_{MMG}$ ($Nm\ m\ s^{-2}$) | 250.21 ± 56.17 | 244.79 ± 36.00 | 326.55 ± 142.96 | 586.09 ± 277.94[c] |
| Coactivation (%) | 25.82 ± 13.59 | 15.99 ± 3.97 | 19.87 ± 3.59 | 21.20 ± 12.50[d] |

[a]Different from LS collapsed across sex

[b]Different from boys collapsed across group

[c]Different from all other groups

[d]Outliers removed ($n=2$; coactivation = 48.73, 43.40%)

## Research design

A cross-sectional design was used for this study. Participants visited the laboratory twice, once for familiarization and once for the experimental trial. Height (HT), weight (WT), seated height, skinfolds, and ultrasound images were taken at each trial. Only data from the experimental trial were used for comparisons. During the familiarization trial, subjects performed two, 4-s MVCs, and two isometric ramp contractions. The familiarization trial was to allow participants experience with the testing equipment and procedures. Two to 7 days after the familiarization trial, participants completed the experimental trial. Based on torque produced during the MVC in the experimental trial, participants were categorized into the LS and HS groups.

## Anthropometrics and body composition

HT, seated height, and WT were measured using a digital scale and stadiometer (Seca 769, Hamburg, Germany). These variables were used to estimate maturity offset (years) using the Mirwald equation (2002). Percent body fat (BF%) was calculated from skinfold measurements taken with a Lange caliper (Model 68902, Cambridge Scientific Industries, Inc., Cambridge, MD, USA). All skinfolds were taken on the right side of the body at the triceps (vertical fold in the middle of the upper arm, midway between the acromion and olecranon process) and suprailiac (diagonal fold immediately superior to the anterior superior iliac spine) sites and were recorded to the nearest 0.5 mm (Jackson and Pollock 1985). Equations established by Housh et al. (2000) and Brozek et al. (1963) were used to estimate body density and BF%, respectively. WT was then multiplied by BF%, which was then subtracted from WT to quantify fat-free mass (FFM).

## Isometric strength

Isometric torque for the right leg extensor muscles was measured using a commercially available leg extension device (Hammer Strength Plate-Loaded, Iso-Lateral Leg Extension Machine; LifeFitness, Rosemont, IL, USA) that was custom fitted with a load cell (Omegadyne, model LCHD-500, 0–500 lbs; Stamford, CT, USA) while seated in a Biodex System 3 chair (Biodex Medical Systems, Inc., Shirley, NY, USA). Subjects were seated with restraining straps over the pelvis, trunk, and contralateral thigh. The lateral condyle of the femur was aligned with the axis of rotation of the leg extension device. All isometric assessments were performed at a leg flexion angle of 60º below the horizontal plane.

For the MVCs, participants were asked to push against the leg extension device as hard as possible for 4 s while strong verbal encouragement was provided. The highest of the two MVCs was used as the 100% torque value during the ramp contractions.

For the ramp contractions, participants were asked to track their torque production on a computer monitor that displayed the real-time, digitized torque signal overlaid onto a ramp template. The ramp contraction began with a 5-s contraction at 5% MVC, followed by a linear increase in torque lasting 6 s, and ending with a 2-s contraction at 100% MVC (Herda et al. 2010; Ryan et al. 2007, 2008). Each participant completed two ramp contractions, and verbal encouragement was provided during each contraction. The trials that satisfied the following criteria were used for analysis: (a) torque reaching at least 85% of MVC and (b) a maximum tracking error of ± 5% MVC around the ramp template. The ramp template and real-time torque overlay were programmed using LabVIEW 2012 software (National Instruments, Austin, TX, USA). Figure 1 illustrates an example of the torque, EMG, and MMG signals for LS and HS participants.

## Electromyography

Surface EMG signals were recorded from the vastus lateralis (VL) and biceps femoris (BF) with pre-amplified, active electrodes (TSD150B, Biopac Systems, Inc., Santa Barbara, CA) with a center-to-center interelectrode distance of 20 mm, gain of 330 (nominal), input impedance of 100 MΩ, common mode rejection ratio of 95 dB (nominal), and bandwidth of 12–500 Hz. For the VL, the electrode was placed 66% of the distance between the ASIS and lateral border of the patella (Hermens 1999). The longitudinal axis of the electrode was arranged parallel to the angle of pennation of the VL fibers (20°) (Jenkins et al. 2017). For the BF, the electrode was placed 50% of the distance between the ischial tuberosity and lateral epicondyle of the tibia (Hermens 1999). The longitudinal axis of the electrode was arranged in the direction of the line between the ischial tuberosity and lateral epicondyle of the tibia (Hermens 1999). A reference electrode (EL503, Biopac Systems Inc. Santa Barbara, CA, USA) was placed over the tibial tuberosity. Placement sites for EMG and MMG sensors were shaved and cleaned with isopropyl alcohol prior to application.

## Mechanomyography

Surface MMG signals were detected from the VL with an active miniature accelerometer (EGAS-S704-10_Rev C, Measurement Specialties, Inc., Hampton, VA; frequency response = 0–200 Hz, sensitivity = 655.1 mV $g^{-1}$,

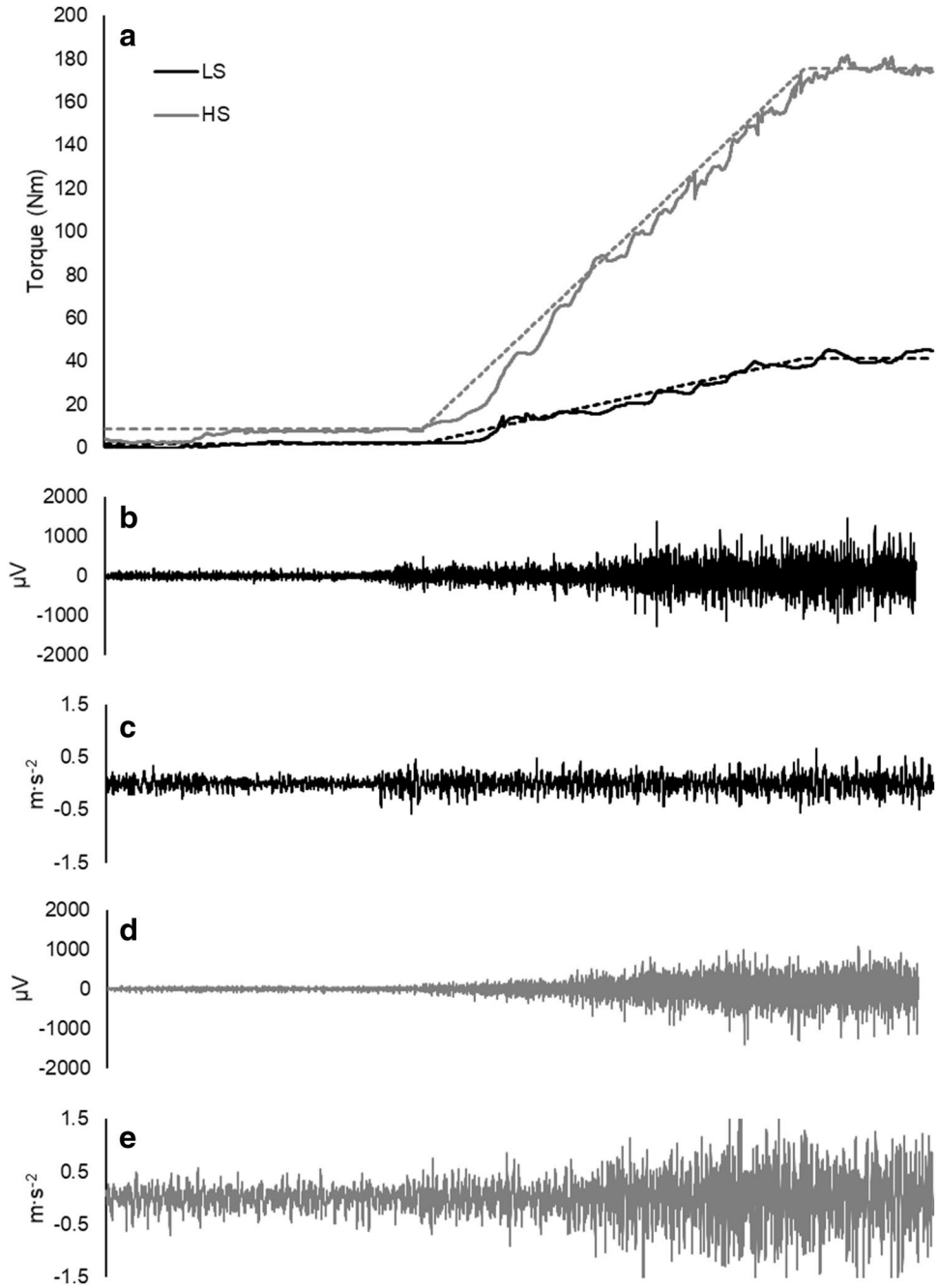


**Fig. 1** **a** An example ramp contraction torque tracing for a low strength (LS) and high strength (HS) participant with **b** LS electromyography (EMG) signal **c** LS mechanomyography (MMG) signal **d** HS EMG signal, and **e** HS MMG signal

range = ± 10 g). Placement for the accelerometer on the VL was 50% of the distance between the anterior superior iliac spine (ASIS) and lateral border of the patella. The accelerometer was fixed to the skin using 3 M double-sided tape.

## Signal processing

Torque (Nm), EMG (μV), and MMG (m $s^{-2}$) signals were sampled simultaneously at 1000 Hz with a Biopac data acquisition system (MP150, Biopac Systems, Inc., Santa Barbara, CA, USA). All signals were stored on a personal computer and processed offline with custom written software (LabVIEW v. 17.0, National Instruments, Austin, TX). MMG signals were digitally filtered (zero-phase 4th order Butterworth filter) with a band-pass of 5–100 Hz. All subsequent analyses were performed on the filtered signals.

MVC torque (Nm) was taken as the highest 0.5-s average torque value during the plateau of the 4-s MVC. The same 0.5-s epoch was taken from the EMG signals of the VL and BF muscles and MMG signals of the VL to calculate EMG and MMG amplitude during the MVC. For the ramp contractions, 0.6-s epochs were taken every 0.1-s for the torque,

EMG, and MMG signals (Ryan et al. 2007, 2008). Torque was the mean of each 0.6-s epoch, while the amplitude for the EMG and MMG signals during each 0.6-s epoch was represented as root mean square ($EMG_{RMS}$ and $MMG_{RMS}$, respectively). To examine coactivation, participants completed a leg flexion MVC using the same procedures as above. $EMG_{RMS}$ of the BF was taken during the highest 0.5-s average torque value during the plateau of the 4-s MVC. Coactivation of the BF was quantified as the $EMG_{RMS}$ of the BF during the leg extension normalized to $EMG_{RMS}$ of the BF during the leg flexion MVC. For the VL, EMG and MMG amplitude were also expressed as neuromuscular efficiency values ($NME_{EMG}$ and $NME_{MMG}$, respectively) during the MVC using previously described calculations (DeVries 1968; Milner-Brown et al. 1986; Moritani and DeVries 1979). In short, $NME_{EMG}$ and $NME_{MMG}$ were calculated by dividing the torque produced by the simultaneously recorded $EMG_{RMS}$ and $MMG_{RMS}$ values, respectively (DeVries 1968; Milner-Brown et al. 1986; Moritani and DeVries 1979).

## Ultrasound imaging

During each visit, panoramic cross-sectional images of the quadriceps were taken to quantify quadriceps muscle cross-sectional area (CSA) and muscle quality [echo intensity (EI)]. Ultrasound images were taken using a portable brightness mode (B-mode) ultrasound-imaging device (GE Logiq e, USA) and a multi-frequency linear-array probe (12L-RS; 5–13 MHz; 38.4 mm field-of-view) (Jenkins et al. 2015). Participants were positioned on a plinth in the supine position while lying fully relaxed with their legs extended and supported on the plinth with feet braced. Panoramic images of the quadriceps were taken at the same site as electrode placement from the most lateral aspect to the most medial aspect of the quadriceps. A generous amount of water-soluble transmission gel was applied to the skin to enhance acoustic coupling and reduce near field artifacts.

Equipment settings were optimized for image quality with a gain of 58 dB and a frequency of 12 MHz. These settings were held constant across participants. Image depth, however, was adjusted based on each participant's leg size and was then held constant for each participant. All images were taken by the same investigator (ZMG) prior to the isometric assessments. Images were taken until three images of acceptable quality were acquired. Images with the highest visual contrast were used for analysis.

Images were analyzed using Image-J Software (National Institutes of Health, USA, version 1.47 v). Prior to analysis, images were scaled from pixels to cm using the Image-J straight-line function. Quadriceps CSA ($cm^2$) was quantified using the polygon function in Image-J to select the maximal region of interest that included as much of the quadriceps muscles as possible while excluding the surrounding fascia (Jenkins et al. 2015). Subcutaneous fat (cm) over the electrode placement site was quantified using the straight-line function that included as much of the subcutaneous fat as possible while excluding the VL. Echo intensity was assessed by gray-scale analysis using the histogram function and was determined using the same region of interest as CSA. Mean EI values were reported as a value between 0 (black) and 255 (white) arbitrary units (AUs).

## Statistical analyses

Means and 95% confidence intervals of all participant demographics (Table 1) and responses during the isometric MVCs were calculated. Percent differences between HS and LS boys and girls were calculated for torque, HT, WT, FFM, CSA, $EMG_{RMS}$, $MMG_{RMS}$, $NME_{EMG}$, $NME_{MMG}$, and coactivation during the MVC (Fig. 2a). To examine the influence of neural factors and growth factors on differences in strength between HS and LS participants, MVC torque was scaled by dividing torque by HT, WT, FFM, and CSA. Scaled variables were then expressed as percent differences between LS and HS participants (Fig. 2b).

For the VL, EMG–torque and MMG–torque relationships were also examined by fitting linear regression models to the natural log-transformed relationships for each subject (Herda et al. 2009, 2011) using the following equations:

$$\ln[Y] = b(\ln[X]) + \ln[a]$$

where $\ln[Y]$ = the natural log of EMG or MMG, respectively, $\ln[X]$ = the natural log of torque, $b$ = slope, and $\ln[a]$ = the natural log of the $Y$-intercept. This can also be expressed as:

$$Y = aX^b$$

where $Y$ is the predicted MMG or EMG values, $X$ is the torque, $b$ is the slope of the original linear equation, and $a$ is the antilog of the $Y$-intercept from the original linear equation.

Prior to analysis, all variables were analyzed for outliers via stem and leaf plots in IBM SPSS v. 25 (Chicago, IL, USA). Nineteen separate two-way factorial analyses of variance (ANOVAs) [sex (boy vs. girl) × group (LS vs. HS)] were used to analyze age, HT, WT, maturity offset, BF%, subcutaneous fat, FFM, CSA, EI, MVC torque, MVC $EMG_{RMS}$ and $MMG_{RMS}$, MVC $NME_{EMG}$ and $NME_{MMG}$, MVC coactivation, and the $a$ and $b$ values from the natural log-transformed models. When appropriate, follow-up analyses included independent samples $t$ tests. One-way ANOVAs were used to analyze scaled data collapsed across sex. All statistical analyses were performed in IBM SPSS v. 25 (Chicago, IL, USA), while percent differences were calculated in Microsoft Excel 2016. An alpha level of $p \leq 0.05$ was considered statistically significant for all comparisons.

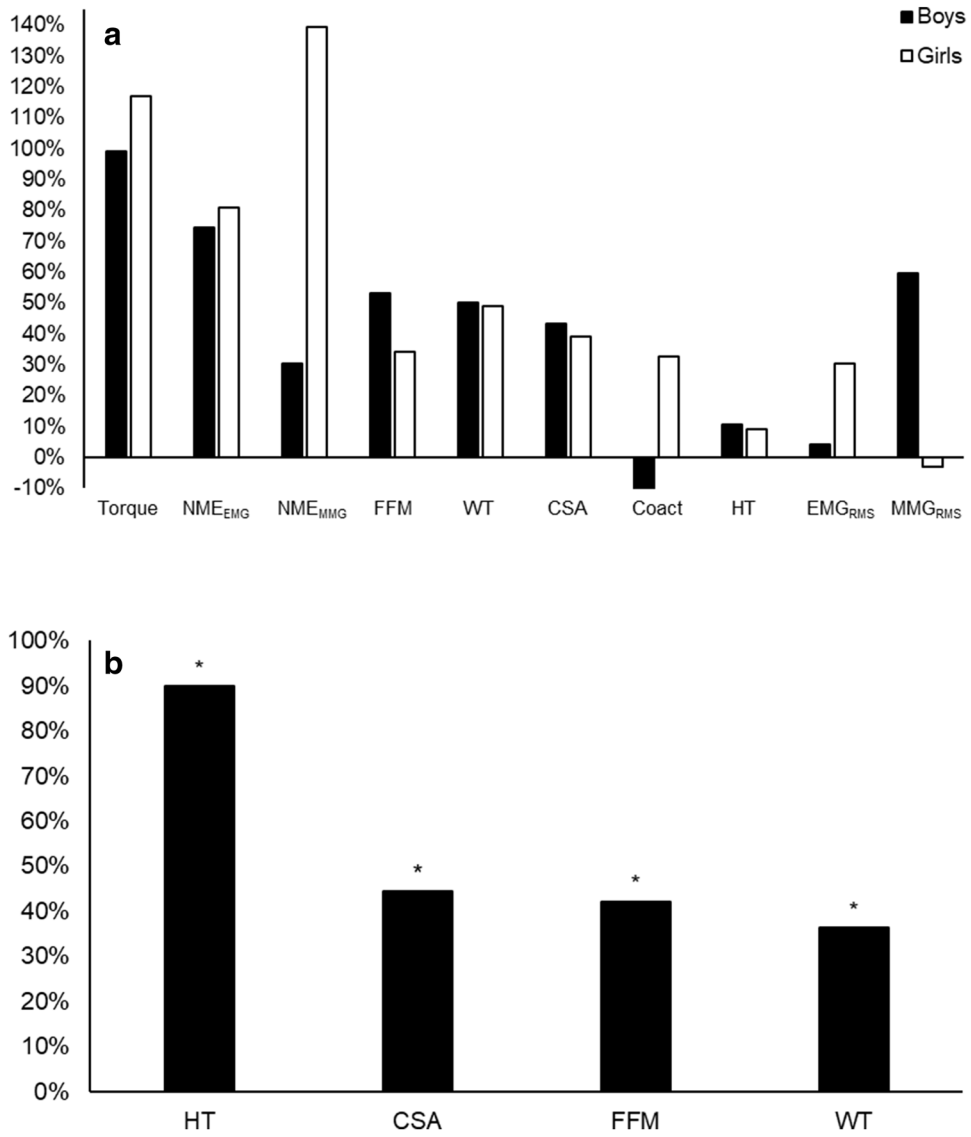


**Fig. 2** **a** Percent differences between low strength (LS) and high strength (HS) boys and girls for torque, neuromuscular efficiency ($NME_{EMG}$ and $NME_{MMG}$, respectively), fat-free mass (FFM), weight (WT), quadriceps femoris cross-sectional area (CSA), coactivation (coact), height (HT), and EMG and MMG amplitude ($EMG_{RMS}$ and $MMG_{RMS}$, respectively) during the MVC, and **b** percent differences between LOW and HIGH groups when torque is scaled for HT, CSA, FFM, and WT. *$p \leq 0.05$

## Results

Results of participant demographics and ultrasound imaging are in Table 1. There were significant main effects for age, height, weight, maturity offset, FFM, and CSA collapsed across sex, such that HS was older, taller, heavier, more biologically mature, and exhibited greater FFM and thigh muscle mass than LS ($p \leq 0.001$). Additionally, there was a significant main effect for maturity offset collapsed across group, such that girls were more biologically mature than boys ($p < 0.001$). There was a significant interaction for EI, indicating higher values (lower muscle quality) for LS boys compared to all other groups ($p \leq 0.037$).

The results of the MVC testing are given in Table 1. There were significant sex x group interactions for $MMG_{RMS}$ and $NME_{MMG}$, indicating that HS boys exhibited greater $MMG_{RMS}$ compared to LS boys and all girls ($p \leq 0.031$), whereas HS girls exhibited greater $NME_{MMG}$ than LS girls and all boys ($p \leq 0.045$). Significant main effects were found for torque and $NME_{EMG}$, showing that HS exhibited greater torque and $NME_{EMG}$ than LS when collapsed across sex ($p \leq 0.022$). There were two outliers for coactivation who were removed from all coactivation analyses. During the MVC, there were no significant interactions or main effects for coactivation ($p \geq 0.135$).

Results of the scaling are given in Fig. 2. Since there were no sex differences in strength and size, we examined the composite HS (boys and girls) and LS (boys and girls) groups for the scaled variables. There were group differences for MVC torque when scaled for HT, WT, CSA, and FFM ($p \leq 0.031$).

Analysis of the *a* and *b* values from the natural log-transformed $EMG_{RMS}$ and $MMG_{RMS}$ vs. Natural log-transformed torque relationships (Fig. 3) showed no significant

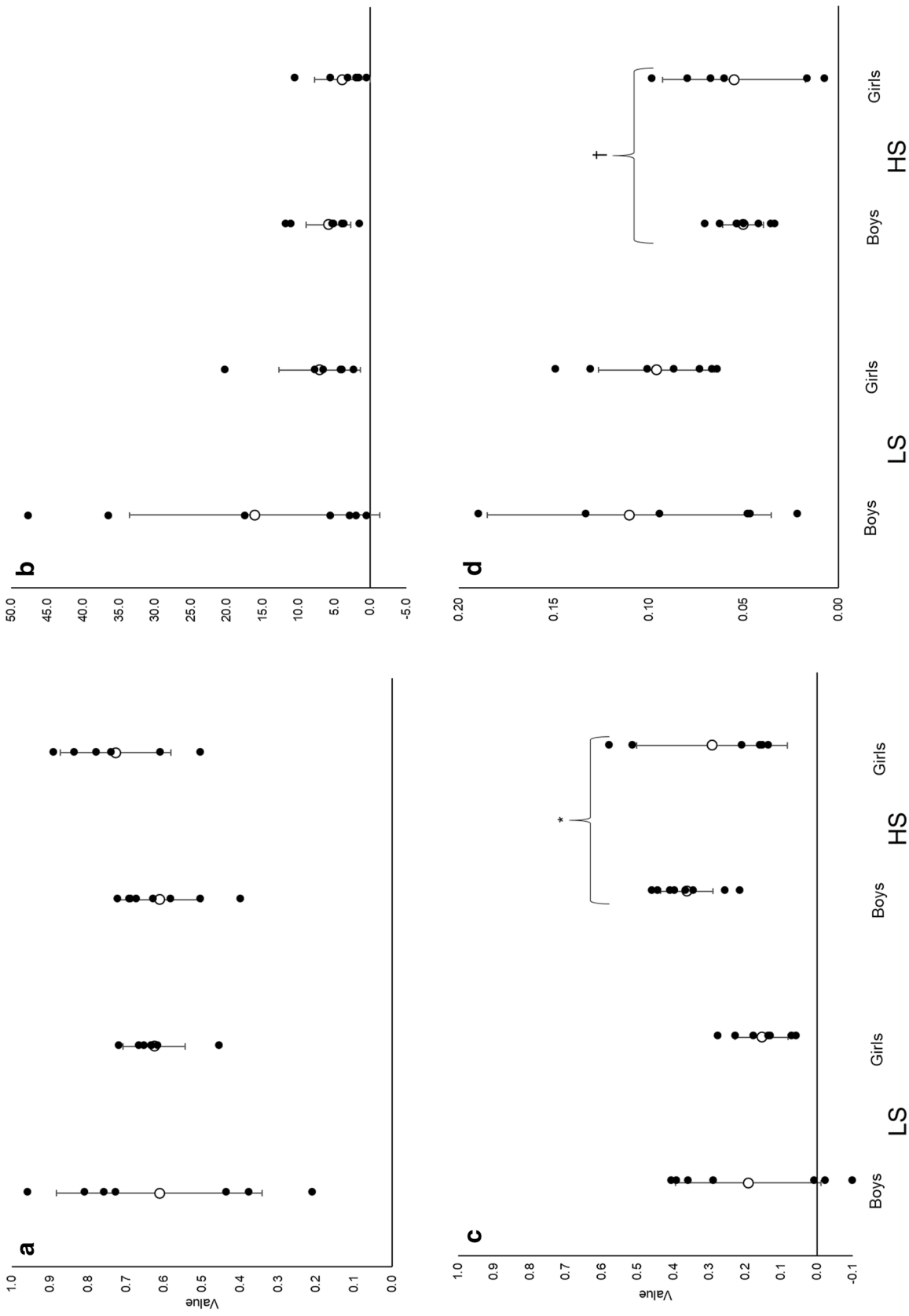


**Fig. 3** Means ± 95% confidence intervals and individual values for the **a** electromyographic (EMG) *b* values, **b** EMG *a* values, **c** mechanomyographic (MMG) *b* values, and **d** MMG *a* values from the natural-log transformed EMG and MMG amplitude vs. torque relationships for the low strength (LS) and high strength (HS) boys and girls. *Significantly greater for HS than LS collapsed across sex, †Significantly lower for HS than LS collapsed across sex

interactions ($p > 0.05$). However, there were main effects for the $MMG_{RMS}$ $a$ and $b$ values, such that $a$ values were lower for HS collapsed across sex ($p = 0.010$) and $b$ values were greater for HS collapsed across sex ($p = 0.015$).

## Discussion

By design, the HS boys and girls in this study were 99–117% stronger during isometric leg extension muscle actions than LS boys and girls (Fig. 2). The HS boys and girls were also 2.3–2.8 years older, 2.3–2.4 years more biologically mature, 9–11% taller, and 49–50% heavier (Table 1; Fig. 2a) than the LS boys and girls, but there were no differences in body fat percent (Table 1). Despite the magnitudes of strength differences, HS boys and girls demonstrated only 39–43% greater quadriceps femoris CSA and total FFM than the LS boys and girls (Fig. 2a). In contrast, the HS groups exhibited 31–139% greater NME (with EMG or MMG) than LS groups (Fig. 2a). Scaling the strength values by HT, WT, CSA, and FFM reduced, but did not eliminate, the statistical differences in strength between HS and LS groups (Fig. 2b). Thus, our findings were consistent with previous studies (De Ste Croix 2007; Housh et al. 1996; Kanehisa et al. 1995; Kluka et al. 2015; Martin et al. 2015; Neu et al. 2002; O'Brien et al. 2010; Seger and Thorstensson 2000; Weir et al. 1999) suggesting that increases in muscle strength during growth and development cannot be fully explained by changes in body size or increases in muscle mass. Our findings also add to a growing body of evidence (Bouchant et al. 2011; Grosset et al. 2008; Kluka et al. 2015; Lambertz et al. 2003; Martin et al. 2015; O'Brien et al. 2010) suggesting that neuromuscular adaptations are contributing factors to strength changes during pubescence.

The unique aspect of the present study is the research design that separated sex and groups by muscle strength as well as clear, balanced differences in biological maturity. Based on maturity offset (Table 1), the LS boys were 2.1 years away from reaching PHV, the HS boys and LS girls were currently experiencing PHV (− 0.5 to + 0.5 years), while the HS girls were 2.1 years past PHV. Thus, despite the similarity in age between boys and girls in their respective strength groups, our sample distinctly represented all three phases of growth and development: pre-pubescent, pubescent, and post-pubescent. Previous studies have reported less disparate differences in muscle strength, muscle size, and/or biological maturity. For example, Fukunaga et al. (2014) reported 34–42% greater muscle strength, 31–42% greater muscle volume, and 10–15% greater muscle thickness in pubescent versus pre-pubescent males separated by an average age difference of only 0.7 years. Pitcher et al. (2012) showed 11–19% increases in muscle strength, 7–14% increases in muscle volume, and 4–11% increases in muscle CSA after a 6-month period of growth and development in 7–8-year-old boys and girls. Tonson et al. (2008) found 53% greater muscle strength, 54% greater muscle volume, and 45% greater muscle CSA in children (Tanner stage ≤ 2) compared to adolescents (Tanner stage > 2) separated by 2 years of age. Yet, these collective studies concluded that the increases in muscle strength observed during growth and development can be accounted for by increases in muscle size alone (Fukunaga et al. 2014; Pitcher et al. 2012; Tonson et al. 2008; Wood et al. 2004). Compared to the 99–117% differences in muscle strength between the LS and HS groups observed in the present study, it may have been easier to account for the smaller magnitudes of strength differences (11–53%) reported in these previous studies. Yet, none of these previous studies reported any data by which to rule out the potential influence of neuromuscular adaptations. Based on the results of the present study, there will be continuing debate regarding the neural versus morphologic factors that contribute to strength changes during growth and development.

Tonson et al. (2008) provided compelling evidence to support the measurement of muscle volume with magnetic resonance imaging (MRI) when calculating ratios of muscle strength:size to equate changes with growth and development in the finger flexor muscles. Although the authors concluded that muscle size, particularly muscle volume, accounted for the differences in finger flexor strength, it was noted that neural factors may have also contributed. Previous research has also demonstrated differences between neural versus hypertrophic influences on strength changes in smaller muscles requiring fine motor control movements (like the finger flexors) compared to larger muscles that control gross force production (Akataki et al. 2003; De Luca et al. 1982; De Luca and Kline 2012; De Luca and Hostage 2010; Herda and Cooper 2015; Lawrence and De Luca 1983; MacIntosh et al. 2006). Motor unit firing rates (i.e., rate coding) exert a larger influence on the force produced by individual muscle fibers and motor units in smaller, fine motor skill muscles (Akataki et al. 2003; De Luca et al. 1982; De Luca and Hostage 2010; Herda and Cooper 2015; Lawrence and De Luca 1983). For example, in the first dorsal interosseous (FDI) muscle, motor unit recruitment is the primary mechanism responsible for increasing force up to 50% of maximal isometric contraction strength, while rate coding increased force output from 50 to 100% (Akataki et al. 2003; De Luca et al. 1982; De Luca and Kline 2012; De Luca and Hostage 2010; Herda and Cooper 2015; Lawrence and De Luca 1983; MacIntosh et al. 2006). In contrast, the VL muscle may rely upon motor unit recruitment up to approximately 80%, after which rate coding determines force output to 100% of maximal force production (Herda and Cooper 2015; Herda et al. 2009; Madeleine et al. 2001; Ryan et al. 2007, 2008). Therefore, although the VL and finger flexor

muscles have similar muscle fiber type distributions (Johnson et al. 1973), this discrepancy in neuromuscular control between small, fine motor skill muscles and large, gross force production muscles may influence the mechanisms by which these muscles get stronger (Granacher et al. 2011; Ozmun et al. 1994; Ramsay et al. 1990). Thus, the differences between the findings of Tonson et al. (2008) and the present study may be related to fundamental differences in neuromuscular control strategies that may influence strength increases in the finger flexor versus leg extensor muscles.

Although the quadriceps femoris is composed of four muscles, the EMG and MMG amplitude values from the VL muscle can accurately represent the other three muscles (Häkkinen and Komi 1983; Jakobi and Cafarelli 1998). Thus, the NME calculated in the present study reflects muscle strength relative to the activation of that muscle, which can be compared across participants to examine the potential neural contributions to strength differences (DeVries 1968; Milner-Brown et al. 1986; Moritani and DeVries 1979). The increases in NME that accompany increases in muscle strength may represent muscle fibers becoming more effective at producing tension (DeVries 1968) or may represent the same motor units being activated at lower stimulation thresholds (DeVries 1968). Furthermore, Milner-Brown et al. (1986) proposed that NME provides a quantification of excitation–contraction coupling, suggesting that increases in NME lower the action potential threshold necessary to result in force production. Several previous studies have quantified $NME_{EMG}$ among adults (Arabadzhiev et al. 2010; Castaingts et al. 2004; Chaves et al. 2012; Deschenes et al. 2002; DeVries 1968; Milner-Brown et al. 1986; Moritani and DeVries 1979) and have generally concluded that stronger, more trained adults have greater $NME_{EMG}$ than their weaker, less-trained counterparts, the implication being that increases in muscle strength include a neural adaptation. However, less is known regarding changes in $NME_{EMG}$ during growth and development among children and adolescents, and no literature has quantified $NME_{MMG}$ in children or adults. Previous authors have examined age-related changes in $NME_{EMG}$ among prepubertal children (Grosset et al. 2008; Lambertz et al. 2003). Collectively, these authors found 69–207% increases in $NME_{EMG}$ across 7- to 11-year-old children. However, both previous studies (Grosset et al. 2008; Lambertz et al. 2003) used prepubertal children and only examined $NME_{EMG}$ changes relative to chronological age. The results of the present study demonstrated 74–81% greater $NME_{EMG}$ in the older, more biologically mature HS group. Furthermore, the most biologically mature group (HS girls) exhibited 80–139% greater $NME_{MMG}$ than all other groups. Thus, in conjunction with the findings of previous studies in adults (DeVries 1968; Milner-Brown et al. 1986), it is possible that the stronger, more mature children have a lower action potential threshold necessary to result force production during a maximal contraction. Collectively, the differences in NME (from EMG and MMG) relative to strength and biological maturity suggest that neural adaptations likely contribute to strength increases during growth and development.

The $EMG_{RMS}$– and $MMG_{RMS}$–force relationships are noninvasive assessments that are thought to contain information regarding muscle activation, motor unit recruitment, and motor unit firing rates that control skeletal muscle force production from low to high force production during non-fatiguing muscle actions (Beck et al. 2006; Herda et al. 2009; Orizio et al. 2003; Ryan et al. 2008). Herda et al. (2009; 2010, 2011) proposed that the *b* and *a* values from the natural-log transformed $EMG_{RMS}$– and $MMG_{RMS}$–torque (or force) relationships provided individual, unique information regarding motor control strategies, and potentially muscle fiber types in adults. It was suggested that the *b* value from the natural log-transformed relationship reflects the patterns of responses for the $EMG_{RMS}$– and $MMG_{RMS}$–torque relationships such that values greater than 1 reflect an exponential acceleration in amplitude, values equal to 1 reflect linear increases in amplitude, while values less than 1 reflect a plateau or deceleration in amplitude across the force spectrum for EMG and MMG signals. Specifically, Herda et al. (2010) found that the $EMG_{RMS}$–torque relationship was incapable of distinguishing among predominantly slow-twitch, trained mixed fiber types, and untrained mixed fiber types in the VL muscle. In contrast, the authors (Herda et al. 2010) hypothesized that the *b* values of the $MMG_{RMS}$–torque relationship reflected fiber type-related differences regarding the emphasis of motor unit recruitment versus motor unit firing frequency to increase force production. Larger *b* values reflected a greater reliance on motor unit recruitment for torque production, which in turn, was associated with a greater proportion of type II fibers (Herda et al. 2009, 2010; 2011). Lower *b* values were thought to reflect an earlier shift in the force spectrum from motor unit recruitment toward motor unit firing frequencies to increase force production, which was associated with a greater proportion of type I fibers. The *a* values were thought to reflect a "gain factor" that represents shifts in the $EMG_{RMS}$– or $MMG_{RMS}$–torque relationships upward or downward, and may be associated with the filtering effects of subcutaneous fat between the surface of the skin and the muscle or the stiffness of the muscle that could restrict the lateral oscillations recorded with MMG (Herda et al. 2009, 2010; 2011). The results of the present study demonstrated no differences between HS and LS for the *b* or *a* values from the $EMG_{RMS}$–torque relationships (Fig. 3a, b). However, there were higher *b* values and lower *a* values for the HS group during the $MMG_{RMS}$–torque relationships (Fig. 3c, d), which were largely consistent with the findings of Herda et al. (2009, 2010, 2011).

The greater *b* values in the HS group in the present study suggest a similarity to the neuromuscular control strategies associated with type II motor unit behavior. Herda et al. (2010) reported that the VL may rely on motor unit recruitment up to approximately 80% of the isometric MVC to increase force output, and the $MMG_{RMS}$–torque relationship *b* values associated with a greater proportion of type II fibers ranged from 0.49 to 1.27. In the present study, although the HS group exhibited greater *b* values than LS for the $MMG_{RMS}$–torque relationships (Fig. 3c, d), the *b* values for the HS group ranged from 0.13 to 0.58 and were comparable to the endurance-trained adults with greater type I muscle fibers reported by Herda et al. (2010) ranging from 0.23–0.39. These findings are consistent with previous studies that have reported greater proportions of type I fibers in children compared to adults (Eriksson et al. 1971, 1973; Ratel et al. 2008) and more recent evidence (Birat et al. 2018) showing similar metabolic profiles between endurance-trained adults and children. Overall, the results of the present study, in conjunction with previous studies (Herda et al. 2009, 2010; 2011), suggest that the $MMG_{RMS}$–torque relationship may be more robust than, yet complimentary to, the $EMG_{RMS}$–torque relationship for addressing the emphasis of motor unit recruitment versus motor unit firing frequencies. Children with lower strength profiles may use neuromuscular control strategies that are typically associated with greater type I muscle fibers in adults that rely on motor unit recruitment to increase force output to approximately 60% MVC and then use motor unit firing rate to increase force thereafter. In contrast, children with higher strength profiles may use neuromuscular control strategies that are typically associated with greater type II muscle fibers in adults that rely on motor unit recruitment to increase force output to approximately 80% MVC and then use motor unit firing rate. Recent studies (Herda et al. 2015, 2019; Pope et al. 2016; Trevino et al. 2018) have hypothesized that a greater rate of growth in the high-threshold, type II motor units may explain the increased reliance on type II motor unit characteristics as children grow. These studies have suggested that the changes in motor unit recruitment strategies, as well as increased size of the high threshold motor units, are the primary neural mechanisms leading to increases in strength. The greater *b* values in the HS group, in conjunction with the greater NME, support this hypothesis, suggesting that changes in motor unit recruitment strategies, such as changes in the motor unit action potential and recruitment threshold relationship, lead to increased force producing capabilities across growth and development (Goldberg and Derfler 1977; Herda et al. 2019; Milner-Brown and Stein 1975; Pope et al. 2016). However, no previous studies have specifically examined the growth rates of high- versus low-threshold motor units and their influences on strength increases during periods of growth and development in children. Future research is needed to examine whether the neuromuscular adaptations that occur with growth and development include shifts in the motor control strategies that increase force production.

Regarding the *a* values, Herda et al. (2010) suggested that greater subcutaneous fat may act as a low-pass filter for the MMG signal that decreases the *a* values. However, if subcutaneous fat causes lower amplitude values and lower *a* values in the MMG signal, similar reductions should also be observed in the EMG signal. The present study found no differences in BF% or subcutaneous fat between LS and HS groups (Table 1), nor were there any differences between groups for the *a* values of the $EMG_{RMS}$–torque relationships (Fig. 3b); therefore, it is possible that something other than subcutaneous fat may affect the *a* values of the $MMG_{RMS}$–torque relationships. However, the present study findings are consistent with Herda et al. (2010), who demonstrated that stronger participants exhibited lower *a* values than weaker participants. This may be due to greater muscle stiffness in the stronger participants that restricted the lateral oscillations that constitute the MMG signal (Akataki et al. 2004; Orizio et al. 2003). In theory, greater muscle stiffness is directly related to greater muscle mass (Ryan et al. 2011), which was also demonstrated in the present study in the HS group (Table 1). Greater muscle stiffness and muscle mass may limit the amplitude of the MMG signal resulting in a shift downward of the $MMG_{RMS}$–torque relationship and a subsequently lower *a* value.

In conclusion, the unique design of the present study demonstrated that separating children and adolescents by maximum isometric strength simultaneously identified the groups by biological maturity, rather than simply age. Thus, the present sample represented male and female children and adolescents who were pre-pubescent, pubescent, and post-pubescent. Furthermore, the present study demonstrated that scaling maximal strength by HT, WT, CSA, and/or FFM was unable to account for the 99–117% differences in strength between HS and LS groups, which implies the presence of factors beyond increases in muscle size (possibly neuromuscular adaptations) that contribute to strength increases during growth and development. Previous authors have suggested that muscle volume alone may account for the increases in strength observed during growth and development (Tonson et al. 2008). However, the magnitudes of strength differences observed and physiological differences between the finger flexor muscles examined by Tonson et al. (2008) and the leg extensor muscles in the present study may at least partially explain the discrepancies between findings. Nevertheless, the results of the present study also indicated 74–139% greater NME in the HS group using EMG and MMG measurements, which strongly suggests that neuromuscular efficiency contributes to strength increases during growth and development. Compounding

evidence from the larger *b* values and lower *a* values exhibited by the $MMG_{RMS}$–torque relationships in the HS group also suggests that shifts in neuromuscular control strategies toward type II fiber characteristics may also occur during growth and development, resulting in greater expressions of strength. Thus, we conclude that neuromuscular adaptations do influence strength increases during growth and development, particularly when cross-sectionally examining large force-producing muscles and large magnitudes of strength differences explained by biological maturity, rather than simply age. Longitudinal research that quantifies both changes in muscle size and neuromuscular control during growth and development are necessary to test the hypotheses underlying the mechanisms associated with strength increases in children and adolescents.

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## Compliance with ethical standards

**Conflict of interest** All authors declare that they have no conflicts of interest.

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