Influence of overweight on the active and the passive fraction of the plantar flexors series elastic component in prepubertal children

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Influence of overweight on the active and the passive fraction of the plantar flexors series elastic component in prepubertal children. J Appl Physiol 114: 73–80, 2013. First published October 11, 2012; doi:10.1152/japplphysiol.00241.2012.—The influence of overweight, as a precursor to obesity, was analyzed on the elastic properties of the plantar flexor tendon, which lead to a slight decrease in the elastic modulus (5). Therefore the precise nature of the adaptive response of the muscle-tendon unit, notably the elastic properties, in overweight/obese children remains unclear.

The fact that musculotendinous stiffness adapts to changes in the functional demand is well known in the literature, both in isolated muscles from animal studies (e.g., 3, 21) and in muscle groups from studies in humans (e.g., 10, 25). In humans, different patterns of musculotendinous stiffness adaptation are reported in the literature, which highlights the fact that the active and passive fraction of the series elastic component (SEC) can adapt differently. For instance, plyometric training causes the musculotendinous stiffness to decrease (25), which can be attributed to a decrease in the stiffness of the active fraction of the SEC (20, 25) due to an increased proportion of fast-type fibers (3). While studies using ultrasound or the alpha method reported an increase in tendon stiffness (10, 20), which may possibly be due to the imposed overload (9). It is, therefore, of interest to separate the mixed mechanical contribution of the active and the passive fraction of the SEC during musculotendinous stiffness evaluation. This separation can be performed using the so-called alpha method, which is based on the animal model developed by Morgan (36) and has recently been used in human testing (e.g., 19, 20).

Finally, the literature suggests that there is a reduced motor unit activation and a lower strength-to-mass ratio in the knee extensors of obese adolescents (6). Because MT stiffness is directly related to the number of cross bridges activated to biomechanics of walking (39). This impairment could be related to changes in the muscle and/or the tendons, such as the muscle fiber type distribution, the mechanical characteristics, the architectural and structural parameters, or even the biochemical properties. It was shown that diet-related obesity in human skeletal muscle (rectus abdominus or vastus lateralis) lead to a decrease in the slow-type fiber percentage and an increase in the fast-type fibers (26, 44). In contrast, transgenically induced obesity in rodents (soleus or EDL) is associated with a decrease in the fast-type fiber content and an increase in slow-type fibers (2). About 1% of the type I content in gastrocnemius muscle in obesity-prone rodents fed a high-fat diet. Even so, these counterintuitive changes are not fully explained (the differences between human and rodent model, in the obesity model or in the muscle studied). These results clearly indicate that obesity affects the proportion of the muscle fiber types. Moreover, transgenically induced obesity in rats results in a decrease in the mass-averaged diameter of collagen fibrils as well as an increase in the maximal strain and stress in the deep digital flexor tendon, which leads to a slight decrease in the elastic modulus (5). Therefore the precise nature of the adaptive response of the muscle-tendon unit, notably the elastic properties, in overweight/obese children remains unclear.

IT IS WELL DOCUMENTED THAT obese children are more likely to remain obese into adulthood (37) and develop a higher incidence of metabolic-related diseases (type-2 diabetes, heart disease, arthritis) (38) leading to a reduced quality of life. In addition to the metabolic and cardiovascular changes, an excess of weight involves extra/excessive loading of the musculoskeletal system due to progressive combined functional influences of the body mass against gravity and movements and displacements during the performance of daily activities (7, 8). Indeed, greater net metabolic costs during walking, which were normalized to the body mass, were observed in obese adolescents and were partly attributed to the differences in the

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develop torque, possible differences in the voluntary activation capacities can influence MT stiffness evaluation (31).

To the best of our knowledge, there have been no studies concerning the biomechanical consequences of overweight prepubertal children, which is seen as a precursor of obesity. Therefore the aim of the present study was to quantify the changes in the musculotendinous elastic properties of the triceps surae muscle group in these children. Additionally, the alpha method will be applied to this human study to differentiate the changes in the active portion of the muscle to the changes in the passive portion of the SEC. Finally, the influence of muscle activation capacities on SEC properties was investigated.

**MATERIALS AND METHODS**

**Subjects**

Forty-four boys and girls, ages 9 years ± 4 mo from six public schools in the city of Recife, Pernambuco, northeast of Brazil, participated in this study. Based on the criteria given by the World Health Organization (WHO) (12), the body mass index (BMI) was adopted to divide children into a control group (CON n = 23, 11 boys/12 girls WHO BMI < 1 SD > Z score < 1 SD) or an overweight group (OW n = 21, 11 boys/10 girls WHO BMI 1 SD > Z score < 3 SD). Each child was evaluated individually using the corresponding WHO reference data matched for their age and gender. The WHO BMI reference data for the age range of 108 ± 4 mo and both gender are for CON < 1 SD Z score to 1 SD Z score = 16.2 ± 1.5 kg m⁻² and for OW 1 SD Z score to 3 SD Z score = 21.5 ± 2.1 kg m⁻².

The experimenter and the legal guardians determined the pubertal status. Based on breast development, pubic hair, and no apparent changes in the voice and skin all children were classified as prepubertal. The study obtained approval by the Committee of Ethics in Research of the Federal University of Pernambuco (CAAE-0345.0.172.000-08) according to the Regulatory Research Standards involving Human Beings, Resolution 196/96 of the Brazilian National Health Council and in accordance to the declaration of Helsinki. The legal guardians were fully instructed about the experimental procedure and signed a term of free and clarified consent. The guardians, as well as the child, were free to withdraw from the study at any time.

**Anthropometric Evaluation**

Anthropometric data included the measurement of subjects’ body mass (BM), height, calf circumference, and calf skinfold all of which allowed for the calculation of the body mass index [BMI = body mass (in kilograms)/height² (in meters)] and the calculation of the theoretical cross-sectional area (CSA₀₀) of the calf (24) without calf adiposity.

The body fat percentage (%BF) was obtained by using anthropometric skinfold measurements of the triceps, biceps, subscapular, and suprailiac in accordance with the formula given by Slaughter et al. (43).

**Biomechanical Evaluation**

The biomechanical testing was carried out using a transportable ergometer device (Bio2M, France) that has been described previously (32).

**Experimental protocol.** The child was comfortably placed on the adjustable seat of the ergometer device without a back support, and the thigh and the knee were maintained by a restraint system to keep them immobilized. The right foot was rigidly attached to the adjustable footplate so that the external malleolus coincided with the axis of rotation of the footplate. Therefore the experiments were performed around this assumed axis of joint rotation. The knee was extended to 120° (180° is full extension) and the ankle was flexed to 90°, i.e., the neutral position.

The surface electromyogram (EMG) was detected on the soleus (Sol) using self-adhesive Ag/AgCl surface electrodes (10 mm in diameter, 3M, USA). The electrode impedance was reduced to below 5 kΩ by exfoliating the skin areas with an abrasive sponge and subsequently cleaned with an alcohol pad. The active electrodes (amplified 20X) were placed over the belly of the Sol, ~2 cm below the insertion of the gastrocnemii to the Achilles tendon. The ground electrode was placed over the tibia. The EMG was recorded differentially, DC amplified (amplified 600X), and band-pass filtered (20–500 Hz) (EMG System do Brazil, Brazil). Postacquisition EMG filtering included band-pass filtering (40–300 Hz) and Fast Fourier transform-based peak frequencies elimination to remove unavoidable environmental noise.

First, the maximal motor direct response of the Sol (Sol Mmax, 10 kHz sampling frequency, amplified 100X) was elicited by applying a percutaneous supramaximal electrical stimulation of the posterior tibial nerve with the cathode located in the popliteal fossa and the anode placed on the thigh and proximal to the patella (a custom-made isolated constant current stimulator was used). The stimulus intensity was adjusted to obtain Sol Mmax, and five Sol Mmax responses were measured.

Then, the absolute force was determined under isometric conditions from a maximal voluntary contraction of the triceps surae (MVC, 1 kHz sampling frequency) in plantarflexion. The MVC was defined as the greatest force maintained for at least 500 ms over three attempts. The absolute MVC force was converted to a torque measurement.

Finally, the elastic properties of the triceps surae musculotendinous (MT) complex were assessed by means of the quick-release technique adapted for in vivo experiments (22). The quick-release technique characterizes the SEC stiffness of Hill’s model (27), where a major proportion of the series elasticity resides in the tendon (passive component of the SEC), and the cross bridges constitute the active component of the SEC (30). This heterogeneity of the MT complex leads to nonlinear tension-extension relationships, and the MT stiffness is classically linearly related to the torque (see Fig. 1A). Quick-release movements (4 kHz sampling frequency) around the neutral position were achieved by a sudden release of the footplate, while the child maintained a submaximal voluntary isometric force in the plantarflexion. Three contractions were performed at 25%, 35%, 50%, or 75% of the MVC.

A full test session including rest periods lasted approximately 1 h and was composed of 1) explanation of the test, 2) preparation of the child, 3) familiarization with the test where children were trained to perform the right contraction during which it was ensured by observation that children executed a plantarflexion, and 4) the actual test. The rest periods were standardized in terms of intratess (30 s) and intertest (1 min). The experimental protocol was always performed by the same two investigators. The cumulative time of contraction for all the mechanical tests never exceeded 90 s per subject. Given the resting time after each voluntary contraction (each of them lasting a maximum of 5 s), it can be assumed that there was no muscular fatigue.

**Data processing.** The SEC characteristics were measured over the first 20 ms of the quick-release movement before any reflex changes in muscle activation (e.g., unloading reflex) were possible (4) and when the elastic elements are supposed to recoil. The data processing was conducted on the measured Θ (angular displacement) and the calculated Θ (angular acceleration) obtained by calculating the angular velocity, Θ and its calculated derivatives were filtered using a moving average smoothing technique over 15 points. Therefore MT stiffness is given as the ratio between variations in Θ and Θ:

\[ K = \frac{\Delta\Theta}{\Delta\Theta \cdot t} \]  

(1)

where \( K \) is the MT stiffness and \( t \) is the inertia.

The inertia is easily calculated at the very beginning of the quick release by considering the transition between the static phase and the...
dynamic phase. At this moment, the static torque (T) equals the dynamic torque and acceleration is maximal \(\dot{\Theta}_{\text{max}}\). Then, for each torque, the inertia can be calculated using Newton’s law for angular motion.

\[
T = \dot{\Theta}_{\text{max}} 
\]

Subsequently, the K values were related to the corresponding isometric torque calculated over the 500 ms preceding the quick-release movement. The slope of the linear K-torque relationship was defined as the MT stiffness index of the musculotendinous complex (SlMT-Torque, see Fig. 1A as well) and was proposed to represent changes in the musculotendinous stiffness (31, 32). Furthermore, the MT stiffness index was also calculated from the relationships between K and the corresponding Sol EMG/Mmax and was termed Sl\(\alpha\)-EMG (31).

Finally, the intercept values of the K-torque or K-Sol EMG/Mmax relationships were used to approach a value of the passive stiffness \((K_p\)-Torque or \(K_p\)-EMG), which is similar to the proposition made by Hof (28).

**Alpha Method**

The alpha method was first developed by Morgan (36) and further detailed by Ettema and Huijing (14, 15) and Ettema (16). The main assumption of this method is that the SEC exhibits both a torque-dependent compliance, which is linearly related to the torque that corresponds to the active part of the SEC \((C_{\text{active}})\), as well as a torque-independent compliance in the range of the torque greater than 20% of maximal torque, which corresponds to the passive part of the SEC \((C_{\text{passive}})\). Because the SEC compliance was considered as two springs in series and taking into account the above assumptions, total compliance of the SEC can be described as

\[
C = C_{\text{active}} + C_{\text{passive}} 
\]

or

\[
C \cdot T = \alpha_0 + C_{\text{active}} 
\]

where \(\alpha_0\) represents the elastic extension of the torque-dependent compliance of the SEC. The alpha-torque relationships \((\alpha\)-torque\) were constructed from the K and torque values and from the K and the Sol EMG/Mmax values for each child. In accordance with Fouré (19, 20), who showed that the nonlinearity at low torque between \(\alpha\) and the torque did not influence the result when using the linear approach of the alpha method, \(\alpha_0\)-Torque and \(C_{\text{passive}}\)-Torque as well as \(\alpha_0\)-EMG and \(C_{\text{passive}}\)-EMG were extracted from the linear regression of the \(\alpha\)-torque and the \(\alpha\)-Sol EMG/Mmax relationship, respectively.

Finally, the passive stiffness values were expressed as a ratio value with regard to BM \(\left[\frac{K_p\text{-Torque}}{BM} \frac{1}{C_{\text{passive}}\text{-Torque} \times BM}; \frac{K_p\text{-EMG}}{BM} \frac{1}{C_{\text{passive}}\text{-EMG} \times BM}\right]\) to know whether passive stiffness is greater than would be expected for children of that body mass.

**Commentary**

A preliminary two-way ANOVA analysis (gender \& WHO criteria) with a Bonferroni post hoc test revealed that none of the parameters (anthropometric and biomechanical data) showed any significant influence by gender, except for the \%BF (\(F = 5.25, P < 0.028\)). For the \%BF, the mean values were 18.2 ± 4.0\%, for CON boys, 21.9 ± 5.2\% for CON girls, 28.4 ± 6.3\% for OW boys, and 31.5 ± 2.8\% for OW girls (data are mean ± SD). The Bonferroni post hoc test indicated that all the pairwise multiple comparisons, including the \%BF, did not show significant differences between CON boys and CON girls or OW boys and OW girls (\(P > 0.05\)). Therefore all data for boys and girls were grouped.

**Statistics**

The anthropometric and the biomechanical data were statistically analyzed using a parametric Student t-test for unpaired changes. The statistical analyses also included linear regression analyses to test the K-torque, the \(\alpha\)-torque, the K-Sol EMG/Mmax, and the \(\alpha\)-Sol EMG/Mmax relationships. A level of \(P < 0.05\) was selected to indicate statistical significance. Data are presented as the mean ± SD.

**RESULTS**

**Anthropometric Data**

Table 1 summarizes the mean anthropometric data for the children in the CON and OW group. The BM, BMI, CSAth, and %BF were measured.

| Table 1. Anthropometric data of children of the control (CON) and overweight (OW) group |
| --- | --- | --- | --- |
| n (men/women) | CON 23 (11/12) | OW 21 (11/10) | Gain Value (%) |
| Age (mo) | 107.8 ± 3.10 | 108.5 ± 3.10** | 0.7 |
| Height (cm) | 132.3 ± 5.10 | 134.6 ± 4.90** | 1.8 |
| BM (kg) | 28.6 ± 3.10 | 37.2 ± 5.40** | 29.9 |
| BMI (kg m\(^{-2}\)) | 16.3 ± 0.90 | 20.4 ± 1.80** | 25.2 |
| CSAth (cm\(^{2}\)) | 41.5 ± 5.10 | 48.9 ± 8.90** | 17.9 |
| Body fat (%) | 20.2 ± 4.90 | 29.9 ± 5.10** | 48.3 |
| Inertia (mm\(^{2}\) rad\(^{-1}\)) \(\times 10^{-3}\) | 2.7 ± 0.2 | 2.9 ± 0.30* | 7.1 |

Data are means ± SD. BM, body mass; BMI, body mass index; CSAth, theoretical cross-sectional area (CSAth). * and ** indicates significant differences at \(P < 0.05\) or \(P < 0.01\), respectively. ns is not significant.
and %BF were significantly greater in the OW group (P < 0.001). The greater BM values for the OW children can hold for the significant higher inertia values observed in the OW group. To approach a measure of homogeneity in the population, the BMI coefficient of variation (CV) was also calculated. For the CON group, the BMI CV was 5.7%, and for the OW group, the BMI CV was 9.0%. The WHO reference BMI CV for the present age range and for both genders was, in both cases, greater than 9% (control or overweight).

Maximal Voluntary Contraction

The MVC torque was significantly greater for the children in the OW group. More precisely, the MVC for the children in the CON group was 28.0 ± 7.2 Nm and was 37.7 ± 11.3 Nm for the children in the OW group.

Musculotendonous Stiffness

Individual K-torque relationships for a child of the CON group (r = 0.96; n = 12; P < 0.05) and a child of the OW group (r = 0.92; n = 12; P < 0.05) are given in Fig. 1A and indicate a significant linear increase in the stiffness with torque. For the population, SI\textsubscript{MT-Torque} data were not significantly different between children in the CON and children in the OW group (P > 0.05), while K\textsubscript{p-Torque} values were significantly greater for children in the OW group (P < 0.05; Table 2). Taking into consideration the BM, the K\textsubscript{p-Torque}/BM ratio did not show any significant differences between the OW and CON children (P > 0.05). When the activation capacities were taken into consideration, significant differences in SI\textsubscript{MT-EMG} and K\textsubscript{p-EMG} were found. More precisely, the OW group demonstrated significantly greater values than the CON group for both parameters (P < 0.05 Table 3, Fig. 2A), while K\textsubscript{p-EMG}/BM was not significantly different between the groups.

Alpha Method

Figure 1B shows a significant linear relationship between α and the torque (CON: r = 0.94; n = 12; P < 0.05; OW: r = 0.91; n = 12; P < 0.05) for a child in the CON and OW group (same children as in Fig. 1A), which allowed for the extraction of α\textsubscript{0-Torque} and C\textsubscript{passive-Torque}. For the population, α\textsubscript{0-Torque} was not significantly different between children in the OW group, while C\textsubscript{passive-Torque} was significantly lower in the OW group, compared with CON (P < 0.05, Table 2). Taking into consideration the BM, 1/(C\textsubscript{passive-Torque} * BM) did not show any significant difference between the OW and the CON children (P > 0.05).

DISCUSSION

The present study was devoted to quantifying the elastic properties of the ankle plantar flexors in OW and CON children. As expected, almost all anthropometric data were greater in the OW children. The evolution of the elastic properties of the SEC, however, depended on the variable studied. More precisely, the stiffness of the passive fraction was more important in the OW children, while the stiffness of the active fraction was influenced by the voluntary activation capacities.

Anthropometric Consideration

The use of the BMI is generally accepted for classifying children as normal, overweight, or obese. Based on these indications, the children of the present study were classified as overweight. However, data in the literature and the WHO reference data indicate that the BMI is different between boys and girls who are 9 years of age (CON and OW), which was not found in the present study. A possible reason for this discrepancy is that the BMI growth rates were different between data in the present study and data in the WHO reference values for the present age range. Then, the greater BMI growth rates in the present study can lead to more important variances (SD) in the mean values and therefore can hold for the observed nonsignificant differences in the BMI between boys and girls (CON and OW).

The value of ~30%BF found in the present study is in good agreement with the mean value of males and females of the literature (35), which indicates that children in the OW group can be classified as obese, although this is not in accordance with the classification of overweight with respect to the BMI. It can be questioned whether the children in the OW group must be classified as overweight or obese. Regarding the %BF, there are an increasing number of studies in the literature using different techniques (predictive equations from skinfold measurements, bioimpedance, DXA, air displacement plethysmography), but the results of these techniques do not appear to be conclusive (13, 40, 45). The fact that post hoc tests in the present result did not reveal sex differences can be due to the predictive equation used to estimate the %BF, differences in the %BF growth rates due to the age range studied, and/or the limited number of children in each group. Finally, because the

Table 2. Elastic properties obtained in relation to torque for children of the control (CON) and overweight (OW) group

<table>
<thead>
<tr>
<th>Property</th>
<th>CON 23 (11 / 12)</th>
<th>OW 21 (11 / 10)</th>
<th>Gain Value (%)</th>
</tr>
</thead>
<tbody>
<tr>
<td>SI\textsubscript{MT-Torque} (rad\textsuperscript{-1})</td>
<td>2.6 ± 0.7</td>
<td>2.6 ± 0.9\textsuperscript{**}</td>
<td>0.01</td>
</tr>
<tr>
<td>K\textsubscript{p-Torque} (Nm rad\textsuperscript{-1})</td>
<td>42.9 ± 13.3</td>
<td>52.3 ± 16.8\textsuperscript{*}</td>
<td>21.8</td>
</tr>
<tr>
<td>K\textsubscript{p-Torque} / BM (NM rad\textsuperscript{-1} kg\textsuperscript{-1})</td>
<td>1.5 ± 0.4</td>
<td>1.4 ± 0.5\textsuperscript{**}</td>
<td>-4.4</td>
</tr>
<tr>
<td>α\textsubscript{0-Torque} (rad) × 10\textsuperscript{-2}</td>
<td>7.8 ± 2.1</td>
<td>8.1 ± 1.8\textsuperscript{*}</td>
<td>3.6</td>
</tr>
<tr>
<td>C\textsubscript{passive-Torque} (rad Nm\textsuperscript{-1}) × 10\textsuperscript{-3}</td>
<td>7.1 ± 1.4</td>
<td>5.8 ± 1.1\textsuperscript{**}</td>
<td>-19.2</td>
</tr>
<tr>
<td>1 / (C\textsubscript{passive-Torque} * BM) (rad\textsuperscript{-1} Nm kg\textsuperscript{-1})</td>
<td>5.1 ± 1.1</td>
<td>4.9 ± 0.9\textsuperscript{**}</td>
<td>-4.8</td>
</tr>
</tbody>
</table>

Data are means ± SD. Musculotendonous stiffness index (SI\textsubscript{MT-Torque}), passive stiffness (K\textsubscript{p-Torque}), active compliance (α\textsubscript{0-Torque}), passive compliance (C\textsubscript{passive-Torque}), body mass (BM), * and ** indicates significant differences at P < 0.05 or P < 0.01, respectively, ns is not significant.
A higher slope value and intercept point, indicating higher stiffness in OW and overweight (OW) (circle, solid line) group. Descriptive analysis indicates overweight (OW) group can be attributed to differences in the mass of the children’s foot. Thus the SI_{MT-torque} values are somewhat overestimated in the OW group due to that calculation of the MT stiffness (see Eq. 1). This can be easily verified by dividing SI_{MT-torque} by the inertia. On average, a nonsignificant decrease of 7.4% ($P > 0.05$) in SI_{MT-torque/inertia} can be observed in the OW group such that the inertia did not statistically influence the calculation of that MT stiffness, i.e., $\Delta\ddot{\theta}/\Delta\theta$. As a matter of fact, the variance (SD) in SI_{MT-torque} was more important than the variance (SD) in the inertia.

### Elastic Properties of the Musculotendinous Unit Under Passive Conditions

The elastic properties of the musculotendinous unit under passive conditions were assessed using both the $K_p$-Torque (or $K_p$-EMG) data as well as the $C_{passive}$-Torque (or $C_{passive}$-EMG) data. When $C_{passive}$-Torque is expressed as stiffness, one can observe that its value is more than three times greater than $K_p$-Torque, which is obtained as the intercept point of the MT stiffness-torque relationship in all groups tested. This brings up the question of which structure is involved in the measurement or the use of the modified Hill model. Hof (28), using the controlled fast release technique in the human ankle, suggested that the passive stiffness, obtained from linear K-torque relationship, can be attributed to an additional elastic component in series with a parallel elastic component (PEC) of a modified Hill model (28). More precisely, Hof (28) suggested that this additional passive stiffness is always engaged and is similar to the proposition of Shorten (42). Because the quick-release technique imposes a joint rotation, it is possible that the passive joint structures, e.g., ligaments, influence the $K_p$-Torque (or $K_p$-EMG) values, while $C_{passive}$-Torque (or $C_{passive}$-EMG) represents the compliance of tendon structures. Furthermore, the $K_p$-Torque values in the present study are comparable to the data describing the passive musculoskeletal stiffness obtained from sinusoidal perturbations in prepubertal children (23).

The present study reported a significantly greater passive stiffness ($K_p$-Torque, $K_p$-EMG) and lower compliance values ($C_{passive}$-Torque, $C_{passive}$-EMG) of children in the OW group. These differences reflect an adaptive response in a similar way, i.e., higher stiffness in both the SEC passive fraction and in the articular structures. Because the OW children should have lower daily activity patterns (29), a greater body mass related loading of the tendon structures should influence the tendon adaptation (47). Therefore the present results are in favor of an adaptive response of the passive stiffness due to the overweight-influenced gravity-dependent extra load of the tendinous structures. This is supported by the invariance in

### Table 3. Elastic properties obtained in relation to voluntary activation (EMG) for children of the control (CON) and overweight (OW) group

<table>
<thead>
<tr>
<th>n (men /women)</th>
<th>CON 23 (11 / 12)</th>
<th>OW 17 (9 / 8)</th>
<th>Gain Value (%)</th>
</tr>
</thead>
<tbody>
<tr>
<td>SI_{MT-EMG} (Nm rad^{-1} %^{-1})</td>
<td>46.4 ± 23.8</td>
<td>80.1 ± 53.8*</td>
<td>72.6</td>
</tr>
<tr>
<td>$K_p$-EMG (Nm rad^{-1})</td>
<td>47.1 ± 14.1</td>
<td>59.2 ± 16.6*</td>
<td>25.6</td>
</tr>
<tr>
<td>$K_p$-EMG / BM (Nm rad^{-1} kg^{-1})</td>
<td>1.6 ± 0.4</td>
<td>1.6 ± 0.6</td>
<td>-1.4</td>
</tr>
<tr>
<td>$\alpha_{0-EMG}$ (rad) $\times 10^{-3}$</td>
<td>4.1 ± 2.9</td>
<td>2.3 ± 1.7*</td>
<td>-43.5</td>
</tr>
<tr>
<td>$C_{passive}$-EMG (rad %^{-1}) $\times 10^{-3}$</td>
<td>7.8 ± 1.3</td>
<td>6.7 ± 1.7*</td>
<td>-14.7</td>
</tr>
<tr>
<td>$1/(C_{passive}$-EMG + BM) (rad %^{-1} kg^{-1})</td>
<td>4.6 ± 0.8</td>
<td>4.3 ± 1.5*</td>
<td>-6.0</td>
</tr>
</tbody>
</table>

Data are means ± SD. Musculotendinous stiffness index (SI_{MT-EMG}), passive stiffness ($K_p$-EMG), active compliance ($\alpha_{0-EMG}$), passive compliance ($C_{passive}$-EMG). * indicates significant differences at $P < 0.05$.

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**Fig. 2.** A: Musculotendinous (MT) stiffness-Sol EMG/Mmax and B: $\alpha$-Sol EMG/Mmax relationships for a child of the control (CON) (square, dotted line) and overweight (OW) (circle, solid line) group. Descriptive analysis indicates higher slope values and intercepts points, indicating higher stiffness in OW (A). In B the intercept value, representing $\alpha_0$, and the slope, representing passive compliance, are lower for OW.
1/(C_{\text{passive-Torque}} \times \text{BM})$ between the CON and the OW children, which is in agreement with the recent result of Waugh et al. (47) who reported that the body mass was the best predictor of tendon stiffness.

**Elastic Properties of the Musculotendinous Unit Under Active Conditions**

The literature indicates that the normalized MT stiffness at 30% of MVC is not different between lean, preobese, or obese postmenopausal women (17), which is in agreement with the nonsignificant difference in $S_{\text{MT-Torque}}$ between the CON and the OW children. However, when taking into consideration the activation capacities, greater $S_{\text{MT-EMG}}$ values are found in the OW children. Indeed, the MT stiffness quantification during submaximal voluntary activation is influenced by the differences in the voluntary activation capacities, as proposed by Lambertz et al. (31) when studying prepubertal children aged 7 to 10 years. In this latter study, a greater MT stiffness index in young prepubertal children was influenced by the lower neuromuscular efficiency and greater coactivation index (31) in the youngest children because coactivation of the antagonist muscles acts in parallel and should add to the quantification of the MT stiffness. Therefore the invariance in $S_{\text{MT-Torque}}$ or $S_{\text{MT-EMG}}$ values can be due to a lower neuromuscular efficiency of the children in the CON group.

To compare the stiffness adaptations due to intrinsic changes in the SEC (i.e., cross bridges and tendon), it is important to take into consideration the voluntary activation capacities (31). The increase in $S_{\text{MT-EMG}}$ and the decrease in $\alpha_{\text{EMG}}$ in the OW children highlighted an intrinsic adaptation in both the active portion (i.e., cross bridges) and the passive portion (i.e., tendon) of the SEC.

The literature indicate that when a training technique increases the slow-type fiber proportion, the SEC stiffness will increase (21), while the opposite result is obtained when the fast-type fiber proportion increased, i.e., greater SEC compliance (3). Therefore the overweight-induced increase in $S_{\text{MT-EMG}}$ or lower $\alpha_{\text{EMG}}$ can be due to a higher content of slow-type fibers. This is in agreement with the greater slow-type fiber content obtained from isolated soleus muscle of transgenetic obesity in humans (44), but this result is in opposition to the reported higher proportion of fast-type fibers in the rectus abdominis of humans (44). These opposing results in these two above-mentioned studies could be related to the differences in the function of the two muscles studied. In effect, the soleus is a postural muscle, which is not the case for the rectus abdominis. Therefore the overweight-induced gravity-dependent extra load of the musculotendinous unit and the corresponding intrinsic signaling pathway could possibly be the reason for a greater proportion of slow-type fibers in the triceps surae muscle in the OW children.

Furthermore, Hof (28) suggested that the increase in the MT stiffness with torque can also result from the parallel addition of passive structures, specifically the active addition of aponeuroses. The present study extends this knowledge because 1) changes in the MT stiffness index could only be revealed when voluntary activation was taken into consideration and 2) important gain changes are obtained in $S_{\text{MT-EMG}}$ (70%) or $S_{\text{EMG}}$ (−43%). It is proposed that both active recruitment of muscle fibers as well as parallel active addition of aponeuroses during contraction contributed to the MT stiffness quantification under voluntary conditions using the quick-release method.

**Functional Consequences**

The SEC active compliance ($\alpha_{\text{EMG}}$) and the SEC passive compliance ($C_{\text{passive-Torque}}$ and $C_{\text{passive-EMG}}$) were less in the OW children with a subsequent increase in the MVC torque because the measure of the generated force was performed at the periphery/joint.

Such an adaptive response in the musculotendinous elastic properties of the triceps surae can have functional consequences in terms of stretch-shortening cycles and movement control in the OW children. The musculotendinous complex contributes to the efficiency during the stretch-shortening cycle but has to fulfill two contradictory requirements: it must be compliant for elastic energy storage and stiff for the subsequent transmission of the generated force (11).

The assessment of the maximal length changes of the musculotendinous complex at MVC from the linear interpolation of the K-torque relationship ($\Delta \Theta_{\text{MVC}} = \text{MVC} / K_{\text{MVC}}$) showed no differences between the groups ($\Delta \Theta_{\text{MVC}}$ – OW: 0.26 ± 0.06 rad and $\Delta \Theta_{\text{MVC}}$ – CON: 0.25 ± 0.03 rad). Taking into account this equality, it can be suggested that the greater MVC in the OW children causes a greater elastic energy storage capacity at maximal strength in these children. This result is confirmed by the greater absolute peak power in obese children reported by Lazzer et al. (34) because no significant differences were found in the peak velocity (34). Then, the greater force generation capacities should affect the stretch-shortening cycle in the OW children as well as the biomechanics of walking. Indeed, it has been reported (39) that a greater net metabolic costs of walking in obese adolescents can be due to greater step-to-step costs associated with wide gait during walking, which compromises the biomechanics of walking in obese subjects. A possible explanation to this is that greater muscle stiffness has important functional consequences in terms of movement control. For instance, the force control should be more difficult because, for a given variation in the displacement, the forces that must be generated should be more prominent to perform a controlled movement (33).

Furthermore, the functional consequences should also concern the evolution of the passive elastic properties and should be taken into account because 1) these characteristics contribute, in part, to the maximal joint range of motion, 2) part of the force developed by the contracting muscle should be devoted to the stretch of the passive antagonist, and 3) a relation between the passive stiffness and 4) the spindle discharge has been reported (41).

**CONCLUSIONS**

The present study demonstrated that weight-related additional loading of the musculotendinous unit results in greater passive stiffness, while adaptation of the stiffness during active conditions was revealed when the activation capacities were taken into account. It is concluded that the adaptive response of the musculotendinous unit due to overweight is influenced by extrinsic pathways, which is due to a relative overweight-
related extra load of the musculotendinous stiffness and intrinsic pathways, which are likely due to a fiber-type transition phenomenon.

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DISCLOSURES

No conflicts of interest, financial or otherwise, are declared by the author(s).

AUTHOR CONTRIBUTIONS

Author contributions: D.L. and K.M.F. conception and design of research; D.L., T.O.S., F.C., and K.M.F. interpreted results of experiments; D.L. prepared manuscript; D.L., F.C., and K.M.F. edited and revised manuscript; D.L., T.O.S., F.C., and K.M.F. approved final version of manuscript; T.O.S. and L.C.X. performed experiments.

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