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Evaluation of different analytical methods for subject-specific scaling of musculotendon parameters

C.R. Winby[a,b], D.G. Lloyd[a,*], T.B. Kirk[b]

[a] School of Sport Science, Exercise and Health, University of Western Australia, 35 Stirling Highway, Crawley, WA 6907, Australia
[b] School of Mechanical Engineering, University of Western Australia, 35 Stirling Highway, Crawley, WA, Australia

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Abstract

Musculoskeletal models are often used to estimate internal muscle forces and the effects of those forces on the development of human movement. The Hill-type muscle model is an important component of many of these models, yet it requires specific knowledge of several muscle and tendon properties. These include the optimal muscle fibre length, the length at which the muscle can generate maximum force, and the tendon slack length, the length at which the tendon starts to generate a resistive force to stretch. Both of these parameters greatly influence the force-generating behaviour of a musculotendon unit and vary with the size of the person. However, these are difficult to measure directly and are often estimated using the results of cadaver studies, which do not account for differences in subject size. This paper examined several different techniques that can be used to scale the optimal muscle fibre length and tendon slack length of a musculotendon unit according to subject size. The techniques were divided into three categories corresponding to linear scaling, scaling by maintaining a constant tendon slack length throughout the range of joint motion, and scaling by maintaining muscle operating range throughout the range of joint motion. We suggest that a good rationale for scaling muscle properties should be to maintain the same force-generating characteristics of a musculotendon unit for all subjects, which is best achieved by scaling that preserves the muscle operating range when the muscle is maximally activated.

Keywords: Scaling; Muscle models; Optimal fibre length; Tendon slack length

1. Introduction

Measuring internal loads in the moving human body is difficult, which has driven the development of musculoskeletal models that estimate the forces generated by each muscle-tendon unit (MTU) (Lloyd and Besier, 2003; Olney and Winter, 1985; Pandy and Anderson, 2000). In these, each muscle is often represented as a normalised Hill-type muscle model (Hill, 1938) where the muscle's normalised isometric force ($F_{\mathrm{m}}^{\mathrm{norm}}$) output depends on its (a) optimal fibre length ($L_{\mathrm{m}}^{\mathrm{O}}$), the fibre length where maximum active force is generated, (b) tendon slack length ($L_{\mathrm{T}}^{\mathrm{S}}$), the tendon length at which it first resists stretch, and (c) muscle pennation angle α (or $\alpha_{\mathrm{m}}^{\mathrm{O}}$ at $L_{\mathrm{m}}^{\mathrm{O}}$), the angle between the muscle fibres and the tendon (Zajac, 1989). In these models, for a given activation level, $F_{\mathrm{m}}^{\mathrm{norm}}$ is characterised by normalised active and passive length–tension relationships (Fig. 1) that are functions of the muscle's normalised fibre length $L_{\mathrm{m}}^{\mathrm{norm}}$, its actual fibre length divided by $L_{\mathrm{m}}^{\mathrm{O}}$. When a muscle has $L_{\mathrm{m}}^{\mathrm{norm}}$ within its force-generating range, to transmit force to a segment via the tendon, the tendon must be strained beyond its $L_{\mathrm{T}}^{\mathrm{S}}$ such that there is a force balance between the tendon and muscle. Because of this force balance different values of $L_{\mathrm{T}}^{\mathrm{S}}$ and $L_{\mathrm{m}}^{\mathrm{O}}$ will shift the overall force-generating range of the MTU.

While muscles can nominally generate force throughout the range $0.5 \leqslant L_{\mathrm{m}}^{\mathrm{norm}} \leqslant 1.5$ (Zajac, 1989), the actual operating range is muscle dependent (Arnold et al., 2007). The $L_{\mathrm{m}}^{\mathrm{norm}}$ adopted by a muscle with given $L_{\mathrm{m}}^{\mathrm{O}}$, $L_{\mathrm{T}}^{\mathrm{S}}$ and $\alpha_{\mathrm{m}}^{\mathrm{O}}$ will change with the posture(s) of the joint(s) it crosses. Let $L_{\mathrm{m}}^{\mathrm{norm}}(\mathbf{q})$ denote the relationship between $L_{\mathrm{m}}^{\mathrm{norm}}$ and the range of joint angles, $\mathbf{q}$, for a maximally activated, isometrically contracting muscle. In musculoskeletal modelling, using the correct

*Corresponding author. Tel.: +61 8 6488 3919; fax: +61 8 6488 1039.
E-mail address: dlloyd@cyllene.uwa.edu.au (D.G. Lloyd).

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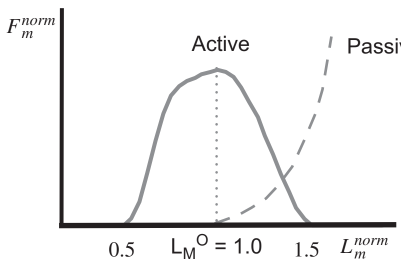

Fig. 1. Relationship between normalised muscle force $F_{\mathrm{m}}^{\mathrm{norm}}$ and normalised muscle fibre length $L_{\mathrm{m}}^{\mathrm{norm}}$.

$L_{\mathrm{m}}^{\mathrm{norm}}(\mathbf{q})$ is important as it influences the MTU's generated force (Lloyd and Besier, 2003; Heine et al., 2003; Garner and Pandy, 2003), with $L_{\mathrm{m}}^{\mathrm{norm}}(\mathbf{q})$ being very sensitive to the values of $L_{\mathrm{m}}^{\mathrm{O}}$ and $L_{\mathrm{T}}^{\mathrm{S}}$ (Heine et al., 2003; Out et al., 1996; Redl et al., 2007).

It is difficult to measure $L_{\mathrm{m}}^{\mathrm{O}}$ and $L_{\mathrm{T}}^{\mathrm{S}}$ (Herzog et al., 1991a), even using costly medical imaging technologies such as ultrasound and MRI (Blemker et al., 2007); so generalised parameters obtained from cadaver studies are often used (Hoy et al., 1990; Lloyd and Buchanan, 1996; Olney and Winter, 1985). However, not all subjects are identical (Abe et al., 2000; Blazevich et al., 2006); so parameters should be scaled to the subject (Blemker et al., 2007). Subject-specific scaling may be viewed as having two components: (1) anthropometric scaling of $L_{\mathrm{m}}^{\mathrm{O}}$ and $L_{\mathrm{T}}^{\mathrm{S}}$ to each subject's segment and MTU lengths, and (2) functional scaling of $L_{\mathrm{m}}^{\mathrm{O}}$ and $L_{\mathrm{T}}^{\mathrm{S}}$ to reflect each subject's specific moment-generating characteristics (Garner and Pandy, 2003; Lloyd and Besier, 2003). Software packages such as SIMM (Software for Interactive Musculoskeletal Modeling–MusculoGraphics Inc., Chicago, USA) or AnyBody (AnyBody Technology, Aalborg, Denmark) facilitate anthropometric scaling that adjusts MTU lengths and moment arms using bone dimensions (Loan et al., 2005), the latter being relatively easy to estimate. However, $L_{\mathrm{m}}^{\mathrm{O}}$ and $L_{\mathrm{T}}^{\mathrm{S}}$ still need to be adjusted and these do not scale linearly with bone length (Ward et al., 2005).

Recent work has sought to identify physiological rationales on which to base scaling techniques. One strategy calculated $L_{\mathrm{T}}^{\mathrm{S}}$ of an individual MTU given *a priori* specification of $L_{\mathrm{m}}^{\mathrm{O}}$ (Manal and Buchanan, 2004). $L_{\mathrm{m}}^{\mathrm{O}}$ assumed the unscaled value from SIMM (Delp et al., 1990), while $L_{\mathrm{T}}^{\mathrm{S}}$ was adjusted in an optimisation process to achieve a muscle–tendon force balance at three different joint postures representing the lower, middle and upper ranges of muscle excursion respectively. Neither passive muscle forces nor low tendon strains were accounted for, and in cases of multiple solutions for $L_{\mathrm{T}}^{\mathrm{S}}$ an average was adopted that may not have been the correct physiological solution. Since this method assumes a value for $L_{\mathrm{m}}^{\mathrm{O}}$ it provides no rationale for its scaling.

Another approach has been presented that, in theory, may address both components of subject-specific scaling, and adjusts $L_{\mathrm{m}}^{\mathrm{O}}$ and $L_{\mathrm{T}}^{\mathrm{S}}$ (Garner and Pandy, 2003). $L_{\mathrm{m}}^{\mathrm{O}}$ and $L_{\mathrm{T}}^{\mathrm{S}}$ were initially adjusted for each muscle such that $L_{\mathrm{m}}^{\mathrm{norm}}(\mathbf{q})$ ranged from 0.5 to 1.2. These $L_{\mathrm{m}}^{\mathrm{norm}}(\mathbf{q})$ bounds were then adjusted within the range of 0.1–1.6 in an optimisation process that minimised the difference between a set of experimentally measured and modelled isometric joint moments. However, even though $L_{\mathrm{m}}^{\mathrm{norm}}(\mathbf{q})$ bounds are different between muscles (Arnold et al., 2007), all muscles had the same starting bounds, which may have influenced the final solutions in the optimisation process. Additionally, the method was only applied to a generic anatomical model. Functional scaling may also require more functional trials to encapsulate other operating conditions, such as eccentric and concentric contractions and other movements (Buchanan et al., 2005; Lloyd and Besier, 2003). So there is value in separating anthropometric and functional scaling. Yet importantly, this method did recognise the influence of $L_{\mathrm{m}}^{\mathrm{norm}}(\mathbf{q})$ on the force-generating properties of an MTU. Preserving $L_{\mathrm{m}}^{\mathrm{norm}}(\mathbf{q})$ between unscaled and anthropometrically scaled subjects is advantageous in that it guarantees a consistency in muscle function which can be used to perform forward dynamic modelling as an 'average scaled model' or as a platform for further functional calibration.

In summary, techniques are required to anthropometrically scale $L_{\mathrm{m}}^{\mathrm{O}}$ and $L_{\mathrm{T}}^{\mathrm{S}}$ to the size of an individual, but which are best? This paper evaluates the effectiveness of seven different techniques for anthropometric scaling of $L_{\mathrm{m}}^{\mathrm{O}}$ and $L_{\mathrm{T}}^{\mathrm{S}}$, with particular reference to a new method that preserves $L_{\mathrm{m}}^{\mathrm{norm}}(\mathbf{q})$ from an unscaled to a scaled musculoskeletal model.

2. Methods

All experimental work was approved by the University of Western Australia Human Research Ethics Committee and all participants gave their informed, written consent prior to testing. Testing involved capturing the static poses of ten subjects in the anatomical position using a 7-camera Vicon Mx motion capture system (ViconPeak, Oxford, UK). These poses were modelled (Besier et al., 2003) to obtain distances between landmarks relating to the pelvis: left and right anterior superior iliac spines, femur: hip and knee joint centres, tibia: knee and ankle joint centres, and foot: calcaneus to 1st metatarsal. Segment scaling factors were obtained by dividing these measurements by those corresponding to a generic baseline lower body anatomical model (Delp et al., 1990) which consisted of the attachment points and paths of 13 MTUs crossing the knee joint as well as their $L_{\mathrm{m}}^{\mathrm{O}}$ and $L_{\mathrm{T}}^{\mathrm{S}}$, and $\alpha_{\mathrm{m}}^{\mathrm{O}}$. An additional "virtual subject" was generated with all segment sizes increased by 10 percent. Using the SIMM software package, scaling factors were then applied to the segments and corresponding MTU attachment and "via" points of the generic model, resulting in a subject-specific anatomical model.

Each MTU was categorised into one of three groups based on the segments it attached to and traversed. These groups were uniarticular spanning the femur and tibia (UA_1), biarticular spanning the pelvis, femur and tibia (BA_1), and biarticular spanning the femur, tibia, and foot (BA_2). For each subject, a scaling uniformity index was created for each MTU group by summing the absolute values of the differences between scaling factors of adjacent segments. For example, given scaling factors of 1.05 for the pelvis, 0.97 for the femur, and 1.08 for the tibia, the calculation for BA_1 would be $|(1.05–0.97)|+|(0.97–1.08)|=0.19$. The average of these three group indices for each subject was termed their proportionality index (PI). The greater the PI, the less uniform the scaling factors of the body segments. The minimum value for the PI was zero, corresponding to equal scaling factors for each segment as for the virtual subject.

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The values of L_m^O and L_T^S from the baseline model were scaled to the subject-specific model using seven different techniques. All techniques only relied on changes in segment lengths, but used this information in different ways that facilitated their division into three categories, now discussed in detail.

Category 1 consisted of two linear scaling methods, LinScal-1 and LinScal-2, which assumed that L_m^O and L_T^S scale linearly by equal amounts according to subject size. LinScal-1 was the simplest method and scaled L_m^O and L_T^S of the thigh muscles by the femur scaling factor and shank muscles by the tibia scaling factor. This method has been shown to be invalid (Ward et al., 2005); so the results were assumed to be a 'worst case scenario' for comparison purposes. In LinScal-2 L_m^O and L_T^S were scaled by the relative length change of their MTU from the generic to scaled anatomical model and differs from LinScal-1 by implicitly incorporating the scaling of all segments relevant to the MTU.

Category 2 methods (ConstTSL-1 and ConstTSL-2) were based on the technique presented by Manal and Buchanan (2004). For each MTU the musculotendon length (L_{MT}) was determined at three different postures, and the corresponding L_m^{norm} adjusted in an optimisation process that minimised the difference between the subsequent L_T^S calculated at each posture (Eq. (1)). In this equation α denotes muscle pennation angle at L_m^{norm} and was calculated assuming constant muscle thickness from α_m^O, which was assumed to be the same for a given MTU regardless of subject size. The final L_T^S was that which varied the least across the three postures, and in cases of multiple solutions an average was taken. In these methods L_m^O for each muscle was determined using the LinScal-2 method. ConstTSL-1 calculated L_T^S according to Manal and Buchanan (2004) (Eq. (2)), while ConstTSL-2 accounted for (a) the passive forces neglected by ConstTSL-1 in its generic muscle fibre–force length relationship, and (b) the non-linearity of the tendon stress/strain relationship at low strain, thereby eliminating the potential for non-zero tendon strains at zero tendon force and subsequent shift of solution L_T^S and resultant $L_m^{norm}(\mathbf{q})$. This non-linear region used an exponential curve passing through point (0,0) and maintaining continuity with the zeroth and first derivatives of the linear component of the curve at tendon strain (ε_T) = 0.0127 (Zajac, 1989) (Eq. (3)):

$$L_T^S = \frac{L_{MT} - L_m^o L_m^{norm} \cos\alpha}{1+\varepsilon_T} = \text{constant} \tag{1}$$

and

$$\varepsilon_T = \frac{F_m^{norm}\cos\alpha + 0.2375}{37.5} \quad \text{for } \varepsilon_T \geqslant 0 \tag{2}$$

OR

$$\varepsilon_T = \frac{F_m^{norm}\cos\alpha + 0.2375}{37.5} \quad \text{for } \varepsilon_T > 0.0127$$

$$\varepsilon_T = \frac{\ln\left(\frac{F_m^{norm}\cos\alpha}{0.06142} + 1\right)}{124.929} \quad \text{for } \varepsilon_T \leqslant 0.0127 \tag{3}$$

In Category 3 three methods that preserved $L_m^{norm}(\mathbf{q})$ between the unscaled and scaled model were assessed: PresMusOper-1, PresMusOper-2, and PresMusOper-3. In these the relationship between F_m^{norm} and joint angle was preserved since L_m^{norm} is the only determinant of F_m^{norm} during a maximum isometric contraction. PresMusOper-1 constrained $L_m^{norm}(\mathbf{q})$ between scaled and unscaled models by using values of L_m^{norm} from two joint postures $P1$ and $P2$ (Table 1) that represented the upper and lower limits of excursion range for all MTUs. For the biarticular muscles simultaneous manipulation of both joints was required (Table 1). In PresMusOper-1, the scaled L_m^{norm} and α were set equal to the unscaled values, i.e. ${L_{m(P1)}}^{norm}$, ${L_{m(P2)}}^{norm}$, $\alpha_{(P1)}$ and $\alpha_{(P2)}$ were equal in the unscaled and scaled model, which permitted a unique L_m^O to be determined:

$$L_M^O = \frac{S_{(P1)}L_{MT(P2)} - S_{(P2)}L_{MT(P1)}}{S_{(P1)}L_{m(P2)}^{norm}\cos\alpha_{(P2)} - S_{(P2)}L_{m(P1)}^{norm}\cos\alpha_{(P1)}} \tag{4}$$

where $S = 1+\varepsilon_T$ and ε_T is given by Eq. (3).

$L_{MT(P1)}$ and $L_{MT(P2)}$ were the MTU lengths from the scaled model. Once L_m^O for the scaled MTU were known they were substituted into Eqs. (1) and (3) to solve for the corresponding L_T^S.

PresMusOper-2 was identical to PresMusOper-1 except that it used a different set of postural pairs to investigate sensitivity of the resulting scaled L_m^O and L_T^S to the postural pairs adopted. The hip and ankle were set in the anatomical zero position, with the knee angle set to 0° and 90° for postures $P1$ and $P2$, respectively.

PresMusOper-3 was designed to preserve $L_m^{norm}(\mathbf{q})$ across the entire range of motion. Eleven evenly spaced posture control points ($\overline{PC}$) were selected that spanned the entire range of excursion of each MTU. For each MTU in the unscaled model the corresponding vectors $\overline{L_{m(PC)}^{norm}}$ and $\bar{\alpha}_{(PC)}$ were determined for $\overline{PC}$. Similarly to PresMusOper-1, $\overline{L_{m(PC)}^{norm}}$ and $\bar{\alpha}_{(PC)}$ were assumed equal in the scaled and unscaled models. For the scaled model, the vector $\overline{L_{MT(PC)}}$ corresponding to $\overline{PC}$ was determined. Given this information, it was possible to establish the following optimisation procedure to solve for L_m^O and L_T^S for each MTU

$$\min \sum_{i=1}^{11} (L_{MT(PC)}(i) - L_{MT(Predicted)}(i))^2 \tag{5}$$

where $L_{MT(Predicted)}(i) = L_T^S + L_T^S\varepsilon_T(i) + L_m^O L_{m(PC)}^{norm}(i)\cos\alpha_{PC}(i)$ and ε_T is given by Eq. (3).

In this the scaled L_m^O and L_T^S were adjusted to minimise the least squares difference between the measured $L_{MT(PC)}$ determined from the scaled model and the $L_{MT(Predicted)}$ calculated at the same points using $\overline{L_{m(PC)}^{norm}}$ and $\bar{\alpha}_{(PC)}$.

The seven scaling methods were evaluated according to how well these preserved each muscle's $L_m^{norm}(\mathbf{q})$ between the scaled and unscaled model. The scaled and unscaled $L_m^{norm}(\mathbf{q})$ were evaluated for each muscle at 2° increments of knee flexion between 0° and 90° with the hip and ankle set in the anatomical zero position. For the biarticular muscles this was repeated with the hip and ankle fixed at 74° flexion and 44° extension, respectively, to encompass the entire MTU excursion range. The sum of squared differences between the scaled and unscaled $L_m^{norm}(\mathbf{q})$ were averaged to yield a mean sum of squared difference (MSSD) for each muscle group.

Two comparisons were performed that examined techniques (1) within and (2) between scaling categories. Both comparisons utilised a two-way repeated measures ANOVA with the log transform of the MSSD the dependent variable and muscle group and either scaling technique (within category) or scaling category (between category) the independent variables, with Bonferroni corrections used for post hoc comparisons. Greenhouse–Geisser corrections were applied as necessary to account for non-sphericity. For the between category analysis one technique from each category was chosen: ConstTSL-2 and PresMusOper-3 because these returned the lowest MSSDs, and LinScal-2 as this method is employed by SIMM.

For each subject, a linear correlation was taken between the MSSD of LinScal-2, ConstTSL-2 and PresMusOper-3 and PI in order to examine the effect of PI on the capacity of these methods to preserve $L_m^{norm}(\mathbf{q})$.

Table 1
Postural positions selected to define the range of excursion of the three different groups of MTUs for PresMusOper-1

Muscle group	Posture $P1$	Posture $P2$
UA1	Knee 16°	Knee 90°
BA1	Hip 0°, knee 16°	Hip 74°, knee 90°
BA2	Knee 0°, ankle 0°	Knee 64°, ankle −44°

3. Results

Mean subject height was 177.5±9 cm, ASIS distance 26.0±2.1 cm, femur length 41.8±3.1 cm, tibia length 41.0±2.2 cm, and foot length 29.0±1.1 cm. PI ranged from 0.015 to 0.21 with a mean of 0.087±0.069. MTU length only scaled uniformly across the full knee joint angle range when PI was zero (Fig. 2).

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For the virtual subject with a PI of zero all techniques except the ConstTSL methods generated identical L_m^O and L_T^S (Table 2), whereas the non-zero PI solutions varied considerably, with greater similarities within the categories than between. For example, for the largest PI of 0.210, values of L_m^O for the medial gastrocnemius ranged from 0.040 (LinScal) to 0.047 (PresMusOper), with L_T^S ranging from 0.368 (PresMusOper) to 0.390 (ConstTSL). PI was highly correlated with the MSSD of LinScal-2 ($R^2 = 0.870$, slope = 0.376), less correlated with that of PresMusOper-3 ($R^2 = 0.474$, slope = 0.0394), and not correlated with that of ConstTSL-2 ($R^2 = 2e^{-6}$, slope = 0.0005).

The within category analysis revealed no significant difference between the MSSD of the LinScal methods (Fig. 4). ConstTSL-2 yielded significantly lower MSSD than ConstTSL-1 for the BA group muscles, but larger MSSD for the UA group muscles ($p = 0.001$). PresMusOper-1 yielded smaller MSSD than PresMusOper-2 ($p<0.001$), yet greater MSSD than PresMusOper-3 ($p<0.001$). However, PresMusOper-2 in some cases provided a better fit within the limited range of motion bounded by its constraint points (Fig. 3). For all categories MSSD was significantly greater for the BA_1 (LinScal: $p = 0.039$, ConstTSL: $p<0.001$, PresMusOper: $p = 0.002$) and BA_2 (LinScal: $p = 0.041$, ConstTSL: $p<0.001$,

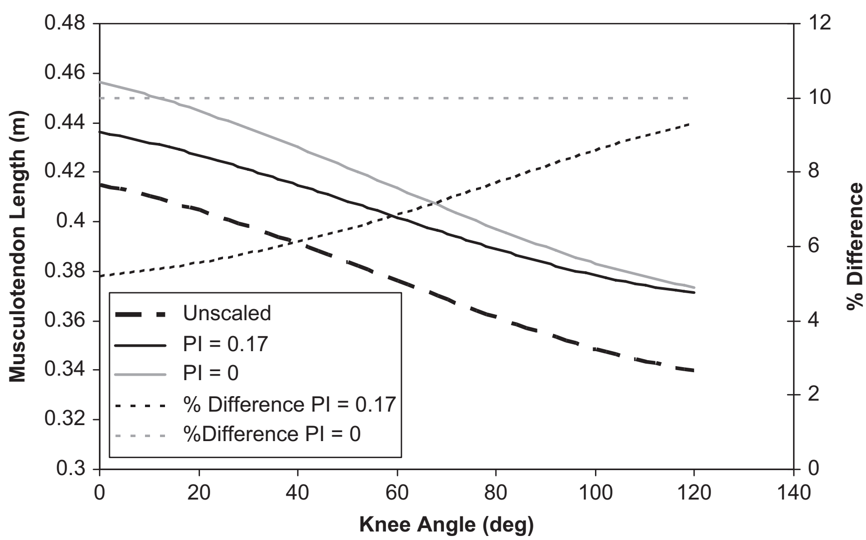

Fig. 2. Musculotendon length changes with knee joint angle for the semimembranous muscle for two subjects with different PIs compared to the unscaled model. For the subject with PI = 0, the % difference is constant and equal to the scaling factor throughout the entire range of motion of the joint. For the subject with PI = 0.17, the % difference varies throughout the range of motion.

Table 2
Sample solutions for L_m^O and L_T^S generated by the seven different techniques for two subjects of different PIs and three muscles corresponding to each of the major muscle groups (UA_1, BA_1 and BA_2)

	Technique	Biceps femoris short head (UA_1)		Rectus femoris (BA_1)		Medial gastrocnemius (BA_2)	
		L_m^O	L_T^S	L_m^O	L_T^S	L_m^O	L_T^S
PI = 0	LinScal-1	0.190	0.110	0.088	0.352	0.044	0.418
	LinScal-2	0.190	0.110	0.088	0.352	0.044	0.418
	ConstTSL-1	0.190	0.125	0.088	0.398	0.044	0.432
	ConstTSL-2	0.190	0.140	0.088	0.373	0.044	0.427
	PresMusOper-1	0.190	0.110	0.088	0.352	0.044	0.418
	PresMusOper-2	0.190	0.110	0.088	0.352	0.044	0.418
	PresMusOper-3	0.190	0.110	0.088	0.352	0.044	0.418
PI = 0.21	LinScal-1	0.191	0.110	0.092	0.370	0.040	0.378
	LinScal-2	0.199	0.115	0.092	0.368	0.040	0.375
	ConstTSL-1	0.191	0.128	0.092	0.411	0.040	0.390
	ConstTSL-2	0.191	0.139	0.092	0.385	0.040	0.386
	PresMusOper-1	0.167	0.131	0.086	0.373	0.046	0.368
	PresMusOper-2	0.167	0.131	0.082	0.380	0.047	0.368
	PresMusOper-3	0.167	0.131	0.088	0.371	0.045	0.369

When PI = 0, all methods generate the same solutions with the exception of ConstTSL.
When PI≠0, the ConstTSL and PresMusOper methods do not scale L_m^o and L_T^S by the same factor.

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PresMusOper: $p = 0.005$) muscles compared to the UA_1 muscles.

For the between category analysis, scaling technique, muscle group and their interaction were found to have a significant effect on the MSSD ($p<0.001$). The PresMusOper methods generated the solutions that best preserved $L_m^{norm}(\mathbf{q})$ between scaled and unscaled models ($p<0.001$), while the LinScal methods yielded a lower MSSD than ConstTSL ($p<0.001$) (Fig. 4). The differences between solution $L_m^{norm}(\mathbf{q})$ returned by the different methods

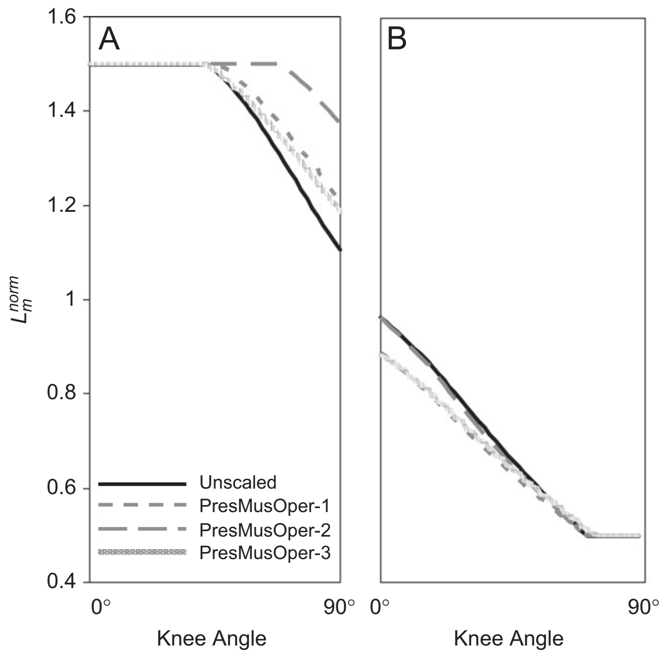

Fig. 3. $L_m^{norm}(\mathbf{q})$ obtained for scaled anatomical model of the subject with PI = 0.17 using the three PresMusOper methods compared to $L_m^{norm}(\mathbf{q})$ for the unscaled model. (A) Hip angle fixed at 0°, (B) hip angle fixed at 44°. PresMusOper-3 provided the best fit over the entire range of motion, but PresMusOper-2 provided a better fit throughout the range of motion bounded by the constraint points $P1$ and $P2$ (B) used for PresMusOper-2.

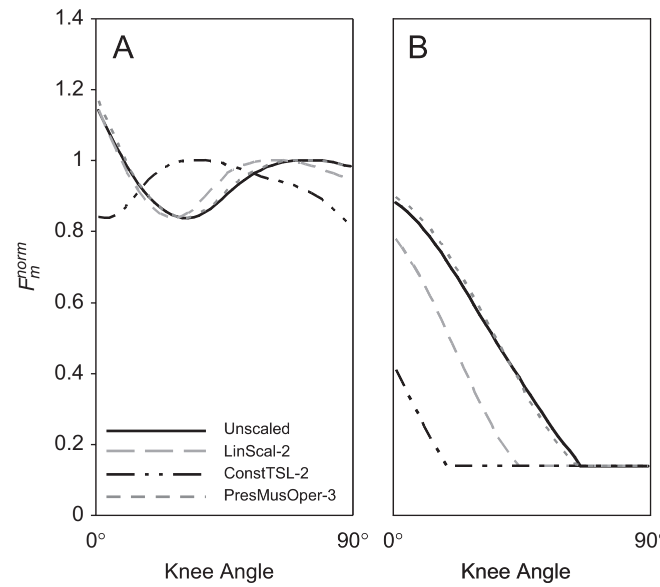

Fig. 5. Muscle forces at maximum activation, F_m^{norm}, corresponding to $L_m^{norm}(\mathbf{q})$ obtained for the medial gastrocnemius of the scaled anatomical model of the subject with PI = 0.21 using LinScal-2, ConstTSL-2 and PreMusOper-3 compared to that of the unscaled model. (A) Ankle fixed at 0°, (B) ankle fixed at 44°. Note the large discrepancy between the LinScal-2 and ConstTSL-2 forces and the unscaled forces, particularly in the shorter MTU range (B), and the close match between the PresMusOper-3 and unscaled forces.

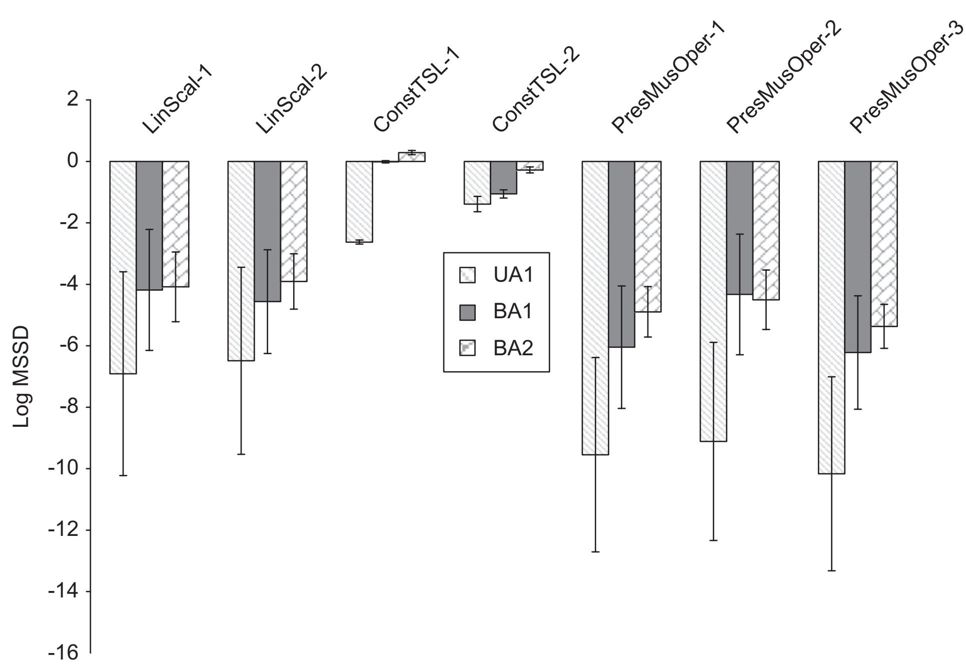

Fig. 4. Log MSSD across all subjects for each of the seven different techniques for the three muscle groups (UA_1, BA_1, and BA_2). Lower values correspond to a better preservation of $L_m^{norm}(\mathbf{q})$ between scaled and unscaled models. Refer to the text for significant differences.

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potentially correspond to very large discrepancies between scaled and unscaled force-generating characteristics—up to 20% for LinScal and 70% for ConstTSL (Fig. 5).

4. Discussion

True to their design, the PresMusOper methods best preserved $L_m^{norm}(\mathbf{q})$, with multiple constraint points performing best (PresMusOper-3), probably because $L_m^{norm}(\mathbf{q})$ like L_{MT} does not scale uniformly (Fig. 2) unless subject PI is zero. Hence, a solution L_m^O and L_T^S yielding an exact match for L_m^{norm} at two constraint points $P1$ and $P2$ may not best approximate $L_m^{norm}(\mathbf{q})$ across $\mathbf{q}$. Moreover, $\mathbf{q}$ in $L_m^{norm}(\mathbf{q})$ should be selected to reflect the movement of interest rather than the entire range of motion for best preservation. The fact that PresMusOper-2 yielded a better fit throughout the limited range bounded by its $P1$ and $P2$ exemplifies this, as does the greater MSSD for the biarticular muscles with their broader range of $L_m^{norm}(\mathbf{q})$. PresMusOper-3 should be adopted where possible since its optimisation process demanded negligible computation time.

ConstTSL methods were least effective in preserving $L_m^{norm}(\mathbf{q})$ despite utilising more information than the LinScal methods. Both ConstTSL-2 and LinScal-2 utilised the same L_m^O, and therefore the differences in solution L_T^S between them reflect the range of potential L_T^S that satisfy Eqs. (1 and 2) for a given L_m^O.

The strong correlation between PI and the MSSD of LinScal-2 suggests non-uniform bone scaling factors reduce linear scaling's capacity to replicate $L_m^{norm}(\mathbf{q})$. PresMusOper-3 was immune to this non-uniformity, which may be the mechanism by which it better replicated $L_m^{norm}(\mathbf{q})$. The differences associated with ConstTSL were independent of PI, perhaps because its solutions were the average of a range. This is reinforced by the case subject with a PI of zero, for which the MSSD was zero for all techniques except ConstTSL.

Caution is required when scaling functional characteristics according to subject size as fascicle length does not correlate with bone dimensions (Ward et al., 2005) or MTU length (Ward et al., 2007). Furthermore, neither the knee torque/angle relationship (Herzog et al., 1991b) nor the sarcomere length/joint angle relationship (Lieber et al., 1997) is the same for all people. These differences may be a functional adaptation (Blazevich et al., 2003; Herzog et al., 1991a), and therefore functional scaling of the MTU properties should be undertaken (Garner and Pandy, 2003; Lloyd and Besier, 2003).

When performing functional scaling, in our experience starting with unscaled L_m^O and L_T^S values may result in $L_m^{norm}(\mathbf{q})$ starting and/or being adjusted to be outside the 0.5–1.5 physiological range (Zajac, 1989). The PresMusOper methods should be used to provide a standardised $L_m^{norm}(\mathbf{q})$ starting point for each muscle that is then refined by suitably constrained functional calibration (Buchanan et al., 2005; Lloyd and Besier, 2003), particularly in light of the large range of solution L_m^O and L_T^S afforded by the methods reviewed here and the sensitivity of F_m^{norm} to the subsequent $L_m^{norm}(\mathbf{q})$ (see Fig. 5). However, to ensure that functional adjustment of L_m^O and L_T^S is within physiological limits, constraints should still be used, such as ± 2 standard deviations around the mean values reported in the literature (Lloyd and Buchanan, 1996), or by constraining $L_m^{norm}(\mathbf{q})$ (Garner and Pandy, 2003).

In this current analysis, the MTU attachment and path constraint points were scaled linearly and equally with segment length in all three dimensions, using a certain set of markers and landmarks. However, the current results would probably not be affected if anthropometric scaling was performed using either a different marker set and associated landmarks, or from imaging methods, such as MRI (e.g. Arnold et al., 2000) and ultrasound (e.g Ito et al., 2000). This is stated as all techniques evaluated, with the exception of LinScal-1, use changes in MTU length between subjects. However, LinScal-1 may be affected and may explain the poor correlation between MTU properties and bone dimensions (Ward et al., 2005). In cases where other methods facilitate subject-specific anthropometric sizing/scaling of the anatomical model, the PresMusOper methods would provide the best option to scale MTU parameters.

A disadvantage of all techniques presented here is that they require *a-priori* knowledge of L_m^O and L_T^S for the unscaled reference anatomical model, obtained in this case from cadaver studies (Delp et al., 1990). However, this is no different from any scenario where muscle properties cannot be directly measured.

This paper examined seven techniques for scaling L_m^O and L_T^S, comparing their capacity to preserve $L_m^{norm}(\mathbf{q})$. Preservation of $L_m^{norm}(\mathbf{q})$ is suggested to be a good rationale for scaling, as the methods which best achieve this provide a good starting point for subject-specific functional scaling of MTU properties, or for forward dynamic modelling using a generic model.

Conflict of interest statement

This study was funded by the Australian National Health and Medical Research Council and the University of Western Australia. Neither funding agency had any role in the design of the study. We do not believe that there are any other items which constitute a conflict of interest.

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