

MUSCLE–TENDON TRANSFORMATION DURING RESISTANCE LOADING: FORCE REFERENCES, ADAPTATION TIMING, AND CLINICAL INTERPRETATION CRITICAL NARRATIVE REVIEW WITH ANALYTICAL ILLUSTRATION

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ABSTRACT

Muscular force capacity and tendon resistance to elongation respond differently to resistance loading. In initially untrained men, 12 weeks of explosive and sustained contractions increased patellar tendon stiffness by 20% and 16%, respectively; tendon–aponeurosis stiffness increased only with sustained contractions. In nine men, Achilles tendon stiffness increased by 50.3% after three months of training and returned to the pre-training value within one month of detraining. Across two years of adolescent development, 18 volleyball athletes increased patellar tendon cross-sectional area by 27% and stiffness by 25%, while tendon stress decreased by 9%. These changes cannot be interpreted solely from muscular strength or from a stiffness slope estimated over a changing force interval. A nonlinear calculation gives lower elongation at 1200 N but higher strain when maximum force increases to 2100 N. Prospective adolescent observations associate strain of at least 9% with subsequent symptoms (risk ratio 2.3; 95% confidence interval 1.1–4.9), without establishing a universal injury threshold. Common-force elongation, current-maximum strain, anatomical dimensions, seasonal variation, and symptoms therefore require separate assessment. Error propagation must retain the dependence between force, elongation, and derived stiffness.

Keywords: resistance training; tendon strain; muscle strength; tendon stiffness; neuromuscular adaptation; mechanical loading

INTRODUCTION

Resistance training changes the force that a muscle can exert on its tendon. The change may arise from neural activation, muscle size, or muscle architecture. After 12 weeks of knee-extensor training, changes in agonist activation, quadriceps volume, and initial strength explained approximately 60% of the variation in strength gains in the study by Balshaw and colleagues (Balshaw et al., 2017). After four weeks of training, Del Vecchio and colleagues identified changes in recruitment and discharge behaviour in tracked motor units that accompanied increased force (Del Vecchio et al., 2019). These measurements describe muscular adaptation; tendon elongation requires an additional mechanical assessment.

Tendon adaptation depends on the loading stimulus. The intervention meta-analysis by Bohm and colleagues identified loading magnitude as a determinant of changes in tendon mechanical properties (Bohm et al., 2015). An increase in

stiffness reduces elongation at a given force only when the compared measurements use compatible reference lengths and loading conditions. At maximum contraction, the muscle may generate a larger force after training. Maximum tendon strain then reflects the combined changes in force capacity and resistance to elongation.

For the knee-extensor and plantar-flexor systems, transformation is defined here as a change in force-generating capacity, tendon force–elongation behaviour, or their interaction after repeated loading. The observation time distinguishes adaptation during training from retention or reversal after loading decreases. External resistance, relative effort, and maximum voluntary contraction impose different comparison conditions: the same external weight need not generate the same internal tendon force, and the same percentage of maximum effort need not represent the same absolute force after strength increases.

Arampatzis, Mersmann, and Bohm described maximum strain in relation to muscular strength and tendon stiffness (Arampatzis et al., 2020). Massey and colleagues assessed tendon stiffness over a common pre-training force interval during resistance training (Massey et al., 2018). These approaches address complementary questions: elongation at a matched force identifies the mechanical response under comparable loading, whereas strain at current maximum contraction identifies deformation under the participant's available force capacity. Anatomical segment, stiffness definition, and measurement timing determine whether results from different studies answer either question.

FORCE-REFERENCE DEFINITIONS

REFERENCE LENGTH AND ENDPOINT SECANT STIFFNESS

Tendon strain is elongation divided by reference length. Stiffness relates tensile force to elongation; stress additionally requires cross-sectional area. Young's modulus requires dimensions and an estimate of the stress–strain slope. Joint torque cannot substitute directly for these quantities. Ultrasound assessments depend on anatomical tracking, tendon-force estimation, synchronisation, and viscoelastic test conditions (Seynnes et al., 2015). A stiffness value alone consequently cannot identify strain at maximum contraction.

Let $j = 0$ and $j = 1$ denote two assessments. For assessment j , choose a reference force $F_{r,j}$ and measure its associated tendon length $L_{r,j} = L_j(F_{r,j})$. At a target force $F_j > F_{r,j}$, define

$$\Delta F_j = F_j - F_{r,j}, \quad \delta_j = L_j(F_j) - L_j(F_{r,j}), \quad K_{s,j} = \frac{\Delta F_j}{\delta_j}, \quad \varepsilon_j^{(r)} = \frac{\delta_j}{L_{r,j}}. \quad (1)$$

Here K_s is endpoint secant stiffness relative to the chosen reference. Positive elongation and compatible reference conventions are required. The resulting identities are

$$\varepsilon_j^{(r)} = \frac{\Delta F_j}{K_{s,j} L_{r,j}}, \quad \frac{\varepsilon_1^{(r)}}{\varepsilon_0^{(r)}} = \frac{\Delta F_1}{\Delta F_0} \frac{K_{s,0}}{K_{s,1}} \frac{L_{r,0}}{L_{r,1}}. \quad (2)$$

For a nominal zero-force reference, $\Delta F_j = F_j$ and the usual engineering strain notation ε_j is used below. For a preloaded reference, the strain describes the additional elongation relative to that preloaded length; it cannot be silently substituted for total strain from an unloaded reference. An unknown passive reference force prevents exact evaluation from net joint torque alone.

With constant reference length, maximum strain increases when maximum force increases proportionally more than endpoint secant stiffness. Eq. (2) is an algebraic identity. On a nonlinear force–elongation curve, selecting a higher endpoint can increase secant stiffness even when the curve is unchanged. The ratio therefore quantifies deformation under the stated references; it does not identify collagen remodelling or injury risk.

TANGENT STIFFNESS AND THE FORCE INTERVAL

Tangent stiffness is instead

$$K_t(F) = \frac{dF}{d\delta}. \quad (3)$$

The derivative requires a specified loading branch, rate, posture, and conditioning history. The unloading branch may differ. A regression slope fitted to part of a ramp contraction estimates stiffness over that interval and is generally unequal to F/δ . Substituting a high-force regression slope into Eq. (2) requires verification of the curve and intercept. After strength increases, a force interval defined as a percentage of current maximum moves to higher absolute forces. Its slope can increase because a different region of the nonlinear curve is sampled.

A comparison of mechanical response requires a common absolute force interval reached at both assessments, together with a separate measurement at each assessment's maximum. Kösters and colleagues demonstrated loading-rate

dependence of patellar tendon stiffness and modulus estimates (Kösters et al., 2014). Equal percentage effort therefore does not ensure equal force or testing rate. Anatomical segment, reference posture, force interval, ramp rate, and loading history must remain compatible when attributing a difference to training.

TABLE 1. FORCE CONDITIONS AND THE TENDON MEASUREMENT OBTAINED UNDER EACH CONDITION.

Force condition	Measurement obtained	Force and test requirements
Common absolute tendon force	Does the tendon elongate less under the same estimated internal force?	Use an overlapping force range and the same anatomical and loading-rate conventions.
Current maximum contraction	What strain occurs at maximum voluntary force at each assessment?	Both force capacity and tendon behaviour may have changed.
Same percentage of current maximum	What strain accompanies equal relative effort at each assessment?	Relative effort can correspond to different absolute forces after strength gain.
Same external exercise load	Has the movement under the same external resistance changed?	Internal tendon force can change with joint geometry, movement strategy, and force sharing.

Table 1 specifies the quantities held constant in four test conditions. Common-force elongation measures response to matched loading. Current-maximum strain measures deformation at available muscular capacity. The second comparison includes a change in the force-generating muscle and therefore cannot isolate tendon adaptation.

NONLINEAR FORCE ELONGATION CALCULATION

Let force F be measured in newtons and elongation δ in millimetres. Prescribe a zero-reference loading curve

$$f_0(\delta) = a\delta + b\delta^2, \quad a = 100 \text{ N mm}^{-1}, \quad b = 50 \text{ N mm}^{-2}, \quad L_r = 80 \text{ mm}. \quad (4)$$

These coefficients define a mathematical calculation, without fitting participant measurements. At a maximum of 1200 N, $\delta_0 = 4 \text{ mm}$, $\varepsilon_0 = 5.00\%$, and $K_{s,0} = 300 \text{ N/mm}$. With the same curve and a maximum of 2100 N, $\delta = 5.557 \text{ mm}$ and $K_s = 377.9 \text{ N/mm}$. The endpoint secant stiffness increases by 26.0% despite an unchanged force–elongation function (Figure 1A).

Now prescribe a changed curve $f_1(\delta) = 1.2f_0(\delta)$ and a new maximum of 2100 N. At the common force of 1200 N, the inverse curve gives

$$\delta_1(1200) = \frac{-a + \sqrt{a^2 + 4b(1200/1.2)}}{2b} = 3.583 \text{ mm}. \quad (5)$$

Common-force elongation decreases by 10.4%. At 2100 N, $\delta_1 = 5 \text{ mm}$, $\varepsilon_1 = 6.25\%$, and $K_{s,1} = 420 \text{ N/mm}$. Maximum force increases by 75%, endpoint secant stiffness by 40%, and maximum strain by 25%; Eq. (2) gives $1.75/1.40 = 1.25$. Lower elongation under the same force thus coexists with higher strain under the larger maximum force (Figure 1B). The calculation concerns force reference alone and does not estimate a treatment effect or injury threshold.

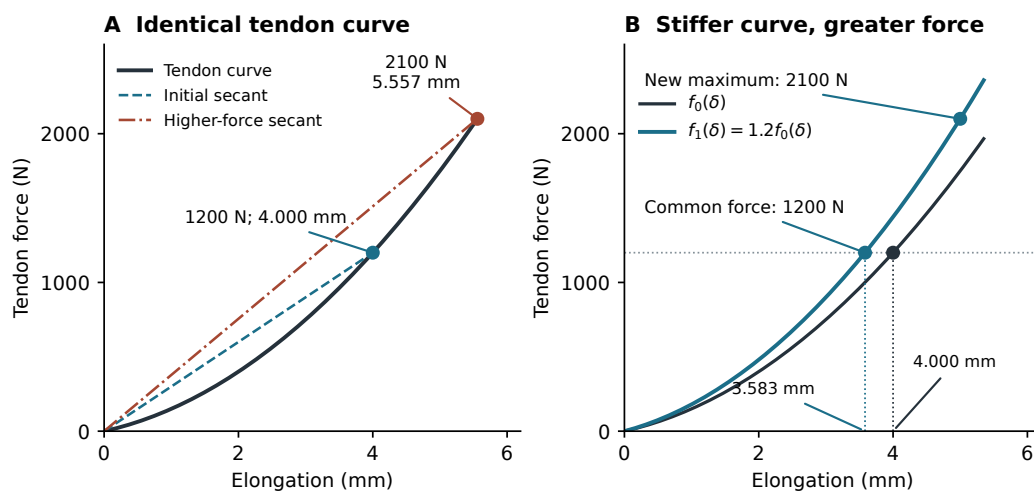


FIGURE 1. CALCULATED DEFORMATION UNDER TWO FORCE REFERENCES. (A) RAISING THE FORCE ENDPOINT FROM 1200 TO 2100 N ON f_0 INCREASES SECANT STIFFNESS FROM 300 TO 377.9 N/MM. (B) ON $f_1 = 1.2f_0$, ELONGATION AT 1200 N DECREASES FROM 4.000 TO 3.583 MM; ELONGATION AT 2100 N IS 5.000 MM. REFERENCE LENGTH IS 80 MM, GIVING MAXIMUM STRAINS OF 5.00% AND 6.25% AT THE RESPECTIVE CAPACITY ENDPOINTS. CURVES AND COORDINATES ARE CALCULATED FROM EQ. (4), WITHOUT PARTICIPANT MEASUREMENTS.

TIME COURSE AND ANATOMICAL DISTRIBUTION OF ADAPTATION

TRAINING AND DETRAINING OBSERVATIONS

The motor-unit changes identified after four weeks of training (Del Vecchio et al., 2019) demonstrate that force capacity can increase through altered neural activation. Tendon adaptation must be dated from repeated tendon measurements rather than inferred from the onset of strength gain.

Kubo and colleagues assessed nine men aged 21–27 years monthly during three months of isometric plantar-flexion training and three months of detraining (Kubo et al., 2012). Achilles tendon stiffness increased by 50.3% at the end of training and returned to the pre-training value after one month of detraining. Ultrasound echointensity increased by 9.1% after two months and remained elevated, whereas magnetic-resonance signal intensity increased by 24.2% after two months and 21.4% after three months before returning to the pre-training value during detraining. Mechanical stiffness and imaging measurements therefore followed different time courses. Imaging changes do not directly quantify collagen remodelling, and the sample does not establish a universal duration required for tendon adaptation.

The 50.3% stiffness increase describes the three-month assessment, whereas the measurement one month later describes reversal during detraining. A final strength value cannot recover intermediate tendon strain, and a final stiffness value cannot establish retention. Follow-up timing and loading between assessments are consequently required to distinguish acquisition from persistence of a mechanical change.

CONTRACTION DURATION, TENDON SEGMENT, AND TRAINING STATUS

In a 14-week Achilles tendon study, Bohm and colleagues varied loading duration and strain rate while maintaining similar strain magnitude and loading volume (Bohm et al., 2014). Three-second loading produced larger increases in stiffness and modulus than twelve-second loading; the jumping condition did not produce a statistically significant stiffness increase. Under these conditions, longer duration or higher strain rate did not increase adaptation consistently. Duration, magnitude, rate, and loading volume must therefore be specified individually.

After 12 weeks, Massey and colleagues found 20% and 16% increases in patellar tendon stiffness with explosive and sustained contractions, respectively, relative to the common pre-training force interval (Massey et al., 2018). Both changes exceeded the control response. Tendon–aponeurosis stiffness increased only with sustained contractions. The free patellar tendon and the tendon–aponeurosis complex thus exhibited different responses within the same training comparison.

In 28 already-trained men, Walker and colleagues compared ten weeks of accentuated eccentric loading, traditional training, and habitual training (Walker et al., 2020). Accentuated eccentric loading increased quadriceps fascicle length without a detectable patellar tendon-property change. The ten-week tendon-property test–retest coefficient of variation was 29.0%, limiting sensitivity to smaller changes. This result applies to the tested population and assessment precision; it cannot establish that eccentric loading has no tendon effect.

Table 2 specifies the force intervals and outcome denominators for these two comparisons. Training status, anatomical segment, and measurement variability restrict direct comparison of their tendon responses.

TABLE 2. RESISTANCE-TRAINING OUTCOMES WITH THEIR ANATOMICAL TARGETS AND FORCE INTERVALS. PERCENTAGES DESCRIBE GROUP-MEAN CHANGES; SAMPLE SIZES IDENTIFY THE RELEVANT ANALYSIS.

Study and design	Mechanical comparison and outcome	Interpretation and limitation
Massey et al. (2018): 42 men randomized; 12 weeks.	Patellar-tendon analyses: 13 explosive, 15 sustained, 12 controls. Common-force slope at 70–80% pre-training maximum torque; stiffness increased by 20% and 16% in the training groups.	Both changes exceeded controls. Tendon–aponeurosis analysis used a different anatomical target and outcome sample; its stiffness increased only with sustained loading. Young, initially untrained men limit transfer to other populations.
Walker et al. (2020): 28 trained men randomized; 10 weeks.	Traditional, accentuated eccentric, and habitual-training groups. Patellar stiffness assessed over 50–100% maximum force and to the same pre-training force level. No detectable tendon-property change.	Tendon-property test–retest CV 29.0%; null results cannot establish equivalence. Fascicle-length adaptation does not establish tendon adaptation.

MATURATION AND SEASONAL VARIABILITY

Mersmann and colleagues found higher patellar tendon stress and smaller tendon cross-sectional area in adolescent than middle-aged athletes (Mersmann et al., 2014). These cross-sectional differences identify distinct force-to-area relationships between age groups. They do not establish the preceding adaptation of any individual athlete.

During one year of follow-up in 12 adolescent athletes and eight similarly aged controls, muscular strength and tendon stiffness increased in both groups (Mersmann et al., 2016). Athletes exhibited less uniform changes in force and stiffness and higher maximum strain: $7.6 \pm 1.7\%$ compared with $5.5 \pm 0.9\%$ in controls (mean $\pm$ standard deviation). Group means describe the tested samples and cannot determine an individual diagnostic threshold.

A two-year study followed 18 elite volleyball athletes from 16 ± 1 to 18 ± 1 years (Mersmann et al., 2017). Muscular strength increased by 13%, muscle anatomical cross-sectional area by 6%, and muscle volume by 6%. Patellar tendon cross-sectional area increased by 27% and stiffness by 25%, while tendon stress decreased by 9%; these changes reached $p < 0.05$. The larger proportional increase in tendon area relative to muscular strength is consistent with the reduction in stress. Adolescent tendon development therefore cannot be described as uniformly slower than muscular development across all ages and observation intervals.

Waugh and colleagues found increased Achilles tendon stiffness and modulus after resistance training in prepubertal children (Waugh et al., 2014). Decreases in peak strain and strain at a given force did not reach statistical significance. Stiffness adaptation consequently did not establish either strain change. Force, tendon dimensions, and deformation must be measured together to identify how growth and training alter the unit.

In 67 healthy elite jumpers, comprising 35 men and 32 women, Epro and colleagues found greater strength and Achilles stiffness in the take-off limb (Epro et al., 2019). Strength and stiffness symmetry indices were correlated ($r = 0.44$). This between-limb association is compatible with coordinated properties at assessment, but cannot determine the sequence of their adaptation or exclude selection of athletes with favourable properties.

Karamanidis and Epro retained 18 jumpers and 12 controls for a one-year analysis and six athletes for four-year follow-up (Karamanidis and Epro, 2020). Successive changes in strength and stiffness had lower similarity in athletes. Between-group differences in one-year stiffness and maximum-strain fluctuations did not reach statistical significance. Thus, coordinated properties in a bilateral cross-sectional assessment can coexist with less uniform changes within individuals. The small retained samples and exclusions limit transfer to other athletic populations.

TENDON STRAIN AND CLINICAL OUTCOMES

STRAIN BEFORE SYMPTOM ONSET

Mersmann and colleagues found higher patellar tendon strain before clinically relevant symptoms developed in adolescent athletes (Mersmann et al., 2023). Among 44 athletes with usable follow-up, 14 were symptomatic and 23 asymptomatic; seven were excluded from this classification. Strain $\geq 9\%$ was associated with a symptom risk ratio of 2.3 (95% confidence interval 1.1–4.9). The temporal sequence supports association in this population. It does not establish that reducing strain prevents symptoms, nor does it determine sensitivity, specificity, or a universal diagnostic threshold.

SEASONAL LOADING INTERVENTIONS

The functional high-load handball intervention by Mersmann and colleagues reduced pain prevalence without significant changes in stiffness or maximum strain (Mersmann et al., 2021). Control and intervention cohorts were recruited in successive years, with five athletes participating in both. Tendon outcomes were available for 12 athletes per group. Because a group-mean strain reduction was not established, these measurements do not identify strain reduction as the mediator of pain benefit. Limited precision also prevents exclusion of an individual association between mechanical and clinical change.

Domroes and colleagues compared 13 adolescent handball athletes per group over successive seasons, with four participating in both cohorts (Domroes et al., 2024). Approximately 31 weeks of personalized loading reduced residual maximum-strain variation ($p = 0.005$), without significant changes in mean maximum strain or the ultrasound structural outcome. Maximum strain was extrapolated to maximum force using a polynomial fit, and stiffness was estimated over 50–80% of current maximum force. Calendar-season differences and extrapolation uncertainty restrict attribution of the strain-variation change to the intervention.

Weidlich and colleagues assigned two adult male volleyball teams to personalized loading or usual training (Weidlich et al., 2024). The analysis included ten intervention and twelve control athletes at four assessments over nine months.

Maximum strain increased in controls; the intervention change did not reach significance. Residual strain fluctuations decreased in the intervention group. Team assignment, attrition, and unequal schedules limit causal attribution. The mechanical outcomes do not establish prevention of clinical injury.

None of these personalized-loading comparisons used a concurrent comparator receiving an otherwise matched tendon-loading dose without personalization. Their findings therefore cannot isolate the effect of personalization from exercise dose or seasonal exposure. Residual variation additionally depends on the fitted time trend and measurement error. Table 3 identifies the allocation and force conventions needed to interpret each outcome.

SAMPLING FREQUENCY AND MOVEMENT LOADING

Four seasonal measurements identify change at the sampled time points and variation around an estimated trajectory. They do not measure the number, amplitude, or duration of tendon-loading cycles during sport. Equal initial and final strain values can occur despite an intermediate increase. A lower final value similarly does not establish lower cumulative exposure between assessments.

Kharazi and colleagues quantified Achilles tendon loading and elastic strain energy during walking and running (Kharazi et al., 2021). Such movement-specific measurements address deformation during locomotion rather than deformation during an isolated maximum voluntary test. Stiffness, strain, symptoms, and performance are distinct outcomes; a change in one cannot substitute for measurement of another.

TABLE 3. ATHLETIC FOLLOW-UP AND LOADING INTERVENTIONS: PARTICIPANT COUNTS, ASSESSMENT TIMING, TENDON OUTCOMES, AND LIMITS OF INFERENCE.

Study and design	Sample and observation schedule	Tendon measurements	Allocation or measurement limitation
Epro et al. (2019): cross-sectional.	67 elite jumpers, 35 men and 32 women; bilateral assessment.	Take-off limbs had greater strength and Achilles stiffness; symmetry indices correlated ($r = 0.44$).	Between-limb coordination cannot demonstrate adaptation over time; healthy-elite selection restricts transfer.
Karamanidis and Epro (2020): longitudinal.	One-year valid sample: 18 healthy jumpers and 12 controls; six athletes followed for four years.	Lower similarity of successive strength and stiffness changes in athletes; one-year stiffness and strain fluctuation differences did not reach significance.	Attrition and small retained samples; stiffness estimated over 30–80% current maximum force, maximum strain extrapolated.
Mersmann et al. (2021): successive-year intervention cohorts.	Recruited 18 controls and 16 intervention players; 12 per group for tendon outcomes; four seasonal assessments.	Pain benefit without significant stiffness or maximum-strain change. Regression stiffness from 50% maximum force to the common ramp-contraction maximum.	Five returning athletes; calendar-year confounding, missing ultrasound observations, and outcome-specific samples.
Domroes et al. (2024): successive-season intervention cohorts.	13 analyzed per group; four assessments; approximately 31-week intervention.	Lower residual maximum-strain variation ($p = 0.005$); no significant mean-strain or structural change. Stiffness from 50–80% current maximum force.	Four returning athletes; noncurrent controls; polynomial extrapolation of maximum strain.
Study and design	Sample and observation schedule	Tendon measurements	Allocation or measurement limitation
Mersmann et al. (2023): prospective symptom follow-up.	44 with usable follow-up; 14 symptomatic, 23 asymptomatic, seven unclassified exclusions; four seasonal time points.	Strain $\geq 9\%$: symptom risk ratio 2.3, 95% CI 1.1–4.9.	Observational association, modest sample and attrition; thresholds require population-specific validation.
Weidlich et al. (2024): team-allocated intervention.	12 controls and 10 intervention players analyzed; four assessments over nine months.	Maximum strain increased in controls; no significant intervention increase; residual fluctuations decreased in the intervention group.	Team confounding, attrition, unequal schedules; injury-prevention benefit and superiority to matched tendon loading remain unestablished.

PAIRED-FORCE ASSESSMENT AND ERROR PROPAGATION

COMMON-FORCE ELONGATION AND CURRENT-MAXIMUM STRAIN

A paired assessment comprises elongation at a common absolute tendon force and strain at the participant's current maximum. The common force must lie within the range reached at every compared assessment. A value outside this overlap requires extrapolation and cannot be treated as an observed matched-force response.

Current-maximum force and strain must be interpreted jointly. Higher maximum strain with lower common-force elongation indicates greater deformation at available muscular capacity despite lower deformation under matched loading. Lower values under both conditions indicate decreased deformation at both force references. A slope available

only over a percentage of current maximum cannot resolve common-force elongation without the underlying curve. None of these comparisons alone diagnoses tendon pathology.

Reference length and cross-sectional area identify dimensional contributions to strain and stress. Their omission prevents separation of elongation from its length denominator and force from its area denominator. Anatomical identification is equally necessary: the free tendon and tendon–aponeurosis complex responded differently in the Massey comparison (Massey et al., 2018). Combining their stiffness values would conceal that anatomical difference.

DEPENDENCE AMONG DERIVED QUANTITIES

Within-person change must be assessed against repeatability under the same test conditions. The 29.0% tendon-property test–retest coefficient of variation in Walker and colleagues' study (Walker et al., 2020) illustrates the uncertainty that can accompany a small apparent change. Confidence intervals and measurement variability determine which effect sizes remain compatible with the observations; statistical nonsignificance alone cannot establish identical tendon behaviour.

The uncertainty of derived quantities must respect their algebraic dependence. If the same measured endpoint gives $\widehat{K}_s = \widehat{\Delta F} / \widehat{\delta}$, then

$$\frac{\widehat{\Delta F}}{\widehat{K}_s} = \widehat{\delta}. \quad (6)$$

The shared force estimate cancels exactly. Treating force and this secant stiffness as independent would introduce an artificial contribution to strain uncertainty. For $\widehat{\varepsilon} = \widehat{\delta} / \widehat{L}_r$, first-order propagation gives

$$\text{Var}(\widehat{\varepsilon}) \approx \frac{\text{Var}(\widehat{\delta})}{L_r^2} + \frac{\delta^2 \text{Var}(\widehat{L}_r)}{L_r^4} - \frac{2\delta \text{Cov}(\widehat{\delta}, \widehat{L}_r)}{L_r^3}. \quad (7)$$

Eq. (7) applies locally to the elongation-to-length ratio, rather than to a fitted tangent or interval slope. Force error affects selection of the common-force position, slope fitting, and extrapolation to maximum force. An error in the estimated moment arm changes the force scale even when elongation divided by reference length remains unchanged. Statistical resampling of participant records must preserve the dependence among force, elongation, reference length, and repeated assessments. The equation specifies that dependence without estimating uncertainty from unavailable participant records.

Training attendance, resistance, contraction duration, and concurrent sport exposure describe the loading received between measurements. The lower-limb tendon meta-analysis by Lazarczuk and colleagues identified mechanical, material, and morphological adaptation across heterogeneous loading interventions (Lazarczuk et al., 2022). Interpretation therefore requires the actual loading dose and test conditions rather than an exercise label alone.

MECHANICAL INTERPRETATION AND CLINICAL VALIDATION

Disappearance of a stiffness difference after force-range matching is compatible with a change in the sampled curve interval. Persistence of lower common-force elongation with higher maximum strain identifies a changed response under increased muscular demand. Both interpretations require compatible joint geometry, loading rate, and anatomical tracking. The adolescent strain–symptom association (Mersmann et al., 2023) does not validate either pattern as a diagnostic test in adults or in other tendons. Clinical interpretation requires symptom onset and function to be observed independently of the mechanical measurements.

CONCLUSION

Resistance loading can increase muscular force capacity and tendon stiffness at different rates and in different anatomical segments. Elongation at a common absolute force identifies mechanical response under matched loading; current-maximum strain identifies deformation under available muscular capacity. A stiffness increase cannot establish lower maximum strain without the corresponding force and length references. Training–detraining observations, adolescent tendon growth, and seasonal strain variation show why observation time is also necessary. The association between high strain and later symptoms in adolescent athletes supports clinical follow-up, but does not convert a mechanical ratio into a universal injury threshold. Force, deformation, dimensions, timing, and symptoms must remain separately identified when interpreting muscle–tendon transformation.

DECLARATIONS

AUTHOR CONTRIBUTIONS

All authors contributed equally and all authors read and approved the final version of this manuscript.

CONFLICT OF INTEREST

The authors declare that there are no conflicts of interest related to this study.

DATA AVAILABILITY

The data supporting the findings of this study are available from the corresponding author upon reasonable request.

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