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TECHNICAL NOTE

A LOW PROFILE HUMAN TENDON FORCE TRANSDUCER: THE INFLUENCE OF TENDON THICKNESS ON CALIBRATION

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Abstract—An *in vitro* calibration method for human tendon force transducers using tendon thickness to predict the calibration factor has been previously proposed (An *et al.*, 1990, *J. Biomechanics* **23**, 1269–1271). However, changes in the calibration factor due to changing tendon geometry during repeated tendon loading are unknown. A new, low-profile transducer design that measures tendon thickness in the transducer, *in situ*, is developed. An empirical model estimating the transducer's calibration factor is developed using data from *in vitro* tension testing of 12 fresh frozen human finger flexor tendons. Each tendon is preseated with ten loading cycles before data collection. Using tendon thickness, the model predicts the measured calibration factor to within 0–15% (average 6%). During repeated loading of an *in vitro* tendon, the calibration factor changes 15% over the first ten cycles (0–50 N) due to the observed changing tendon thickness. After the first ten loading cycles the variability of the calibration factor is reduced to less than 1% for the next three loading cycles. Hence this new, modified *in vitro* calibration procedure with tendon preseating reduces the cycle-to-cycle variability caused by the associated change in the tendon thickness. © 1997 Published by Elsevier Science Ltd.

Keywords: Tendon; Force transducer; Finger; Force.

INTRODUCTION

Tendon force transducers, commonly referred to as buckle transducers, have been used to measure tensions in the human finger flexor tendons (Schuind *et al.*, 1992), the human Achilles tendon (Komi *et al.*, 1987; Komi, 1990), the primate extraocular muscle tendon (Miller and Robins, 1992) and elbow tendons (Peres *et al.*, 1983), the cat gastrocnemius tendon (Gregor *et al.*, 1988; Sherif *et al.*, 1983; Whiting *et al.*, 1984), and the horse digital extensor tendon (Barnes and Pinder, 1974). Most animal studies use *in situ* methods for calibrating the tendon transducers, yet this technique is not practical for human studies. An *et al.* (1990) proposed and calibrated their transducer *in vitro* using an unchanging tendon thickness to determine the calibration factor for the *in vivo* applications. Because the tendon transducer applies a compressive load on the tendon at the points of contact, it is expected that the tendon thickness in the transducer will change over repeated loading cycles, i.e. the tendon seats itself into the transducer. Preliminary experiments confirm this expectation. Because of this seating process, the calibration factor changes with repeated loading cycles. A new low-profile transducer design, with *in situ* thickness measurements that account for the effects of tendon seating, is developed and validated. The effects of repeated tendon loading on *in situ* tendon thickness are presented.

METHODS

The transducer consists of a 9 × 16 × 4.5 mm stainless steel frame and a removable stainless steel fulcrum (Fig. 1) and fits tendons up to 5 mm wide, 3 mm thick. The tendon lies and self-aligns in the semi-circular arches ($r = 2.5$ mm) in the frame and fulcrum. The cylindrical design and arrangement of the wires allows the transducer to slide between other tendons with minimal interference. It does not twist the tendon. Lengthening of the muscle–tendon unit by passing the tendon over the fulcrum is small, only 1.2 mm for a 3 mm thick tendon. Material and transducer dimensions were designed for loads between 0 and 50 N, the physiological range of interest.

Two 500 Ω silicon uni-axial strain gauges placed on the top and bottom of the narrow beams of the frame form a half-Wheatstone bridge measuring the bending strain. The half-bridge allows for increased sensitivity to strain created from the bending load, minimal sensitivity to temperature, and sums the bending strains from both beams of the transducer. The wiring and gauges are waterproofed with an air-drying acrylic coating.

A three-post caliper was fabricated to measure tendon thickness in the transducer. Two posts rest on the transducer, and a central sliding post with a spring and locking mechanism is lowered to the top of the tendon to gauge its thickness at the fulcrum (Fig. 2). The spring provides a small, repeatable pressure perpendicular to the tendon path to reduce measurement variability.

To assess linearity, repeatability and drift, the transducer was supported at the ends and weights (0–14 N) were hung from the fulcrum simulating the three-point loading of the device at room temperature (18°C). These direct loading tests were repeated with the transducer heated, with a heat lamp, to body temperature of 37°C with the bridge completion circuit at room temperature. Differences were tested at 95% confidence levels with sample t-tests ($n = 5$). The dynamic response of the transducer was tested by tapping the unloaded transducer alone with a hammer instrumented with an accelerometer.

An empirical model predicting the calibration factor (converting transducer output to tension) was developed using 12 fresh frozen human finger flexor tendons of different thicknesses taken from five hands. The tendons were mounted in a uniaxial loading machine with a 111 N load cell and submerged in a 0.9% saline solution (20°C). To reduce variability the tendons were then preconditioned with 10 loading cycles as described by Woo (1982) and Viidik (1987). The cycles ranged from 0 to 50 N, the expected physiological force range of interest (An *et al.*, 1990; Rempel *et al.*, 1994; Schuind *et al.*, 1992). Transducer

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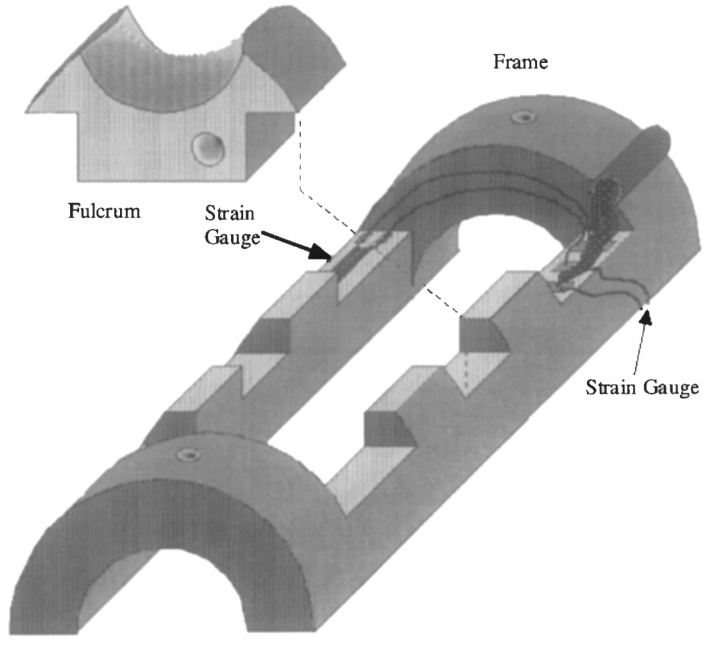

Fig. 1. The buckle transducer has two parts, the frame and removable fulcrum. Two strain gauges form a half-Wheatstone bridge to measure the deformation resulting from increased tension of the tendon over the fulcrum. The slim profile allows for easy insertion during carpal tunnel release surgery and minimizes interference from adjacent structures.

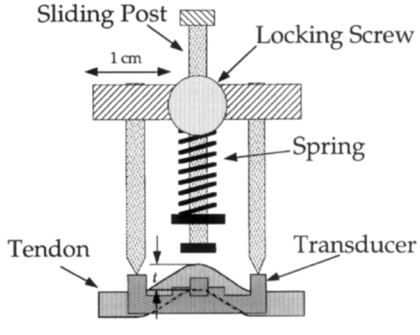

Fig. 2. Side view of the tendon thickness caliper with the transducer and tendon. The spring-loaded central sliding post slides down to rest on top of the tendon at the fulcrum to measure tendon thickness. Conversion of transducer output to tendon tension is a function of the tendon thickness, t.

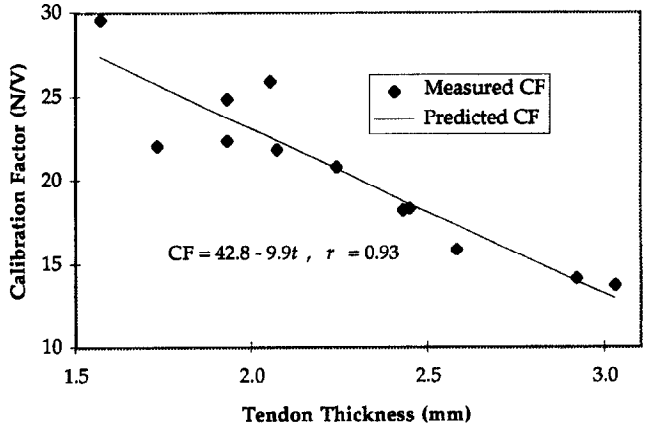

Fig. 3. Measured and predicted calibration factors for the 12 tendons. The linear model, $r = 0.93$, predicts the measured calibration factor within, on average, 6%. The model is used to determine the calibration factor for the human *in vivo* experiments, since *in situ* calibration methods are not practical.

output and load cell data were collected during three additional loading cycles. The calibration factor is defined as the slope of a straight line best fitting the transducer output versus tendon tension measured from the load cell for the three additional loading cycles. Tendon thickness was recorded five sequential times immediately following the last loading cycle while the tendon was under 1–3 N of tension.

During first calibration attempts it was observed that tendon thickness changed during repeated loading, affecting the calibration factor. To study this tendon seating effect, a fresh frozen human finger flexor tendon was loaded over 100 sinusoidal cycles (0–50 N). The tendon was submerged in a 0.9% saline solution. Tendon thickness was measured five times after every 20 cycles while the tendon was under 1–3 N of tension.

RESULTS

The transducer's output is linear ($r \geqslant 0.99$), repeatable (S.D. $\leqslant 1\%$, $n = 10$) and has no significant drift over 5 min for the direct loading tests. The gains measured at the body or room temperature range were not significantly different. Using the time response data from both the transducer and the instrumented hammer, the magnitude of the empirical transfer function estimation (Ljung, 1987) is flat until resonance at 660 rad s^{-1} (105 Hz).

Variations in the five repeated tendon thickness measurements for each tendon were small (S.D. 0.07 mm). The transducer's output for the three loading cycles was linear ($r \geqslant 0.99$) and the variation in the calculated calibration factor was small ($< 1\%$). A least-squares fit of the thickness and calibration factor (CF) data (Fig. 3) provides the linear model $CF = 42.8 - 9.9t$, ($r = 0.93$). The prediction error for the calibration factor ranges from 0 to 16% (mean = 6%, S.D. = 5%).

The error in the calibration factor creates a bias error in predicting the tension. Average relative errors calculated for each tendon tested ranged from 1 to 15% of scale (mean = 6%, S.D. = 4%) and maximum

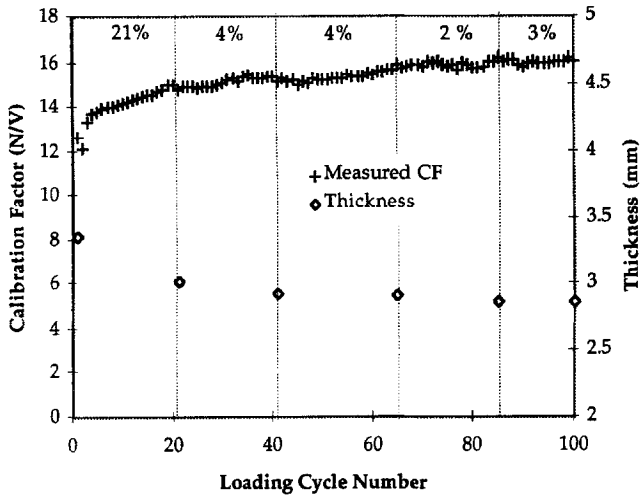

Fig. 4. The measured calibration for 100 loading cycles and the six thickness measurements taken every 20 cycles. The calibration factor (+) changes as the cadaver tendon seats in the transducer (15% over the first 10 cycles). The observed tendon thickness (◇) predicts similar changes in the calibration factor. As a result *in vitro* calibration procedures should preseat the tendons with several loading cycles to reduce the cycle-to-cycle variability.

relative errors ranged from 6 to 23% of scale (mean = 11%, S.D. = 5%). The absolute maximum errors ranged from 0.8 to 8.3 N (mean = 3.2 N, S.D. = 2.4 N).

Figure 4 shows the change in the calibration factor over the 100 loading cycles along with the change of the six thickness measurements taken after every 20 cycles. The calibration factor increases 21% over the first 20 cycles and then 4, 4, 2, and 3% for the second, third, fourth and fifth 20-cycle intervals, respectively. Using the model above, the observed change of thickness measurements predict an increase in the calibration factor of 30, 6, 1, 3, and 0% for the same 20 cycles intervals.

DISCUSSION

This study identifies effects at the tendon–transducer interface. For example, the transducer's output is linear with the applied tendon tension; however, the linear relationship described by the calibration factor changes with repeated loading (Fig. 4). During the repeated loading, the tendon deforms under compression at the tendon–transducer interface. Indentations in the tendon were visible at the point of contact with the transducer after removal. This change in thickness is measurable with the thickness caliper (Fig. 4) and, using the empirical model developed here, predicts similar changes in the calibration factor. To ensure accurate prediction of the calibration factor, thickness measurements should be made immediately following the loading cycles with the tendon in the transducer.

Standard tension testing methods for tendons require preconditioning the specimens with ten loading cycles to obtain repeatable data (Woo, 1982; Viidik, 1987). *In vitro* calibration methods require a similar procedure during which the tendon seats in the transducer. The cycle-to-cycle variability in the calibration factor is reduced by an order of magnitude after 10 cycles, and the change over the three cycles used to create the model averages less than 1%. After 20 cycles, the change of calibration factor does level off. Therefore, pre-seating reduces the cycle-to-cycle variation in the calibration factor.

Does tendon seating take place *in vivo*? Most likely it does. *In vivo* procedures having cyclic loading should include preseating the tendon before the thickness measurement and data collection, following the *in vitro* procedures. Monitoring the thickness during an *in vivo* experiment can record seating during the procedure and lead to correction of the calibration factor if needed.

This transducer design has three new features. First, the cylindrical design allows the transducer to slide more freely between other tendons. This is especially useful for the tight space in the carpal tunnel. The arrangement of the wires allows for the transducer to rotate freely with the tendon and therefore does not alter the tendon path. Second, the self-aligning arches reduce misalignment error described by Miller and Robins (1992). Finally, the thickness measurements are taken in the transducer, *in situ*, thereby tracking the current tendon seating. The tendon seating described above shows a need to know the tendon thickness in the transducer.

The calibration procedure presented here provides a new method to reduce variability and provide repeatable results for *in vitro* calibration methods used for human tendon force transducers. The transducer design itself further refines the utility of the buckle concept with *in situ* thickness measurements, and cylindrical geometry minimizing interference with surrounding structures.

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REFERENCES

An, K. N., Berglung, L., Cooney, W. P., Chao, E. Y. S. and Kovacevic, N. (1990) Direct *in vivo* tendon force measurement system. *J. Biomechanics* **23,** 1269–1271.

Barnes, G. R. G., and Pinder D. N. (1974) *in vivo* tendon tension and bone strain measurement and correlation. *J. Biomechanics* **7,** 35–42.

Gregor, R. J., Roy, R. R., Whiting, W. C., Lovely, R. G., Hodgson, J. A. and Edgerton, V. R. (1988) Mechanical output of the cat soleus during treadmill locomotion: *in vivo* vs *in situ* characteristics. *J. Biomechanics* **21,** 721–732.

Komi, P. V. (1990) Relevance of *in vivo* force measurements to human biomechanics. *J. Biomechanics* **23,** 23–34.

Komi, P. V., Salonen, M., Jarvinen, M. and Kokko, O. (1987) *in vivo* registration of achilles tendon forces in man, I. Methodological development. *Int. J. Sports Med.* **8,** 3–8.

Ljung, L. (1987) *System Identification: Theory for the User.* Prentice-Hall, Englewood Cliffs, NJ.

Miller, J. M. and Robins, D. (1992) Extraocular muscle forces in alert monkey. *Vision Res.* **32,** 1099–1113.

Peres, G., Maton, B., Langerit, B. and Philippe, C. (1983) Electromyographic and mechanical aspects of the coordination between elbow flexor muscles in monkeys. In *Biomechanics VIII-A* (Edited by Matsui, H. and Kobayashi, K.), pp. 455–463. Human Kinetics Publishers, Champaign.

Rempel, D., Dennerlein, J., Mote, C. D. Jr. and Armstrong, T. (1994) A method of measuring fingertip loading during keyboard use. *J. Biomechanics* **27,** 1101–1104.

Schuind, F., Garcia-Elias, M., Cooney, W. P. and An, K. N. (1992) Flexor tendon forces: *in vivo* measurements. *J Hand Surg.* **17A,** 291–298.

Sherif, M. H., Gregor, R. J., Liu, L. M., Roy, R. R. and Hager, C. L. (1983) Correlation of myoelectric activity and muscle force during selected cat treadmill locomotion. *J. Biomechanics* **16,** 691–701.

Viidik, A. (1987) Properties of tendons and ligaments. In *Handbook of Bioengineering* (Edited by Skalak, R. and Chien, S.), pp. 6.1–6.19. McGraw-Hill, New York.

Whiting, W. C., Gregor, R. J., Roy, R. R. and Edgerton V. R. (1984) A technique for estimating mechanical work of individual muscles in the cat during treadmill locomotion. *J. Biomechanics* **17,** 685–695.

Woo, L. Y. (1982) Mechanical properties of tendons and ligaments: I. quasi-static and nonlinear viscoelastic properties. *Biorheology* **19,** 385–396.