

Short communication

In vivo tendon force measurement of 2-week duration in sheep

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Accepted 23 June 2003

Abstract

Tendon tension in vivo may be determined indirectly by measuring intratendinous pressure, by using a buckle transducer or by measuring the tendon strain. All of these methods require appropriate calibration, which is highly dependent on various variables. To measure the tendon load in vivo during a period of 2 weeks in sheep, a measurement technique has been developed using a force sensor interposed serially between the humeral head and the tendon end.

Within a supporting frame, a flexion-sensitive force transducer is subjected to three-point bending stress. The load is transmitted by sutures from the tendon end through a hole in the sensor frame, orthogonal to the force transducer. In this configuration, the sensor measures the tensile force acting on the tendon, largely independent of the loading direction. The sensor was screwed to the humeral head and connected to the tendon end which was previously released from its insertion site along with a bone chip, using sutures. Connecting wires passed subcutaneously to a skin outlet about 30 cm away from the transducer.

The sensor output was linear to the measured load up to 300 N, with maximum hysteresis of 18% full scale. All sensors worked in vivo without drift over a period of up to 14 days with no change in the calibration data. Forces up to 310 N have been recorded in vivo with daily tension measurements.

This study shows that serial tendon tension measurement is feasible and allows for reliable, repeatable recording of the absolute tendon tension at the expense of tendon integrity.

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Keywords: Tendon tension measurement; Sensor, In vivo; Shoulder; Muscle force

1. Introduction

In vivo measurement of tendon or muscle tension has proved to be a challenging task. The complex muscular balancing around the shoulder joint makes it impossible to deduct from the external force measured on the limb, the tension of a single tendon crossing the joint. Therefore, tendon tension can only be determined by using an implanted transducer. While a large number of authors report methods and results from short-term measurements of less than 24 h, hardly any method was successful in recording tendon tension over a longer period of time (Gregor et al., 1988; Hodgson et al., 2001). The methods reported up to date comprise implantation of a pressure-sensitive transducer into the tendon (Hall et al., 1999; Herzog et al., 1996a,b; Korvick et al., 1996; Platt et al., 1994; Scott et al., 1997; Xu et al., 1992) passing of the tendon through a

buckle-transducer, sensitive to flexion or pressure (An et al., 1990; Dennerlein et al., 1997; Komi et al., 1987) measuring intratendinous pressure with fiberoptic cables (Erdemir et al., 2002; Komi et al., 1996) or concluding from a measured strain to the load of the tendon, by attaching a length-measuring device to the tendon (Fleming et al., 1993; Kear and Smith, 1975). All these methods leave the tendon in continuity and conclude from a second variable (pressure, strain) the tendon tension after appropriate calibration. However, these variables may vary significantly between different loading conditions, different specimens and over time, which will affect calibration to an unknown degree. Therefore, in the few reports of buckle sensor implantation over a longer period, calibration shortly after the measurements was necessary (Gregor et al., 1981, 1988).

This paper presents a new method of direct tendon force measurement in vivo over a period of 2 weeks. The method was developed in order to measure the load in the infraspinatus tendon of sheep in which tendon and muscle degeneration and regeneration were studied.

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Therefore, the main interest was to assess the decrease of residual passive tension in the muscle after tendon repair and to determine the maximal contractile force of the muscle. In preliminary experiments in two sheep, intratendinous c-shaped pressure-sensitive sensors (Hall et al., 1999) were evaluated for the purpose. These failed due to cable breakage and movement of the sensor in the tendon, despite encouraging previous *in vitro* tests. Therefore, a new method was developed using specifically assembled sensors which were interposed between the humeral head and the tendon end. In this configuration, the sensors were in series with the tendon, measuring the absolute tendon tension while eliminating the confounding variables related to calibration.

2. Material and methods

2.1. Sensor design

The sensors were assembled inside a frame (Fig. 1) that held a force transducer (F7A-3 Tendon force transducer, Konigsberg Instruments, Pasadena, CA, USA) which was subjected to bending under three-point load in the frame (Fig. 2). These force transducers were originally envisaged by the manufacturer to be used as buckle transducers (Hodgson et al., 2001), but in this case were used as flexion-sensitive force transducers, resting at each end on supports and loaded in the middle by a short pin (Fig. 1). Around the pin, two double suture loops of 10 mm length (Ethibond #3, Ethicon, Norderstedt, Germany) were passed through a hole of 2 mm diameter in the frame and connected to a steel ring of 4 mm outer diameter (Figs. 1 and 2). The frame was then screwed to the humeral head, while the ring was attached to the repair-sutures (Fig. 3). In this manner

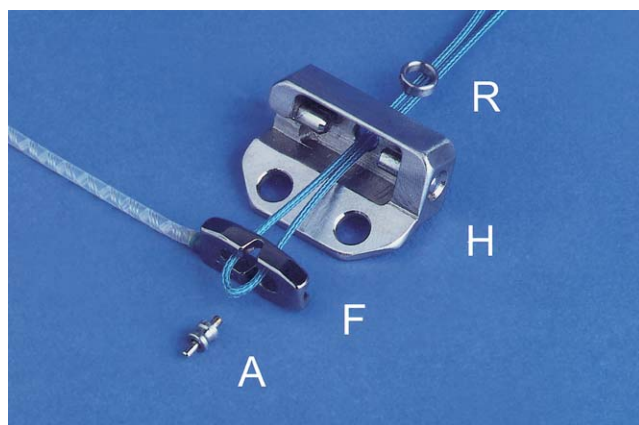


Fig. 1. The disassembled sensor: *F*: force transducer with silicone cable, *H*: Frame serving ($25 \times 10 \times 7 \text{ mm}^3$) as a holder for the force transducer (length 15 mm) with a hole through which the suture is passed, connecting the axis (*A*) with the ring (*R*), to which the tendon is later attached.



Fig. 2. The assembled sensor as it was implanted into the sheep.

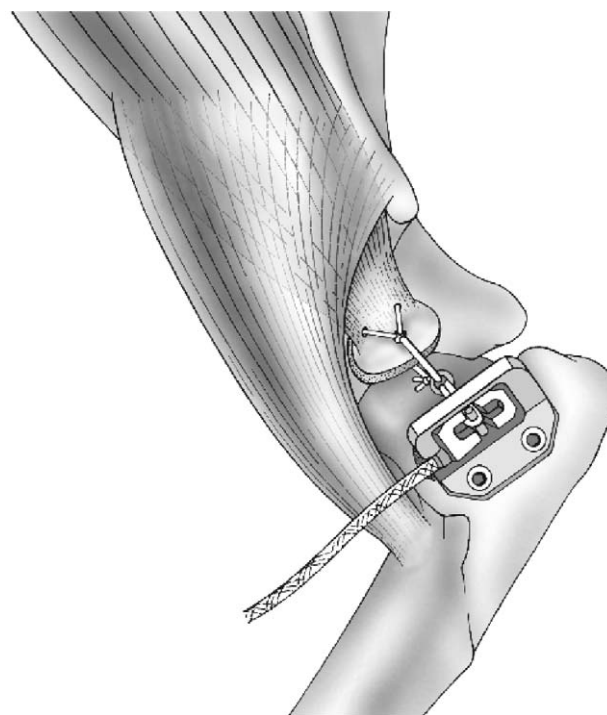


Fig. 3. Lateral view on a right sheep shoulder. Sensor implanted on the humeral head, connected to the retracted infraspinatus tendon with sutures. The tendon end is partly covered by the deltoid muscle, originating from the infraspinatus muscle and scapular spine.

the load on the sutures resulted in a three-point bending stress of the force transducer.

The connection from the force transducers consisted of a silicone tube, containing three spiraled steel wires that were leads from the strain-gauge half-bridge circuit. The complimentary half-bridge with DC amplifier and 5 V supply were in an external unit. An operational amplifier of Typ OP-77 (Precision Monolithics Inc., Norwood, MA, USA) with ultra-low offset voltage and with an adjusted gain of about 100 was used. The analogue output signal from this unit (max. $\pm 1 \text{ V}$) was

then digitized using a converter (DAQ Card, National Instruments, Austin, TX, USA) and recorded on a computer (software LabVIEW™, National Instruments, Austin, TX, USA) at a rate of 30 Hz.

2.2. Calibration

2.2.1. Straight loading

The sensors were routinely calibrated by loading with free-hanging weights in 50 N steps up to a force of 200 N. Hysteresis was evaluated by loading and unloading the sensor on a material testing machine (model 4204, Instron Limited, High Wycombe, UK) at a rate of 7 N/s up to 300 N. Two transducers were loaded up to 400 N to explore the maximum possible range of loading.

2.2.2. Loading at an angle

Influence of the loading angle on the sensor output was measured by tilting the transducer frame in the horizontal and in the vertical plane, in order to simulate the alteration in loading direction of the sensor when the humerus is flexed or rotated, respectively. A constant load of 20 N was applied and the output voltage (43 mV) measured at intervals of 10° tilt, between –50° and 50° for the vertical plane and between –30° and 30° for the horizontal plane, which represents the physiologically maximal range of motion plus 20° in each direction. The tests were repeated also with loads of 50 and 70 N.

2.3. In vivo experiments

The experiments were conducted with 8 Swiss Alpine Sheep. The mean age of the sheep at the beginning of the study was 3.75 years, with a range from 3 to 5 years. The mean weight was 48.4 ± 3.6 kg. All experiments were conducted according to the Swiss law of animal welfare and use for animal experiments (*permission number ZH 160/99*).

2.3.1. Implantation

After surgical exposure through a curved incision over the shoulder joint under standard general anesthesia (Akens et al., 2001), the infraspinatus muscle was released by osteotomy of the greater tuberosity. In two sheep, the sensors were implanted immediately after tendon release, in six sheep, the sensors were implanted after a period of 40 weeks to mimic chronic tendon retraction. The sensors were attached with two 3.5 mm cortical bone screws to the humeral head, so that the connection-ring faced the free tendon end. Through the tendon end and the bone chip, two Ethibond #6 sutures were passed in a figure-of-eight configuration and connected to the steel ring of the sensor with a surgical knot (Fig. 3). To prevent spontaneous healing, the end of the tendon was covered with a silicon rubber tube of 5 cm length (Silicone Penrose drain tube, 12 mm,

Fortune medical instrument corp. Taipei, Taiwan), which was closed at its distal end. In order to reduce the risk of infection, the sensor cable was passed through a subcutaneous channel of 30 cm to the sheep's neck, where it perforated the skin. It was intraoperatively connected to the measurement equipment. Under continuous recording of the tension, the repair was performed at either high tension (50–80 N, 4 sheep 40 weeks after tendon release) or low tension (0–20 N, 2 sheep with immediate repair after release, 2 sheep 40 weeks after tendon release), measured in 0° of rotation, abduction and flexion of the shoulder. The length of the connecting sutures depended on the distance of tendon retraction, and ranged between 5 mm (no retraction) and 30 mm (far retraction).

2.3.2. In vivo force measurements

For two weeks immediately after surgery, the sheep were prevented from lying down by the use of loose suspension belts. A ball, attached to the sheep's claw (Gerber et al., 1999), prevented full weight bearing. This was necessary in order to reduce the risk of repair failure. Between the measurements, the connection line was stored in a small cloth bag, attached to the sheep's back. The tension on the sensor was recorded each day during a phase of resting and during maximal activity (feeding), for a duration of at least 10 min. Through gentle passive movement of the leg through the examiner, the minimal muscle tension in maximal abduction with external rotation was evaluated daily.

2.3.3. Sensor removal

Either after failure of the sutures connecting the sensor to the tendon or after 14 days, the sensors were removed. During the surgical intervention the sensor wires were connected to the measuring and recording equipment. The greater tuberosity and humeral head with sensor and tendon were exposed and the sensor zero-output checked after disconnecting the tendon from the sensor. The sensor and tendon covering tube were subsequently removed. After removal of the sensor, calibration up to 200 N was repeated to check whether it had remained unaltered. Thereafter, the sheep were utilized for further biological observations, reported separately.

3. Results

3.1. Calibration

3.1.1. Straight loading

Typical hysteresis curves in straight loading from 10 N up to upper limits of 150 and 300 N, respectively, are displayed in Fig. 4. Static loading during one minute with 400 N resulted in slight deformation of the buckle. However, even though deformed, the sensitivity

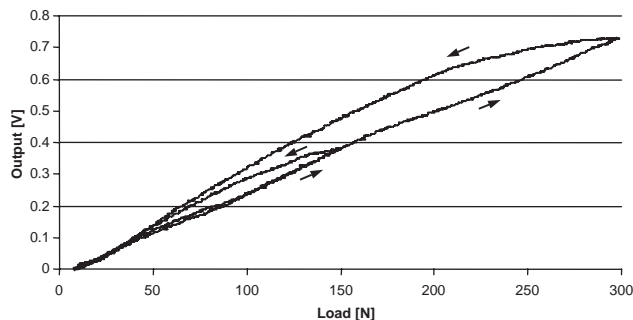


Fig. 4. Hysteresis of the sensor, loaded once from 10 to 150 N and once from 10 to 300 N.

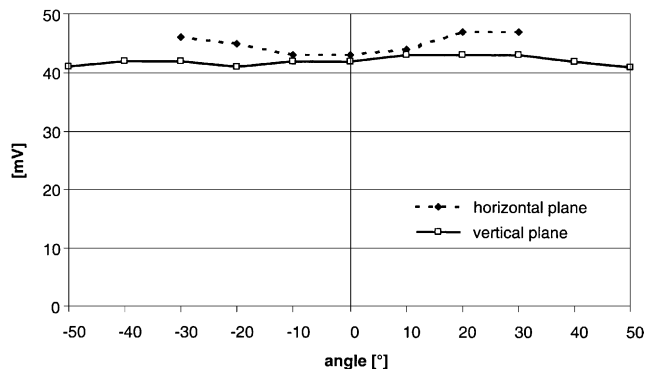


Fig. 5. Output of the sensor, loaded with a hanging load of 20 N at different angles with increments of 10° up to 30° in the horizontal and up to 50° in the vertical plane, corresponding supra-maximal leg flexion and abduction, respectively.

remained unaltered. Although the buckle transducer showed a non-linear behavior above a load of about 200 N, linearity as well as hysteresis proved to be within $\pm 10\%$ of the upper limit (300 N).

3.1.2. Loading at an angle

The calibration curves displaying the output voltage when the sensor was loaded at an angle to the suture in both planes (horizontally and vertically) are displayed in Fig. 5. Within the physiological range of motion during suspension in the barn (humerus flexion up to 10° , rotation up to 30°), the maximal errors were 2.3% along the horizontal plane (flexion) and 2.4% along the vertical plane (rotation or abduction). Maximal error of the sensor (beyond the range of shoulder motion) was 8.5% and 4.8%, respectively. All errors related to a full scale of 20 N. Further tests using higher loads showed similar, but slightly smaller percentages of error.

3.2. In vivo measurements

No difficulties with wound healing or infection occurred. All the sensors and their connection lines stayed intact for the entire period of implantation. Suture failure between sensor and tendon due to fraying

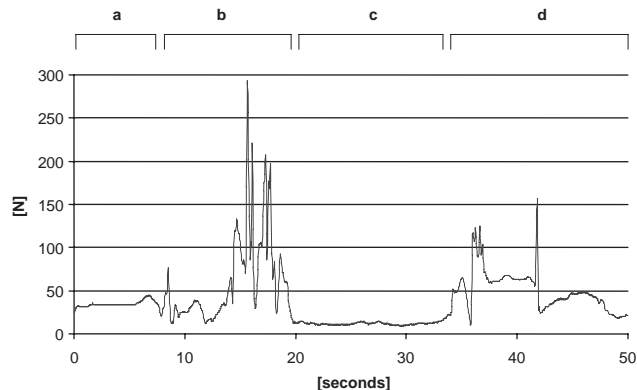


Fig. 6. A typical in vivo-measurement sample on first postoperative day over a period of 50 s from a sheep with a ball on the claw and prevented from laying down with belts. Subsequent time sequences: (a) undisturbed standing; (b) sheep refuses actively to be examined at the claw by the investigator; (c) sheep is left quiet; and (d) sheep is rewarded by feeding with bread.

of the suture at the connecting ring, resulting in no further tension measurement and early sensor removal, occurred postoperatively at day 1, 4, 6 and 11 in one animal each. The sutures in the other four animals stayed intact for the entire period of 14 postoperative days without fraying of the sutures at the connecting ring or knot loosening.

Throughout the period of force measurements, the recording of muscle tension appeared to correspond well with the sheep's activity (Fig. 6). The maximal tension on the tendon registered was 310 N. By bringing the sheep's leg gently in maximal abduction and external rotation, the minimal passive tension was recorded, which was found to be about zero, 48 h after implantation.

At sensor removal after 14 days of measurement, following intraoperative detachment of the tendon from the sensor, a negative shift of 5–15 N was found, which had been observed as drift during the in vivo measurements, starting seven days after implantation. After removal of the soft tissue newly formed around the sensor, the drift immediately disappeared. Sensitivity had not altered within the accuracy of measurement when hanging weights to the sensors, and no macroscopic external alteration to the sensor shapes have been noticed.

In Fig. 6, a typical measurement curve from a sheep in the suspension apparatus at the first postoperative day is displayed, comprising resting tonus, movement when refusing to be examined, and movement during feeding.

4. Discussion

Various methods have been developed during the last decades with the common goal to measure the tensile load on tendons in vivo. To our knowledge, all of these techniques leave the tendon in continuity and attach or insert a sensor to the tendon or the muscle, which is

deformed when the tendon is loaded. After appropriate calibration, the load on the tendon is derived from the sensor output.

The main concern thereby is that the sensitivity will be affected by several mechanisms: In short term applications, by bending of the tendon due to limb movement (Herzog et al., 1996a; Holden et al., 1995; Rupert et al., 1998), by the loading rate (Herzog et al., 1996a), by the position of the sensor in the tendon (Erdemir et al., 2002), and by the elasticity of the tendon tissue and tendon thickness (An et al., 1990; Dennerlein et al., 1997). In long-term applications, by angular and longitudinal movement of the sensor in the tendon (Erdemir et al., 2002) and by adaptation of the tendon tissue to the load (Dennerlein et al., 1997; Xu et al., 1992). Adjacent structures to the tendon such as contracting muscles may impinge and compress pressure-sensitive sensors in the tendon, influencing the sensor output to an unknown degree. Calibration will always be specific for the tendon investigated, which requires either intraoperative calibration at implantation, or exact measuring of the tendon diameter. In summary, parallel measuring techniques which leave the tendon in continuity may be of great value when studying relative tension in the tendon to recognize loading patterns such as during gait, but will be problematic when an absolute value of tension is required.

Commercially available intra-tendinous c-shaped pressure sensitive sensors were tested in vivo in two preliminary experiments but failed due to cable breakage and relative movement of the sensor on the tendon. Therefore, a new method for in vivo long-term tendon force measurements was developed. This study demonstrated that it is feasible to measure the tendon load in series with the tendon over a period of 2 weeks. The main advantages of the herewith introduced serial measurement technique is that no intraoperative calibration is necessary and possible hysteresis and calibration errors are known and derive from the sensor design itself. The design which transmits the load with sutures through a hole in the sensor frame, makes the sensor very insensitive to the direction of loading (error maximally 8.5%). As a result, a relatively large hysteresis (18% at loading up to 300 N) has been found, probably related to friction of the suture with the frame and of the pin inside the sensor. Apart from some mechanical influence by in-growth of soft tissue into the sensor, no electronic drift or alteration of the sensitivity has been found during the maximal period of implantation of 2 weeks. To our knowledge, no other studies report comparable measurements.

During experimental loading, a linear signal output within the accuracy necessary for practical use has been found for the entire range of muscle load.

The obvious main drawback of the method is that the tendon is not kept in continuity and hence does not

represent the physiological situation, and the repair with sutures limits the maximal load on the tendon. Also a possible mismatch of the original length of the muscle unit due to the interposition of the transducer between the excised tendon and anchored transducer may influence the muscle tension. Therefore, the results will not represent the situation in healthy sheep, but the maximal or minimal tension on the muscle being studied. Pull-out of the sutures from the tendon as known from previous studies (Gerber et al., 1999) was successfully prevented by release of the tendon with a bone chip, which enhanced grasping the tendon end with sutures. This technique seems to be advantageous also for other possible applications such as measurement of ligament tension. However, possible changes in joint kinematics must be evaluated for each application. To reduce the risk of suture breakage, the sutures may be replaced in future studies with now available modern high-performance suture material and any sharp edges on the sensor must be meticulously smoothed. Because of the size of the available force transducer, the whole implant became correspondingly large, which however was well tolerated by the animals and hardly palpable through the skin. Still, the invasive nature of this measurement technique will most likely limit the application on experimental animals.

The force transducers in the sensor were not designed for the application they were used in this work and were certainly subjected to higher bending load than if used as buckle transducers. However, even though minimal plastic deformation did occur in vitro once, drift or alteration in sensitivity was never found. As it proved to be very demanding to implant electronic parts without cable breakage and with water immersion for over 2 weeks, these implants which were designed for long-term implantation (Hodgson et al., 2001) seemed to be the best suited.

The goal of the study was to establish the procedure of serial tension measurement using the most simple and reliable approach possible. Based on the positive results obtained, smaller sensors capable of measuring higher loads with specifically designed force transducers may be developed, exhibiting even less measurement error and resistance to in-growth of soft tissue.

Acknowledgements

This study was funded by the Swiss National Science Foundation.

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