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Clinical Biomechanics 19 (2004) 433–447
CLINICAL BIOMECHANICS
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Review

Strain and force transducers used in human and veterinary tendon and ligament biomechanical studies

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Received 20 January 2004; accepted 20 January 2004

Abstract

Biomechanical studies often aim at determining the contribution (in terms of load or strain) of a tendon or ligament in posture, gesture or locomotion. To this end, many transducers have been developed since 30 years. These devices implanted within or attached to the inside of the tendon or ligament must be compliant enough to measure in vivo the tissue load or strain without interfering with the movement of man or animals. They can be transducers with variation of electrical resistance (liquid metal strain gauge, buckle transducer, implantable force transducer and pressure transducer), variation of magnetic field (Hall effect transducer) and variation of light flow (optic fibre). Their use requires surgery in order to implant them and it is limited in time because of their invasive character and the development of fibrous healing reactions. Besides, the transducer dimensions and its position in the tendon can influence the transducer output signal. Moreover, the latter may not reflect the behaviour of the tendon as a whole but only locally. In addition, a calibration is required in order to convert the output signal into a strain or a force. In animals, this calibration is generally made by a post-mortem procedure on dissected anatomical specimens; in man, an indirect calibration procedure using inverse dynamic calculations is generally performed. However, the calibration conditions cannot reproduce exactly the in vivo conditions. So far, only invasive transducers have allowed to measure strain or force in tendons with all constraints and limits mentioned above.
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Keywords: Transducer; Strain; Force; Tendon; Ligament

1. Introduction

The mechanical behaviour of ligaments and tendons in situ and in particular the tension generated by a muscle or the elongation undergone by a tendon and ligament during motion is of particular concern in biomechanics. The force or strain that is developed within these structures can be measured directly via a compliant transducer attached to the tendon or ligament, or indirectly using inverse dynamic calculations (non-invasive method). For the direct approach only invasive devices are applicable. This means that the transducer must be implanted within the tendon or ligament and it must be compliant enough to measure the tissue tension without interfering with its normal use during movement. It also has to be able to measure in vivo strains (about 1–20%) (Howe et al., 1990; Jansen et al., 1993b, 1998; Kear and Smith, 1975; Keegan et al., 1991, 1992; Lochner et al., 1980; Riemersma et al., 1988b; Stephens et al., 1989; Van Weeren et al., 1992) or heavy loads (hundreds or thousands of Newton) (Finni et al., 1998, 2000; Fukashiro et al., 1993; Gregor et al., 1987, 1988; Hasler and Herzog, 1997; Holden et al., 1994; Jansen et al., 1993a; Komi et al., 1984; Komi, 1990; Platt et al., 1990; Riemersma et al., 1985, 1988a; Roberts et al., 1994; Sherif et al., 1983).

In this bibliographic review of the main transducers used in vitro and in vivo in human and veterinary medicine, the description and physical principles of the transducers, theirs conditions of use and lastly their respective limitations, will be successively presented.

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0268-0033/$ - see front matter © 2004 Elsevier Ltd. All rights reserved.
doi:10.1016/j.clinbiomech.2004.01.008

2. Description and physical principles of strain or force transducers

Different types of transducers are used to measure strains or loads within the tendons or ligaments. Three types of transducers can be defined according to physical principles on which they are based: electrical resistance for liquid metal strain gauges and other extensometry gauge transducers (buckle transducer, implantable force transducer and pressure transducer), variation of magnetic field for Hall effect transducer, variation of light flow for optical strain-measuring system (Toux, 1990).

2.1. Variation of electrical resistance

The liquid metal strain gauge (LMSG) consists of one (Brown et al., 1986; Glos et al., 1993; Keegan et al., 1991; Keegan et al., 1992; Riemersma and Lammertink, 1988; Riemersma et al., 1988a; Van Weeren et al., 1992) or two (Lochner et al., 1980) compliant capillaries (in a soft polymer as silicone), filled with a liquid metal (liquid at room temperature like mercury or a mixture of gallium and indium). The open tube ends are closed by platinum wire to which two lead wires are soldered (Fig. 1). The liquid metal column is enclosed in the tube under a slight overpressure. This slight overpressure prevents subpressure during straining, which might predispose to water penetration through the slightly permeable silicone tube wall (Riemersma and Lammertink, 1988). The basic functioning principle of the LMSG (Table 1) is the measurement of strain-induced electrical-resistance-change of the liquid metal capillary that is contained within the soft polymer encasement. The electrical resistance ($R = \rho L/A$) of the LMSG is a function of the metal column length (L) and the column cross-sectional area (A) (ρ being the resistivity of the liquid metal, independent of L and A). The more the gauge is stretched, the more the liquid metal column's length increases and its cross-sectional area decreases, and the more its electrical resistance increases. The gauge is part of a circuit with constant current. Thus, the variation of electrical resistance induces a variation of the voltage at the output of the gauge, which represents the variation of gauge strain and of tissue (e.g. tendon) strain equipped by this type of transducer. The electrical resistance variation of the gauge (dR/R) is measured by a Wheatstone bridge (Brown et al., 1986; Glos et al., 1993; Lochner et al., 1980; Riemersma and Lammertink, 1988; Riemersma et al., 1988a; Stone et al., 1983). The LMSG is implanted with some prestrain to ensure that any change in length of the tendon will be recorded (Riemersma et al., 1985).

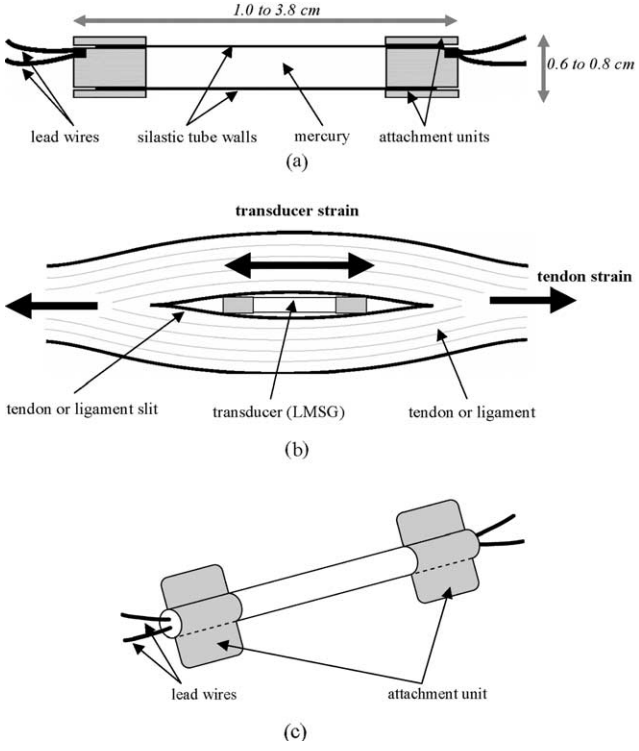

Fig. 1. Schematic diagram of the liquid metal strain gauge (LMSG) (Komi et al., 1996; Marlkof et al., 1998b; Platt et al., 1990, 1994; Stephens et al., 1989). (a) Constitution and dimensions of the LMSG. (b) Position of the transducer within the tendon or the ligament. (c) General schematic view of the LMSG.

The extensometry gauges transducers include three types of transducers according to their design: the buckle transducer, the implantable force transducer and the pressure transducer. All of these devices are equipped with strain gauges which deform when the device undergoes a strain. When the strain gauges attached to the surface of the transducer deform, their resistivity changes and the output signal (electrical resistance) is modified.

(a) *The buckle transducer* consists of an outer frame and removable crossbars that are seated upon the frame. This type of transducer has the form of a rectangular or oval buckle (Barnes and Pinder, 1974; Fukashiro et al., 1993; Herrick et al., 1978; Komi et al., 1987; Komi, 1990; Lewis et al., 1982; Salmons, 1969; Sherif et al., 1983; Silver and Rossdale, 1983) or an E-form (Komi et al., 1984; Komi et al., 1987; Walmsley et al., 1978) (Fig. 2a). A loop of ligament or tendon is drawn through the frame (Fig. 2b) and secured in position by the narrow cross-piece. This design was modified a little by some authors. Barry and Ahmed (1986) have developed a I-form buckle transducer which is laid on the top of the tissue fibres under study and parallel to their length and is maintained on the tissue by a small clip passed underneath the tendon or ligament. The buckle transducer works by slightly deflecting the normal configuration of a load-carrying flexible element (crossbar and frame). Any tension in a tendon or ligament stretches it and causes it to straighten itself to its normal orienta-

Table 1
Characteristics of different transducers used in vitro or in vivo in tendons or ligaments

Transducer family	Physical principle	Applications	Dimensions (mm)	Calibration	Range measurement (according the studied tendon and movement)	Precision	Reliability, repeatability	Ref.
1.1.1. Liquid metal strain gauge	Variation of electrical resistance of liquid metal column	Strain measurement (sometimes force measurement)	10 to 38×6 to 7.6	• Traction directly on the isolated transducer (in vitro) • Traction on the dissected tendon equipped with the transducer (in vitro or post-mortem)	Up to 11%	+/−0.1% of the elongation		Brown et al., 1986; Jansen et al., 1993a,b 1998; Keegan et al., 1991, 1992; Lochner et al., 1980; Riemersma and Lammertink, 1988; Riemersma et al., 1985, 1988a, 1988b, 1996a,b Stone et al., 1983; Van Weeren et al., 1992
1.1.2.a Buckle transducer	Variation of electrical resistance of strain gauges bonded on transducer surface Measurement of the deformation of the transducer crossbar and frame	Force measurement	9 to 16×5 to 7.5×0.5 to 2.5 or 34 to 38×20 to 25×13.5	• Application of the force known directly on the isolated transducer • Traction on the dissected tendon equipped with the transducer or electrical nerve stimulation of the muscle (in vitro or post-mortem) • Use of different calibration apparatus to evaluate the tendon force from joint moment data (in vivo in the man)	Up to 4000 N			Barnes and Pinder, 1974; Barry and Ahmed, 1986; Gregor et al., 1987, 1988, 1991; Hall et al., 1999; Komi et al., 1987; Komi, 1990, Lewis et al., 1982; Sherif et al., 1983; Silver and Rossdale, 1983; Walmsley et al., 1978

Table 1 (*continued*)

Transducer family	Physical principle	Applications	Dimensions (mm)	Calibration	Range measurement (according the studied tendon and movement)	Precision	Reliability, repeatability	Ref.
1.1.2.b Implantable force transducer	Variation of electrical resistance of strain gauges bonded on transducer surface Measurement of the transversal deforming force	Force measurement (sometimes strain measurement)	4 to 5×1.2 to 1.8 or 5×2 to 2.5×1 or 8.5 to 11. 5×6.5 to 8×2.4 or 11.2×6.4×2.5	• Traction on the dissected tendon equipped with the transducer or electrical nerve stimulation of the muscle (in vitro or post-mortem) • Use of a load cell by reproducing the anatomical configurations of the bones (in vitro or post-mortem)	Up to 2500 N		• Coefficients of variation for resultant force after several repeated measurements: 0.038 to 0.111 • Coefficients of variation for strain after several repeated measurements: 0.209–0.342 • Coefficients of variation for output voltage according to deep or proud transducer placements: 0.156–0.359 • Coefficients of variation for resultant force after remove and re-implantation of the transducer: 0.001–0.713 (same location) or 0.021–2.028 (other location)	Butler et al., 1989; Fleming et al., 1994, 1996, 2000; Glos et al., 1990, 1993; Hall et al., 1999; Hasler and Herzog, 1997; Herzog et al., 1996a,b; Kear and Smith, 1975; Korvick et al., 1996; Marlkof et al., 1998b; Platt et al., 1990, 1994; Roberts et al., 1994; Rupert et al., 1998; Xu et al., 1992
1.1.2.c Pressure transducer	Variation of electrical resistance of strain gauges bonded on transducer surface Measurement of the transversal deformation force	Force measurement	3.5 to 4.5×0.9 to 1.9	• Use of a load cell by reproducing the anatomical configurations of the bones (in vitro or post-mortem)	Up to 700 N		• Variation of the sensibility with the flexion ankle of the knee: 20–100 mV/N	Glos et al., 1990; Holden et al., 1994, 1995
1.2. Hall effect transducer	Variation of magnetic field	Strain measurement (sometimes force measurement)	5 to 7×2	• Use of a dial-type Vernier calliper (in vitro)	Up to 19%	0.2–0.5% according the transducer model		Fleming et al., 1994; Howe et al., 1990; Marlkof et al., 1998a; Stephens et al., 1989

Table 1(*continued*)

Transducer family	Physical principle	Applications	Dimensions (mm)	Calibration	Range measurement (according the studied tendon and movement)	Precision	Reliability, repeatability	Ref.
1.3 Optic fibre	Variation of light intensity detected with a photodetector	Force measurement (sometimes strain measurement)	0.2–0.5	• Traction on the dissected tendon equipped with the transducer or electrical nerve stimulation of the muscle (in vitro or post-mortem) • Use of different calibration apparatus to evaluate the tendon force from joint moment data (in vivo in the man)	Up to more 6000 N		• Variation induced by the fibre movement: $P < 0.0000$ lou less 27% maximum load • Influence of loading rate: less 17% maximum load • Maximum root-mean-square (RMS) error within any specimen: 26% maximum load	Erdemir et al., 2001, 2002; Finni et al., 1998, 2000; Komi et al., 1996

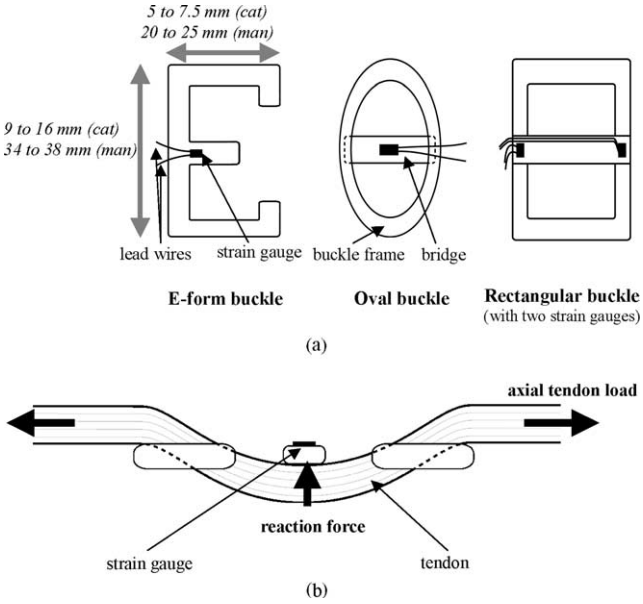

Fig. 2. Schematic diagram of the buckle transducer (Keegan et al., 1991, 1992; Roberts et al., 1994; Stone et al., 1983). (a) Schematic view and dimersions of the buckle transducer (top view of the transducer). (b) Position of the transducer within the tendon or the ligament (side view of the transducer).

tion, thereby deforming the crossbar and frame. Strain gauges are mounted on the frame and when the frame is deformed by the tension in the tendon, this strain is detected by the strain gauges (Table 1 and Fig. 2). The strain gauges (of reduced rigidity compared to that of the tendon) produce an output signal (electric voltage) proportional to the tension developed within the device (Herrick et al., 1978; Lewis et al., 1982; Salmons, 1969). So that the device does not injure the tendon or the ligament and that the tendon shortening is minimised, the shape of the crossbar and the dimensions of the transducer must be adapted to the width of the tendon (Fig. 2a) (Gregor et al., 1991; Komi et al., 1987; Komi, 1990; Lewis et al., 1982).

(b) *The implantable force transducers* (*IFT*) that have been used in soft tissues research employ two different mechanical modes of action: one which is based on three-point-bending and another that is subjected to two-point compressive loading during tissue loading (Hall et al., 1999). The three-point bending transducer consists of a slightly curved miniature steel bar (Fig. 3a) (Butler et al., 1989; Glos et al., 1990; Glos et al., 1993; Herzog et al., 1996a; Kear and Smith, 1975; Korvick et al., 1996; Xu et al., 1992). The two-point compressive transducer consists of a C-shaped tube with a longitudinal slit (Fig. 3a) (Fleming et al., 1994; Fleming et al., 2000; Hall et al., 1999; Marlkof et al., 1998b). The strain gauges are attached, in the case of curved bar, both on top and bottom surfaces (Butler et al., 1989; Glos et al., 1990; Glos et al., 1993; Herzog et al., 1996a; Xu et al., 1992) and in the case of the split tube, on the edge

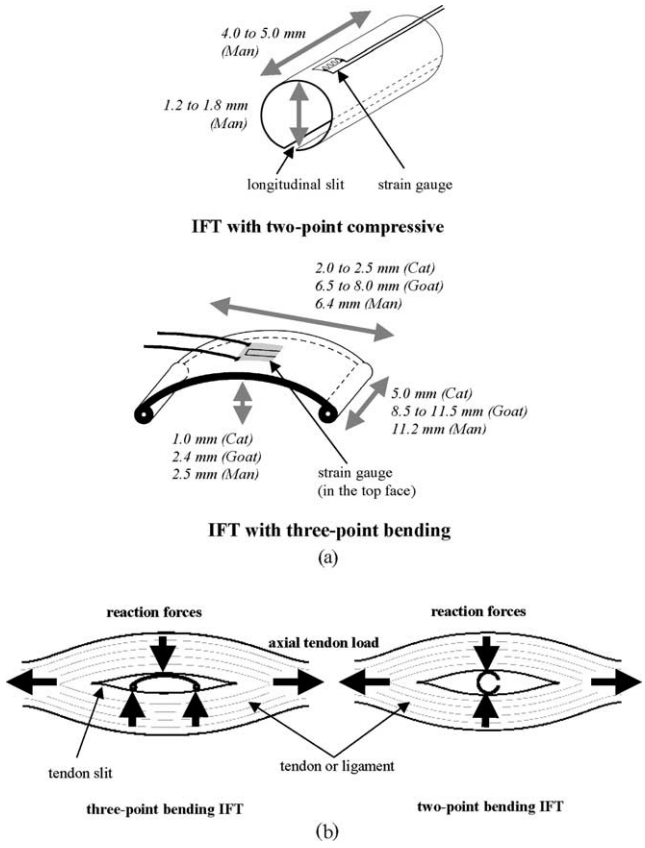

Fig. 3. Schematic diagram of the implantable force transducer (IFT) (Finni et al., 2000; Lewis et al., 1982). (a) Schematic view and dimensions of the IFT. (b) Position of the transducer within the tendon or the ligament.

opposite to the slit (Fleming et al., 1994; Hall et al., 1999) (Fig. 3b). Another design is used in some tendon studies and consists of a base-plate with a central deflection leaf; two strain gauges are bonded to the leaf, one on either side (Platt et al., 1990; Platt et al., 1994). Once positioned within a tendon or a ligament, the IFT displaces the tissue fibres in contact with it slightly from their normal axial alignment due to its presence. The fibre displacement causes a transverse force creating surface device strains that can be measured and calibrated against the axial force in the tissue (Fleming et al., 2000; Glos et al., 1993; Hall et al., 1999; Sherif et al., 1983; Xu et al., 1992). Indeed, the resulting strain-gauge signal is proportional to the transversely deforming force (and through an appropriate calibration, to the tension in the tendon or ligament fibres) (Marlkof et al., 1998b; Xu et al., 1992). The strain of the device can then be evaluated by the strain of gauges fixed on the device (Table 1) (Glos et al., 1993; Salmons, 1969; Xu et al., 1992). All the reaction forces are supposed to be normal to the surface of the IFT because of absence of friction forces between the transducer surface and the tissue fibres (assumption made because of the strong water content of the tendons) (Xu et al., 1992). The IFT must be a smaller as possible in order not to damage the tissue itself and alter its mechanical behaviour. It must also have sufficient sensitivity to allow the detection of the small transverse compressive forces exerted by the fibres against the device. The curved bar or the slit tube constituting the transducer should thus not be too stiff nor too flexible lest the signal is contaminated with a high level of noise, rendering the analysis irrelevant. The sensitivity of this type of transducer can be increased or decreased by modifying its geometry (curvature, length, thickness) or by choosing a material with varying degree of compliance (Butler et al., 1989; Xu et al., 1992). The transducer width can be adapted to measure either the average tissue force or the force in a specific fraction of the tissue (Xu et al., 1992).

(c) *The pressure transducers* used within tendons or ligament are commercially available transducers modified for in vivo force measurements (Glos et al., 1990; Holden et al., 1994; Holden et al., 1995). They are miniature transducers (3.5–4 mm diameter, 0.5–1.5 mm thick) on which a dome-shaped Teflon button is bonded on the diaphragm (1.9 mm diameter, 0.4 mm height) (Glos et al., 1990; Holden et al., 1994; Holden et al., 1995). The implanted device causes the tissue fibres passing over it to be displaced slightly from their usual paths, producing a transverse pressure. The transverse force is proportional to the axial force exerted on the tendon or ligament fibres and is detectable against the strain-gauged transducer diaphragm (Table 1) (Holden et al., 1994; Holden et al., 1995).

2.2. Variation of magnetic field

The Hall effect strain transducer is a steel tube (covered with Teflon) coupled to a Hall generator. A magnetic core or a metal wire charged magnetically (magnet) slides freely within the tube. One end of the tube is bonded to a fixation pin and the distal end of the magnet is similarly armed. The fixation pins serve to attach the transducer to the tendon or ligament (Fig. 4a) (Glos et al., 1993; Howe et al., 1990; Stephens et al., 1989). Marlkof et al. (1998a) used a slightly different model (“differential variable reluctance transducer”), a small cylindrical electrical coil and a metal rod that can slide within the coil. These strain transducers are based on the principle of the Hall effect, in which the transducer signal (the voltage output of a Hall generator) is proportional to the displacement of magnetic core and the strength (proximity) of a magnetic field (Table 1). As the tendon stretches, the distance between the fixation pins increases and the magnet displaces relative to the Hall generator. Over a short distance, the voltage output is proportional to the distance (Glos et al., 1993; Howe et al., 1990; Stephens et al., 1989).

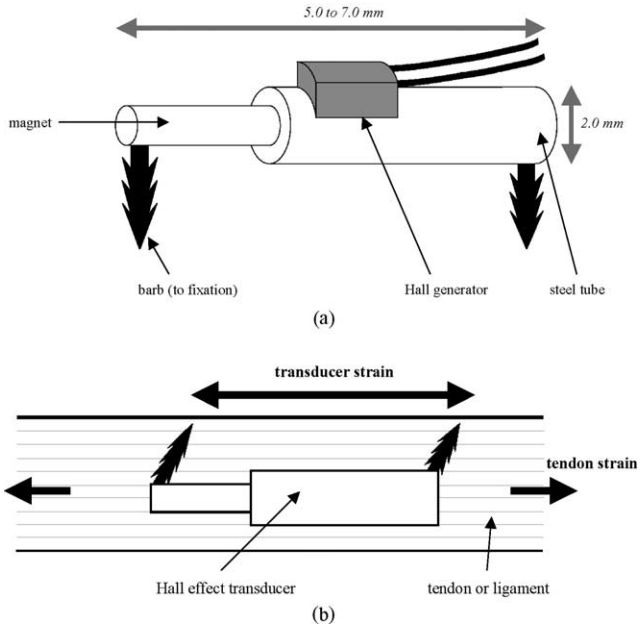

Fig. 4. Schematic diagram of the Hall effect transducer (Finni et al., 2000; Herzog et al., 1996b; Salmons, 1969). (a) Schematic view and dimensions of the Hall effect transducer. (b) Position of the transducer within the tendon or the ligament.

2.3. Variation of light flow intensity

An optic fibre has been used as a transducer of tendinomuscular forces in the Achilles tendon (Erdemir et al., 2001, 2002; Finni et al., 1998, 2000; Komi et al., 1996) and in the patellar tendon (Finni et al., 2000). It is a commercial optic fibre (with a diameter of 265 or 500 μm to 0.5 mm) covered by a plastic buffer. The optic fibre is inserted through the skin then through the entire cross-section of the tendon using a needle. After this insertion, both ends of the optic fibre are carefully prepared for receiving and transmitting light. A transmitter–receiver unit (consisting of a light emitting diode and a photodiode receiver) is attached to the ends of the optic fibre (Finni et al., 1998, 2000). The light signal travels in the core of the optic fibre and returns to the unit for conversion into an analog signal. The measurement with this transducer is based on light flow intensity modulation of the optic fibre by mechanical modification of the geometric properties of the fibre. When the optic fibre is bent or compressed within the tendon due to the compressive stresses which occur when the tendon fibres are loaded, the light flow reaching the receiver (and therefore the voltage output) is modified since the optic fibre changes the conditions of propagation of the light (Table 1) (Erdemir et al., 2002; Finni et al., 2000; Komi et al., 1996; Toux, 1990).

These three types of transducers are used to measure either strains or forces: LMSG and Hall effect transducers are generally used as strain transducers whereas buckle transducers, implantable force transducers, pressure transducers and optic fibres usually allow load measurements (Table 1). However, the LMSG can also be used as a load transducer (Jansen et al., 1998; Riemersma and Lammertink, 1988; Riemersma et al., 1988a), and the IFT as a strain transducer (Glos et al., 1993; Kear and Smith, 1975; Marlkof et al., 1998a) (see Section 3.3).

3. Conditions of use of the transducers

3.1. General goals of the studies using strain or force transducers

The different transducers presented are used in tendinomuscular and articular biomechanics in order to record tendon or ligament strains and tendon or ligament loads both in vitro and in vivo, in order to estimate the contribution of every tendon or ligament. In vivo tendon strains and loads are studied during movement of animals (walk, trot, gallop, treadmill locomotion, jumping) or man (walk, jumping, cycling, running), sometimes in conjunction with muscular activity (EMG), kinematics or force plate measurements (Tables 1 and 2). The aim of several locomotion research studies which used these invasive transducers, was to correlate the measured strain pattern with the phase of the stride (Jansen et al., 1993a, 1998; Keegan et al., 1991, 1992; Lochner et al., 1980; Riemersma et al., 1985, 1988a, 1988b , 1996a; Van Weeren et al., 1992) or to estimate the modifications of the strain pattern with variations of hoof angle (Lochner et al., 1980; Riemersma et al., 1996b) or ground characteristics (Riemersma et al., 1996a).

3.2. Implantation and position of the transducers

Several transducers (LMSG, extensometry gauge transducer, pressure transducer, optic fibre) have to be implanted within the midsubstance of the tendon or ligament, either through a longitudinal incision made in the tissue (LMSG: Jansen et al., 1993a, 1998; Lochner et al., 1980; Riemersma et al., 1985, 1996a, 1996b; Van Weeren et al., 1992; implantable force transducer: Butler et al., 1989; Fleming et al., 1996; Glos et al., 1990, 1993; Hall et al., 1999; Herzog et al., 1996a, 1996b; Korvick et al., 1996; Marlkof et al., 1998a; Platt et al., 1990, 1994; Roberts et al., 1994; Rupert et al., 1998; Xu et al., 1992; Hall effect transducer: Fleming et al., 1996; pressure transducer: Glos et al., 1990; Holden et al., 1994, 1995), with a needle (Hall effect transducer: Stephens et al., 1989; optic fibre: Erdemir et al., 2002; Finni et al., 1998, 2000; Komi et al., 1996) or with arthrocopic instruments (Fleming et al., 2000; Hall et al., 1999; Howe et al., 1990;

Table 2
In vivo use of different transducers in tendons or ligaments

Transducer family	In vivo use of the transducer			Aims of experiment	References	Advantages	Disadvantages
	Species	Tissue	Gait of locomotion or movement studied				
1.1.1 Liquid metal strain gauge (LMSG)	Horse	SDFT, DDFT, SL, DCL, LDET	Standing, walk and trot	Determination of the tendon strains and loads during the locomotion, influence of shoeing, ground characteristics, support bandaging and hoof wall angle on the tendon strains, comparison with strain patterns calculated from kinematic data	Jansen et al., 1993a; Jansen et al., 1998; Keegan et al., 1991, 1992; Lochner et al., 1980; Riemersma et al., 1985, 1988a,b; 1996a,b; Van Weeren et al., 1992	• Small size	• Brittleness during handling and implantation • Limited service life (slow oxidation of the liquid metal) • Temperature influence on the signal output • Risk of release of toxic substance (liquid metal)
	Rabbit	Achilles tendon	Slow hopping	Development of a pilot in vivo experiment to measure dynamic tendon forces	Brown et al., 1986		
1.1.2.a Buckle transducer	Horse	SDFT and common extensor tendon	Walk	Determination of the tendon load and bone strain during the locomotion	Barnes and Pinder, 1974; Silver and Rossdale, 1983	• Rigid • Easy to manufacture in a size adapted to the tendons • Simple recording equipment	• Modification of course of the tendon • Shortening of the implanted tissue • Possible impingement with bone
	Cat	Tendons of the medial gastrocnemius and soleus muscle	Posture, treadmill locomotion, jumping	Determination of muscular forces during locomotion, correlation with myoelectric activity, comparison with data measured during maximal nerve stimulation	Gregor et al., 1988; Sherif et al., 1983; Walmsley et al., 1978		
	Rabbit	Achilles tendon	?	Development of a force transducer usable in man	Komi et al., 1987		
	Man	Achilles tendon	Walk, running, jumping, sprinting, cycling	Determination of the tendon forces and the moment produced by the triceps surae muscles, comparison with the residual muscle moment at the ankle and the tendon force estimated from the ankle joint moment	Fukashiro et al., 1993;Gregor et al., 1987, 1991; Komi et al., 1987; Komi, 1990		

Table 2 (*continued*)

Transducer family	In vivo use of the transducer			Aims of experiment	References	Advantages	Disadvantages
	Species	Tissue	Gait of locomotion or movement studied				
1.1.2.b Implantable force transducer (IFT)	Horse	SDFT, DDFT, SL	Walk and trot	Determination of the tendon loads during the locomotion	Platt et al., 1990; Platt et al., 1994	• Less bulky than buckle transducers • Induction of a very small deformation of the tendon or ligament	• Calibration sensible to the least minimal movements of the transducer within the tissue and to the tissue form (noncircular tissue)
	Goat	Patellar tendon	Posture, walk and trot	Determination of the patellar tendon force	Korvick et al., 1996		
	Sheep	Tendon of lateral digital extensor	Walk and trot	Determination of the tendon strain and correlation with the muscular activity	Kear and Smith, 1975		
	Cat	Patellar tendon	Treadmill locomotion	Determination of the patellar tendon force (to predict patellofemoral contact forces)	Hasler and Herzog, 1997		
	Man	ACL	Knee extension	Determination of the ligament load during knee extension exercises	Roberts et al., 1994		
1.1.2.c Pressure transducer	Goat	ACL	Walk	Determination of the ligament load during the locomotion	Holden et al., 1994	• Small size • Induction of a very small deformation of the tendon or ligament	• Calibration performed at multiple joint positions for each specimen
1.2. Hall effect transducer	Horse	SDFT	Walk, trot, canter and gallop	Determination of the tendon strains during the locomotion, influence of hoof angle changes	Stephens et al., 1989	• Possible arthroscopic implantation	
	Man	ACL	Knee extension and flexion	Determination of the ligament strains during knee movements and quadriceps contractions	Howe et al., 1990		
1.3. Optic fibre	Man	Achilles tendon, patellar tendon	Walking and jumping	Determination of the tendon forces during walk and jumps to study muscle loading	Finni et al., 1998, 2000	• Implantation: quick and virtually painless • Light modifications of geometry and structure of the tendon	• Possible artefact related to the movement of the skin

ACL = anterior cruciate ligament of the knee, SDFT = superficial digital flexor tendon, DDFT = deep digital flexor tendon, SL = suspensory ligament (or 3rd interosseus muscle), DCL = distal accessory ligament, LDET = long digital extensor tendon.

Marlkof et al., 1998a; Roberts et al., 1994). In some studies, the transducers are attached to the inside of the tendon by sutures (LMSG: Jansen et al., 1998; Lochner et al., 1980; Riemersma et al., 1985; Riemersma and Lammertink, 1988; Riemersma et al., 1988a; Van Weeren et al., 1992; implantable force transducer: Marlkof et al., 1998a; Marlkof et al., 1998b) or barbs (Fleming et al., 1996; Howe et al., 1990; Stephens et al., 1989). After the transducer implantation, the tendon incision is generally closed with a simple interrupted suture to prevent the transducer being detached from the tendon during stretching or loading tissue. In rare cases, transducers are not implanted within the tendons but sutured to the surface of tendons (LMSG: Brown et al., 1986; Jansen et al., 1993b; Keegan et al., 1992; Stone et al., 1983; implantable force transducer: Kear and Smith, 1975). On the other hand, the buckle transducer is positioned around the tendon or ligament (Barnes and Pinder, 1974; Fukashiro et al., 1993; Gregor et al., 1988; Herrick et al., 1978; Komi et al., 1987; Komi, 1990; Lewis et al., 1982; Sherif et al., 1983; Walmsley et al., 1978); this implantation is accomplished by interlacing tissue through the buckle.

The different transducers are implanted either parallel to the tendon or ligament fibres (on surface (LMSG: Brown et al., 1986; Jansen et al., 1993b; Keegan et al., 1992; Stone et al., 1983; implantable force transducer: Kear and Smith, 1975), within the midsubstance (LMSG: Jansen et al., 1993a, 1998; Lochner et al., 1980; Riemersma et al., 1985, 1988a,b, 1996a,b; Van Weeren et al., 1992; Hall effect transducer: Fleming et al., 1994; Howe et al., 1990; Stephens et al., 1989; extensometry gauges transducer: Butler et al., 1989; Fleming et al., 1994, 1996; Glos et al., 1993; Hall et al., 1999; Herzog et al., 1996a,b; Korvick et al., 1996; Marlkof et al., 1998a; Platt et al., 1994; Roberts et al., 1994; Rupert et al., 1998; Xu et al., 1992; pressure transducer: Holden et al., 1995), or around the tendons or ligaments (buckle transducer: Barnes and Pinder, 1974; Fukashiro et al., 1993; Gregor et al., 1987, 1988; Komi et al., 1987, 1990; Lewis et al., 1982; Sherif et al., 1983; Silver and Rossdale, 1983; Walmsley et al., 1978) or transversally to the fibres within the tendon or ligament (implantable force transducer: Finni et al., 2000; Fleming et al., 2000; Hall et al., 1999; Marlkof et al., 1998b; optic fibre: Erdemir et al., 2001, 2002; Finni et al., 1998, 2000; Komiet al., 1996).

3.3. Calibration procedure

The conversion of the signal collected (electric tension) into a strain (relative elongation of the tissue) or a load (internal force within the tissue tested) is possible only after calibration of the transducer. Conversion of the transducer signal to tendon strain can be done by stretching of the sensor itself (with a dial-type Vernier calliper) (Stephens et al., 1989) or by stretching of the transducer sutured or implanted in the tested tissue (in conjunction with a extensometer attached to the surface of the tendon which is stretched (Herrick et al., 1978) or with clamping forceps and a calibrated spring scale (Lewis et al., 1982) (Table 1).

Conversion of the transducer signals to tendon loads can be done by direct compressing on the transducer during which the external force and the transducer output is recorded (Platt et al., 1994). One can also use known weights: Sherif et al. (1983) and Walmsley et al. (1978) carried out the calibration of buckle transducers prior to implantation by hanging a series of known weights to a nylon cord passing through the buckle. The conversion can also be done within the tendon, submitting the tendons equipped with the transducer to tensile force tests in a material testing machine equipped with a force transducer, 1. by traction on the isolated tendon (Herrick et al., 1978; Jansen et al., 1993b; Jansen et al., 1998; Kear and Smith, 1975; Lochner et al., 1980; Platt et al., 1994; Riemersma et al., 1985; Riemersma and Lammertink, 1988; Riemersma et al., 1988a,b; 1996b), 2. by attaching known weights to the isolated tendon (Barnes and Pinder, 1974), 3. by stretching of the in situ tendons within the limb by compression of the isolated limb (Riemersma and Lammertink, 1988; Riemersma et al., 1996a), 4. by flexion-extension, translation or rotation movements exerted on the knee (Hasler and Herzog, 1997; Holden et al., 1994; Stone et al., 1983) or 5. by electrical stimulating of the muscle acting on the tendon under study (Herzog et al., 1996b; Kear and Smith, 1975; Komi, 1990) (Table 1). The isolated tendons (or ligaments) are submitted to traction test during which the force applied to the tendon, the transducer signal and the temperature (the temperature only in the case of the LMSG) are simultaneously recorded. In an equine study (Riemersma et al., 1996a), the isolated limbs are often submitted to a vertical loading simulating a mid-stance phase position at the walk. The transducer signals (electrical resistance) are thus converted into tendon strains or loads using the relationship between transducer signals and direct tendon loads obtained from these tensile force tests.

In animals the calibration of the in situ transducer in the tendon or ligament can be made directly by a post-mortem procedure on dissected anatomical specimens (isolated tendons, knees or legs for example). For that, after the in vivo measurements, the animal is generally euthanasied, the limbs are isolated and tendons—still containing the transducers—are dissected apart from the limb (Herzog et al., 1996b; Holden et al., 1994; Jansen et al., 1993a, 1998; Riemersma et al., 1985, 1988a).

However, in human experimentation, a direct calibration (post-mortem) cannot be used; it is thus necessary to perform an indirect method using the

mathematical expression of the static equilibrium. For example, with a calibration table, the force developed within the Achilles tendon during dorsiflexing of the foot with known calibration weights is calculated taking into consideration the geometrical arrangements of the transducer, axis of rotation and the pulley system (Fukashiro et al., 1993; Gregor et al., 1991; Komi et al., 1987, 1990). In a study of forces developed within the anterior cruciate ligament (ACL) during knee extension exercises (Roberts et al., 1994), the transducer (a force transducer) was calibrated in vivo relatively to anterior–posterior shear forces at different flexion angles of the knee. The anterior–posterior shear force was applied to the proximal tibia with an instrumented force handle and the transducer calibration to ACL force was obtained using an equation determined in vitro that predicted the ACL force induced by the anterior–posterior shear force (Table 1).

Moreover, it is necessary to calibrate in strain or load the transducer for each type of tendon or ligament in which it is used (Holden et al., 1995). For a type of anatomical structure (tendon or ligament), it is necessary to repeat the procedure of calibration to several tendons (five superficial digital flexor tendons, for example) (Platt et al., 1994) to take into account the inter-specimen variability in the calibration device within the same anatomical structure. The in vitro calibration should be therefore performed in the site and in the anatomical structure that will be instrumented in vivo.

4. Limitations

4.1. Problems induced by the implantation of transducers in tendons or ligaments

All of the above invasive transducers require surgery under local (in man) or general (in animals) anaesthesia for their implantation within or around the tendon or ligament. The patients must thus recover from surgery and general anaesthesia before being able to carry out the in vivo experiments. However, the delay in recovery is short: the in vivo measurements are often made the same day of surgery (Fukashiro et al., 1993; Howe et al., 1990; Komi et al., 1987, 1990; Riemersma et al., 1985, 1996b; Stephens et al., 1989), within 1 to 4 days (Barnes and Pinder, 1974; Holden et al., 1994; Jansen et al., 1993a; Keegan et al., 1992; Lewis et al., 1982; Platt et al., 1994; Van Weeren et al., 1992), up to 10 days post-implantation (Kear and Smith, 1975; Stephens et al., 1989) or even later (Walmsley et al., 1978). All the transducers except the buckle transducers seem well tolerated by the human and animal patients. With the buckle transducer, after the implantation, a significant level of pain may be provoked, especially if the dimensions of the transducer are too large. In a human study (Komi et al., 1987), the patient could only walk slowly during the first post-op week because of a persistent pain. This problem disappeared after size reduction of the buckle transducer. Cats equipped with a buckle transducer walked again normally about one week after implantation (Walmsley et al., 1978). Moreover in man, the complete healing of the wound after the experiments, takes 2–3 weeks before the subject can resume normal walking or running (Komi et al., 1996).

Many transducers implanted in tendons are positioned within a slit made in the tissue. The slit created for implantation of the transducer seems to have no appreciable effect on the tendon load–strain response and it heals rapidly (Platt et al., 1994).

The installation of a buckle transducer causes inevitably some ligament or tendon curve and shortening to occur, potentially altering the distribution of loads. The size of the buckle transducers must also be selected carefully because an excessive bending could cause too much unnecessary stretch in the wrong directions and it may even damage the tendon at higher stretch loads (Komi, 1990). During movement, these transducers can also impinge on bone or be compressed by surrounding soft tissues, producing a false output. Another problem with some buckle transducers is that they may preload the tissue when they are applied, thereby changing the way the tissue functions in vitro or in vivo. These problems are encountered in particular with short ligaments like the cruciate ligaments of the knee (Lewis et al., 1982; Xu et al., 1992) or with tendons located in a restrained space (in the knee, for example). Therefore, they should be applied principally to the long tendons (Achilles tendon for example) because of the transducer size and the space available around the tendon. The implantation and use on the knee ligaments (cruciate ligaments) can be more difficult (Lewis et al., 1982). However, tissue shortening and impingement with bone can be reduced using buckle transducers adapted (not too large) to the tendons under study (Barry and Ahmed, 1986; Lewis et al., 1982). Barry and Ahmed (1986) succeeded using a modified buckle transducer to measure the loads simultaneously in the anteromedial and posterior fibres of the ACL, as well as the loads developed within the medial collateral ligament of the knee or the loads in the fibres connecting the ACL with the menisci.

4.2. Significance and reliability of the parameters measured by the transducers

The transducer technique is a direct and, therefore, quite a reliable technique for tendon strain or load measurements but it has the disadvantage of measuring strain or load locally. Indeed, the majority of the transducers (except the buckle transducer) used in biomechanics studies are not as wide as the tendon or

ligament in which they are implanted (Herzog et al., 1996b; Holden et al., 1995; Marlkof et al., 1998b; Rupert et al., 1998). This local strain or load is not necessarily representative for the total strain or load of the structure. Therefore, according to the position of the transducer within the tendon (implanted more or less deeply within the tendon and more or less near to the edge sides), one can obtain differences in the transducer signal (Finni et al., 1998; Fleming et al., 2000; Marlkof et al., 1998b; Platt et al., 1994; Rupert et al., 1998). Indeed, Marlkof et al. (1998b) showed that the depth of insertion of the transducer has a marked effect on voltage output of the transducer (IFT) because all tendon fibres do not contribute in the same way to the resultant force. This problem can be minimised by using a wider transducer so that the device is in contact with a greater number of tissue fibres (Rupert et al., 1998). On the other hand, as the buckle transducer surrounds the tendon or ligament under study, one can consider that this type of transducer measures an average force implying all tissue fibres in the implantation site of transducer. If the tendon under study is more or less homogeneous throughout its length and width, as long tendons found in the distal limb of the horse, the strain or load measured locally in the tissue can be considered reflecting the total load of the tendon.

The output signal can also be modified by a slight deviation of the axis of the transducer compared with the axis of the tendon fibres. This problem has been shown by Herzog et al. (1996b) with an IFT used in vitro in the patellar ligament of cat. With this type of transducer, a misalignment from 5° to 10° of the position of the transducer can cause significant changes in the output signal for a given externally applied load. This problem is also typical of the LMSG; an artefact strain appears when the distension (e.g. transversal displacement of one end of the gauge while the other end is held fixed) angle reaches about 15° (Brown et al., 1986). The ends of the transducer must thus be sutured to the tissue after having taken care to control its positioning compared to the orientation of the tissue fibres (parallel positioning) (Herzog et al., 1996b).

It is moreover necessary to check that the transducer position and alignment are the same during the in vivo recordings and the calibration procedure. Herzog et al. (1996b) suggested that the transducer be implanted two weeks before the realisation of the in vivo experiments so that the local inflammatory reaction encourages the device to adhere to the tissue, thus minimising any displacement of one relative to the other. However, in some cases (Korvick et al., 1996), problems of wire break appear at the time of the in vivo recordings when there is a time between the transducer implantation and the recording of the signal output.

4.3. Calibration problems

The data obtained finally (after calibration) from every type of transducer must be interpreted with caution. Several parameters can influence the relation obtained between the transducer signal and the tendon load during the calibration test: the tendon temperature, the course of the tendon different in the calibration test compared with the experiment,... (Lochner et al., 1980; Riemersma et al., 1985; Riemersma and Lammertink, 1988).

The signal amplitude of the LMSG is temperature dependent because the resistance of the gauge increases with temperature rise. In experiments performed to calibrate the LMSG, an increase in temperature of 1 °C produced a mean increase in the signal amplitude corresponding to a strain of 0.051% (Riemersma and Lammertink, 1988) to 0.185% (Brown et al., 1986). The influence of temperature on the signal appears as a light shift of the gauge signal. However, significant differences in temperature (15–20 °C) may be expected between in in vitro and in vivo circumstances which require exact determination of these changes. Ignoring the influence of temperature difference may induce an artefact of approximately 1% of strain, which represents a third of the maximum (SDFT) tendon strain estimated in horse during the walk (Riemersma et al., 1985). On the other hand, the other transducers (extensometry gauges transducer, Hall effect transducer and optic fibre) do not seem to be sensitive to temperature (Holden et al., 1995).

As the transducer output can be modified by removing and reinserting the device, the calibration must be performed without removing the transducer from the tendon or ligament between the experiment and the calibration procedure (Fleming et al., 2000).

Also, since the tendon cannot be fixed in vitro in exactly the same way as it is in vivo, the fibres to which the transducer is attached can be strained differently and the load distribution within a tendon can differ between the in vivo experiment and the in vitro calibration. The fibres to which the transducer is attached may be strained differently in vitro than they are in situ under the same tendon load. The mutual position of the tendon or ligament fibres may change because of the lack of deviation around joints and a change in the positions of origin and insertion. In an experiment carried out with LMSG by Riemersma and Lammertink (1988), a decrease of the signal amplitude was observed when the tendon was isolated and clamped for stretching compared to the tests on intact limb. In order to avoid these problems, the authors (Riemersma and Lammertink, 1988) have developed a calibration procedure in which the amplitude shifts of the LMSG due to tendon isolation and temperature differences were determined.

The alternative of clamping is the use of bone–tendon or ligament–bone preparation. Several studies have used

this technique. However, fixing a tendon in this way will not ensure an exact reproduction of in vivo load distribution within the tendon or ligament fibres. This depends on the complexity of origin or insertion of the structure and the degree of deviation around joints in vivo.

Moreover, in some studies (Glos et al., 1990; Hall et al., 1999; Herzog et al., 1996a; Holden et al., 1994; Korvick et al., 1996), when dissected anatomical specimens are used for the calibration procedure, some tissues (others tendons or ligaments than the tissue under study, muscle and soft tissues, subcutaneous tissue, skin) of the anatomical specimen (knee for example) are removed from around the tendon or ligament under study in order to ensure that the external force applied is transmitted entirely through the latter. The tissue under study (tendon or ligament) still containing the transducer in situ is preserved intact. However, changes in joint position can exist and it can affect force distribution in the tendon or ligament. The in vitro relationship between the internal force within the tendon or ligament and the external force applied to the specimen during the calibration procedure is not comparable with the in vivo relationship when all tissues are present in the anatomic specimen (Glos et al., 1993). It appears therefore necessary to verify the measurements of strain or force obtained from the transducers by independent methods.

4.4. Particular risks during the use of LMSG

The life of an LMSG is theoretically limited (about 6 months) (Brown et al., 1986) because the silastic tubing around the liquid metal column is porous and its integrity can be affected by passage of liquid and dissolved gas. A slow oxidation of the mercury due to oxygen diffusion through the silastic tubing can take place. In in vivo experiments, small differences in the signal are observed between measurements made 48 h after surgical implantation of the mercury-in-silastic strain gauge in the tendons and those carried out 2 weeks after the surgery (Keegan et al., 1992). However, this integrity loss of the liquid metal strain gauge does not seem to be a major problem in experiments (Jansen et al., 1998; Keegan et al., 1992).

The service life of these transducers is also short because of a possible break of connection wires or a contact loss between the connection wire and the liquid metal column following movements of tendon or ligament against the surrounding tissues. There are also local changes of tendon mechanical properties (changes of stress–strain relationships) because of an acute inflammatory reaction (fibrin, serosity and scar tissue) in the transducer implantation (Brown et al., 1986; Jansen et al., 1998; Keegan et al., 1992). In order that these changes do not affect the transducer signal, only a limited number of in vivo measurements can be collected and the measurements should thus be performed soon (3–4 days) after transducer implantation (Jansen et al., 1998).

With the LMSG, there is also a risk of release of toxic substance (mercury or other liquid metal) into the body following a rupture of the silastic tubing containing the liquid metal column (Glos et al., 1993; Riemersma et al., 1985). This risk could exist in vivo for example in horse, even when moving at a slow gait (e.g. walk) (Riemersma et al., 1985).

5. Conclusion and clinical relevance

The internal loading exerted on a tendon or ligament is of importance in various contexts. This applies to muscle physiology or musculoskeletal biomechanics regarding the contribution of a muscle and its tendon during a specific movement, to surgery regarding the reproduction of stretching patterns in prostheses and in human or veterinary medicine regarding the effects of different appliances during the locomotion.

In terms of measurement, the ideal transducer should be reliable, induce a minimal distortion or length change in the tissue, not cause a significant inflammatory reaction and be technically simple to implant. The transducers implanted in tendons or ligaments enable avoidance of impingement artefacts with the surrounding tissues, contrary to the buckle transducers in which tendons or ligaments are placed. However, the attachment of the transducer to the tendon or ligament is problematic because the position influences the signal which in turn reflects the behaviour of a small part of the tendon only. Moreover, the implantation of all transducers within tendons or ligaments causes a varying degree of tissue damage which may affect the transducer signal output and limit the number of possible measurements.

The use of the different types of transducer requires surgery under local (in man) or general (in animals) anaesthesia. Surgical implantation has the disadvantage of inducing a local reaction of the tissue which can disturb the measurements and increase the time for recovery before the experimental measurements can be conducted and recorded. Thus a non-invasive transducer model which could be used in vivo by setting the transducer directly on the skin racing the tendon or ligament under study is definitely a challenge for the future.

Acknowledgements

The authors wish to thank M. Sylvain Falala for his contribution to the illustrations of this article.

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