Noninvasive Measurement of Torque Development in the Rat Foot: Measurement Setup and Results From Stimulation of the Sciatic Nerve With Polyimide-Based Cuff Electrodes

Thomas Stieglitz, *Member, IEEE*, Martin Schuettler, *Member, IEEE*, Andreas Schneider, Elena Valderrama, and Xavier Navarro

***Abstract*—In neural rehabilitation, selective activation of muscles after electrical stimulation is mandatory for control of paralyzed limbs. For an evaluation of electrode selectivity, a setup to noninvasively measure the force development after electrical stimulation in the rat foot was developed. The setup was designed in accordance to the anatomical features of the rat model to test the isometric torque development at given ankle positions in an intact leg. In this paper, the setup design and development is presented and discussed. In a first study, the selectivity of small nerve cuffs with 12 electrodes implanted around the rat sciatic nerve was investigated. Special attention was drawn to the performance of the torque measurement setup in comparison to electrophysiological data obtained from compound muscle action potential recordings. Using one cuff around the nerve, electrical stimulation on different electrode tripoles led to plantarflexion and dorsiflexion of the foot without an *a priori* alignment of the cuff.**

***Index Terms*—Cuff-electrode, electrical stimulation, peripheral nerve, polyimide, torque, force.**

I. Introduction

CUFF ELECTRODES became widespread for interfacing peripheral nerves in neural prostheses during the last three decades. The development of multichannel stimulation sites inside the cuffs allowed partial excitation of different nerve fascicles with selective control of different muscles [1]. The force development on the animal's foot was monitored with a noninvasive measurement setup [11]. It allowed repeated measurements on the cat foot, the commonly used animal model in the studies of Mortimer's group (CWRU, Cleveland, OH). The knee of the animal was fixed with a clamp and the foot was inserted into a "shoe" that was connected to a measurement device (JR3, Woodland, CA) with sensors for three force axes ($F_{\mathrm{x,y,z}} < 100$ N) and three torques ($M_{\mathrm{x,y,z}} < 8$ N·m). Measurement of all directions of movement is necessary in the cat [18]. The maximal measured torque was in the range of 3 N·m in the plantar-/dorsiflexion plane and selective recruitment of different muscles for movement was reported using twelve-polar cuffs [12].

For use in smaller animal models, the mass of the JR3 transducer and measurement setup might be too large (and too expensive) for widespread use. A setup with a custom made single axis torque transducer was established at the University of Michigan, Ann Arbor, for the mouse model [2]. Force development was measured in the plantar-/dorsiflexion plane with a maximum value of 25 N·mm. The development of force and torque of the commonly used rat model should be somewhere in between the mouse and the cat. Therefore, a measurement setup for the rat was developed with abilities to monitor the plantar-/dorsiflexion and the medial/lateral rotation plane. Investigations on the rat sciatic nerve, that is much smaller than the cat one, should evaluate if the possibility to selectively control antagonist muscles exists with the recently developed polyimide-based small nerve cuff electrodes.

In this paper, we present our development of a simple and noninvasive torque measurement device for the rat animal model and the first results from stimulation experiments investigating the selectivity of newly developed polyimide-based cuff electrodes implanted around small nerves.

II. Materials and Methods

A device for noninvasive torque measurement of the foot of the rat was developed in accordance to its anatomical features. A theoretical consideration of the main muscles and their insertion points led to a design of the measurement setup for the two main directions of movement and torque development.

A. Anatomical Considerations of the Rat Foot With Regard to a Torque Measurement Setup

In a short theoretical summary, the main aspects of the rat animal model with regard to an access to the sciatic nerve were described. The sciatic nerve was established as a common model for peripheral nerve stimulation with multipolar electrodes, mainly cuff electrodes. The nerve is easily accessible at a side above the knee joint. It innervates a large number of muscles in the thigh, the shank, and the foot of the lower limb. Therefore, selective stimulation of distinct portions should be

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TABLE I
INNERVATION OF MUSCLES BY THE SCIATIC NERVE AND THEIR FUNCTION. DF—DORSIFLEXION. PF—PLANTARFLEXION. Su—SUPINATION. Pr—PRONATION. MR—MEDIAL ROTATION. LR—LATERAL ROTATION. Ex—EXTENSION. Fl—FLEXION. Ab—ABDUCTION. Ad—ADDUCTION. THE MUSCLES THAT ARE MARKED IN GRAY DO NOT DIRECTLY INDUCE A MOVEMENT AT THE ANKLE JOINT, BUT MAY AFFECT THE TORQUE MEASUREMENTS

Name **Musculus ...**	**Innervation** **Nervus ...**	**Function**	
Tibialis posterior	Tibialis	ankle joint:	PF/Su
Gastrocnemius	Tibialis	ankle joint:	PF
Soleus	Tibialis	ankle joint:	PF
Plantaris	Tibialis	ankle joint: toe joint:	PF Fl
Flexor digitorum longus	Tibialis	ankle joint: toe joint:	PF Fl
Flexor hallucis longus	Tibialis	ankle joint: big toe joint:	PF Fl
Tibialis anterior	Peroneus profundus	ankle joint:	DF/Su
Extensor digitorum longus	Peroneus profundus	ankle joint: toe joint:	DF Ex
Extensor hallucis longus	Peroneus profundus	ankle joint: big toe joint:	DF Ex
Peroneus longus	Peroneus superficialis	ankle joint:	DF/Pr
Peroneus brevis	Peroneus superficialis	ankle joint:	DF
Quadratus plantae	Plantaris lateralis	ankle joint:	DF
Semitendinosus	Cutaneus surae caudalis	hip joint: knee joint:	Ex Fl
Popliteus	Tibialis	knee joint:	MR
Extensor digitorum brevis	Peroneus profundus	toe joints:	Ex
Peroneus digiti quarti	Peroneus profundus	toe joint IV:	Ex
Peroneus digiti quinti	Peroneus profundus	toe joint V:	Ex
Flexor digitorum brevis	Plantaris medialis	toe jointe:	Fl
Flexor hallucis brevis	Plantaris medialis	big toe joint:	Fl
Adductor indicis	Plantaris medialis	toe joints:	Ad
Flexor digiti quinti brevis	Plantaris lateralis	toe joint V:	Fl
Abductor digiti quinti	Plantaris lateralis	toe joint V:	Ab
Interossei plantares	Plantaris lateralis	toe joints:	Fl/Ab/Ad
Lumbricales I-IV	Plantaris medialis/lateralis	toe joints I-IV:	Fl

obtained to control as many muscles as possible with as little implanted hardware as necessary. The anatomy of the sciatic nerve at the implantation site above the trifurcation is well known [15]. It consists of fascicles splitting more distally to the Nervus cutaneus surae caudalis, the N. cutaneus surae lateralis, the N. tibialis, and the N. peroneus [27]. The sciatic nerve and its branches innervate many muscles of the lower limb that have impact on the torque development of the ankle joint (Table I) [4], [9], [15], [16], as well as torques and rotations in the knee joint and the toe joints. The directions with the largest force and torque development is in the dorsiflexion/plantarplexion plane. The orthogonal directions in the supination/pronation and medial/lateral rotation plane are much smaller. The torques in the supination/pronation plane are so small that they were neglected in a cat model with even higher torque values [11], [12]. From the biomechanical point of view, the ankle joint allows three axis of rotation that depend from the direction of rotation and the actual joint position. In humans, the axis of rotation crosses in a central rotation point [20]. In cats, the ankle joint has a central point in the middle of the talus [18] and the ankle joint of rats may also be assumed as a joint with a fixed rotation point [1]. The ligaments are assumed as unelastic [1] and their torques were neglected because of the small clearance in the joints [28]. For an isometric contraction of muscles that will be investigated here, the inertia forces need not to be taken into account [33]. The force lines of the different muscles are difficult to obtain but they can be approximated by force lines that connect origin and insertion of each muscle. The main pronator is the M. peroneus longus, the main plantarflexor is the M. gastrocncemius. Medial rotation is induced mainly by the M. popliteus, dorsiflexion by the M. tibialis anterior. Due to the sophisticated synergisms of agonists and antagonists and the throwing of some tendons before their insertion points, a measurement of generated torques in an unsevered lower limb with all agonisms and antagonisms after electrical nerve stimulation should deliver a more realistic view than the measurement of isolated muscle-tendon preparations. In some investigations, *in vivo* measurements were compared

TABLE II
MUSCLE FOR PLANTAR-/DORSIFLEXION OF FOOT IN THE RAT *IN VIVO* AND CONSIDERATION AT *IN SITU* MEASUREMENTS

Investigation	Dorsiflexors	Plantarflexors
torque *in-vivo*	M. tibialis anterior M. peroneus longus M. peroneus brevis M. quadratus plantae M. extensor digitorum longus M. extensor hallucis longus	M. tibialis posterior M. gastrocnemius M. soleus M. plantaris M. flexor digitorum longus M. flexor hallucis longus
force *in-situ*	M. tibialis anterior M. extensor digitorum longus	M. gastrocnemius M. soleus M. plantaris

TABLE III
TORQUE TRANSDUCER AND SETUP SPECIFICATIONS

Specification	Unit	Name/value
torque transducer	-	**Wazau Type DMA/VM-2000**
measurement range	Nmm	0-2000
supply voltage	V	5
sensitivity (full wheatstone bridge, R=350 Ω)	mV/V	1.474
maximum transducer voltage	mV	7.37
bellows couplings	-	**Baeuerle Type BSK 39 08 40**
distortion of coupling at nominal torque (2000 Nmm)	°	1.52
signal processing		**custom made at IBMT**
first stage	-	INA 131, Burr Brown
gain	-	100
supply voltage	V	± 12
second stage (optional)	-	OPA 404, Burr Brown
amplification of second stage (optinal)	-	100

with results from single muscle preparations *in situ* (Table II) [11], [12] to validate the *in vivo* data. With the anatomical and biomechanical considerations in mind, we decided to design and develop a torque measurement setup for the rat foot with transducers for the plantar-/dorsiflexion and the lateral/medial rotation plane.

B. Torque Measurement Setup

A setup was designed for the noninvasive measurement of torque development of the rat foot during electrical stimulation. The rat has to be fixed laying on its side with an electrode implanted in the upper laying leg. The following are the basic components of the torque measurement setup (TableIII):

1) sensors for torque measurement around the ankle joint;
2) flexible couplings for decoupling of measurement axis;
3) shoe that is stiff against torsion with elements for adjustment of axis of rotation and fixation of rat foot;
4) connecting elements between the shoe, decoupling unit, and toque transducer(s);
5) mount for fixation of all system components;
6) clamp for fixation of the knee joint of the rat;
7) amplifier to adapt output signal range to conventional analog or digital recording systems.

The torques that were expected during the experiments were originated by isometric contractions of the muscles, i.e., quasi-static but mainly static torque courses have to be measured. For static acuity, we omitted piezoelectric-based transducers and focused on strain gauge-based transducers. Only few companies offered devices with measurement ranges up to 1 N·m, a value that has been extrapolated from other animal models to the rat model [12], [1]. Finally, we chose a strain gauge based torque sensor (DMA/VM-20000, Watzau, Berlin, Germany) with a measurement range up to 2000 N·mm. These sensors have small dimensions with a diameter of 23 mm, a height of 10 mm, and a resolution of 1 N·mm. Transducers with the same dimensions and lower measurement range [1 N·m with 2.0-N·mm resolution (0.1% of maximum value), 0.5 N·m with 0.5-N·mm resolution] were also available and might be used at smaller maximum torque values to increase the possible resolution.

Couplings were chosen that fit the mechanical and geometrical requirements of the setup. The torque transduction range

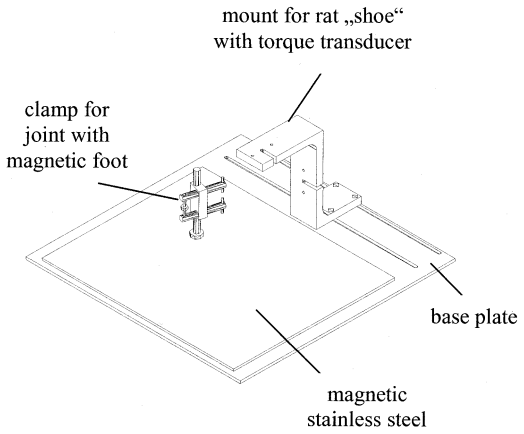

Fig. 1. Design of the base plate with mount for the torque measurement device. An adjustable clamp for mechanical fixation of the knee can be placed all over the base plate; an integrated magnet ensures a stable position.

of the coupling must be above the expected value of the torque at the ankle joint. The stiffness of the coupling should be tenfold above the stiffness of the sensor under a dynamic load. For a static load, a smaller stiffness was expected to be sufficient. Two flexible bellows couplings (BSK 39 08 40, Baeuerle, St. Georgen, Germany) were chosen to separate the torques of plantar-/dorsiflexion from those of supination/pronation and medial/lateral rotation. These couplings were fabricated with means of precision mechanics and got small dimensions (25-mm diameter, 33-mm length) at a maximum torque transmission of 800 N·mm and a high torsion stiffness of 300 N·m/rad. They allow a maximum angular displacement α of 5°. The displacement was calculated according to

$$\alpha_{\text{ges}} = \alpha_S + \alpha_K = \frac{M_{\max}}{T_S} + \frac{M_{\max}}{T_K} \tag{1}$$

with a maximum torque $M_{\max} = 1$ N·m, the torsion stiffness $T_S = 100$ N·m/rad of the sensor and the torsion stiffness $T_K = 300$ N·m/rad of the coupling and

$$\alpha\,[^\circ] = \alpha\,[\text{rad}] \cdot 180/\pi.$$

It resulted in a total maximum displacement of

$$\alpha_{\text{ges}} = \left(\frac{1}{100} + \frac{1}{300}\right)\left[\frac{\text{Nm}}{\text{N}\cdot\text{m/rad}}\right] \cdot \frac{180\,[^\circ]}{\pi\,[\text{rad}]} = 0,76^\circ \tag{2}$$

a value much lower than the maximum allowed displacement.

A base plate with guides for a mount was built onto which the whole torque measurement device was placed (Fig. 1). A further plate made of magnetic stainless steel allowed the fixation of an adjustable clamp via a magnet for the rat's knee.

The shoe was made out of a transparent material to allow adjustment of the foot under visual inspection in respect to the ankle joint rotation axis. Polymethylmetacrylate (PMMA) was chosen. The size of the shoe (inner dimensions: 20-mm width; 55-mm length; 15-mm height) was adapted to the anatomy of rats. The material is less stiff than stainless steel but stiff enough for the expected torque values below 1 N·m. Three adjustment screws for horizontal alignment were fit into the shoe. Height adjustment could be performed via PMMA washers. A final fixation of the foot could be done with a fixation screw in the top

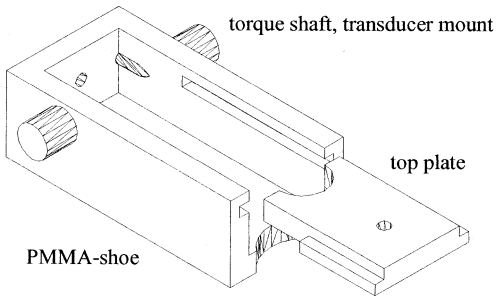

Fig. 2. "Shoe" with top plate to fix rat foot. The shaft allows fixation to torque transducers on the mount. Holes with threads are for screws for fine adjustment of the rat foot.

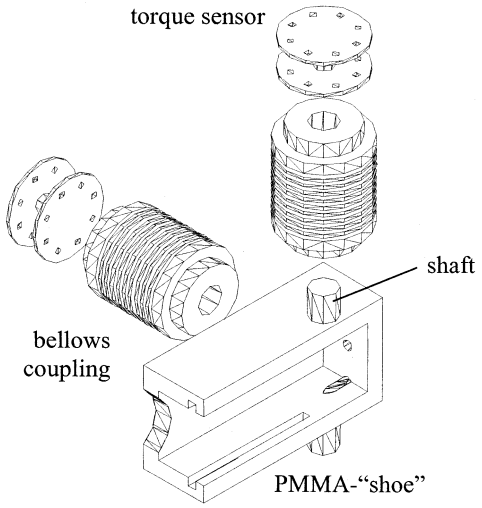

Fig. 3. "Shoe" with bellows couplings and torque transducers to insert the rat foot for measurements.

plate of the shoe (Fig. 2). Two orthogonal shafts were integrated into the shoe to allow assembling with the bellows couplings and the torque sensor(s) (Fig. 3). This setup containing shoe, couplings, and sensors was fit into the mount on the base plate and was finally fixed with only four screws.

The amplification of the sensor output signal was performed in a two-stage amplifier with a gain of 100 in the first stage (INA 131 instrumentation amplifier, Burr Brown) and a gain of 1 or 100, respectively, in the second stage (OPA 404 operational amplifier, Burr Brown). Thereby, the sensor output signal was set to a sensitivity of 0.4 and 40 V/N·m, respectively, for adaptation to standard data acquisition systems.

C. Polyimide-Based Nerve Cuff Electrodes

Cuff electrodes with 12 dot electrodes that could be arranged to four tripoles were designed (Fig. 4, Table IV). They were fabricated using micromachining technologies established to fabricate flexible micromachined electrodes with integrated interconnection lines and contact pads [30]. Polyimide (Pyralin PI 2611/HD Microsystems, Bad Homburg, Germany) was chosen as substrate and insulation material for cuff electrodes. The resin like polyimide was spin-coated on silicon wafers. Imidization was performed at 350 °C. Then, thin-film metallization of platinum was sputter deposited and patterned for electrode areas, interconnection lines, and connection pads. A top layer of poly-

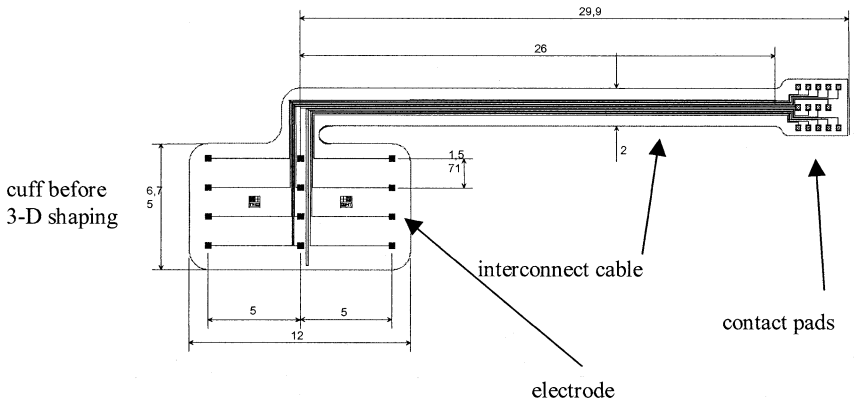

Fig. 4. Design of a 12-channel cuff electrode. All dimensions are in millimeters.

TABLE IV
CUFF ELECTRODE SPECIFICATIONS

Specification	Dimension
thickness of polyimide device	10 μm
cuff diameters	0.7 / 1.0 - 2 mm in steps of 0.2 mm
number of electrodes	12 dots, arranged in 4 tripoles
longitudinal electrode pitch	5 mm
interconnect cable width	2 mm
interconnect length	26.4 mm
cuff length	12 mm
dot electrode area	70685 μm² (diameter 300 μm)

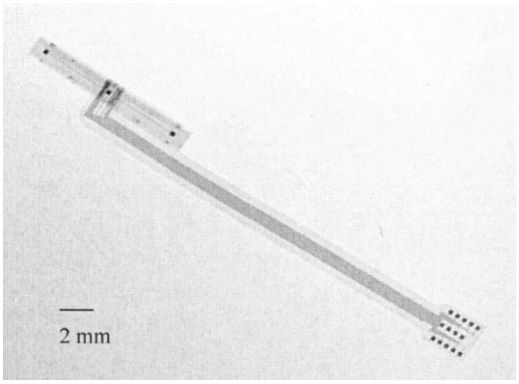

Fig. 5. Polyimide-based cuff-electrode with 12 stimulation sites. Cuff Diameter: 1.6 mm.

imide was spin-coated and imidized for insulation and mechanical support. Reactive ion etching (RIE) was applied for opening the electrode and contact areas. RIE was also used to define the outer shapes of the devices. The devices were separated from the silicon wafer and were rolled to cuffs. In a temper step at 340 °C for 2 h, the mechanical stresses in the devices were mainly released. The cuffs stayed stable in the rolled position (Fig. 5). For stimulation experiments during acute implantations, the devices were assembled with a miniaturized plug. The electrodes could be addressed individually via this plug. An arrangement to tripoles was made by connecting the outer electrodes to one stimulation channel as anode and using the electrode in between as cathode.

D. Acute Implantation

Polyimide-based cuff electrodes with 12 stimulation sites arranged to four tripoles were used with a diameter of 1.6 mm for acute implantation. Surgical intervention was performed in female Sprague–Dawley rats under pentobarbital anaesthesia (40 mg/kg i.p.) with the aid of a dissecting microscope. The left sciatic nerve was exposed at the midthigh and carefully freed from surrounding tissues from the sciatic notch to the knee. Its diameter was around 1.2 mm. The cuff was opened and placed around the sciatic nerve avoiding compression and stretch without any prealignment to the nerve fascicles. The electrode interconnect ribbon was routed through the muscle plane excision, avoiding tension, placing the ending enlargement on the lateral side of the hind limb. The animal was placed on the stainless steel plate of the torque measurement setup. The left foot was inserted into the “shoe” and gently fixed with adjusting screws to prevent any relative movement inside the “shoe.” The knee was fixed with the adjustable clamp that was placed onto the base plate with the magnet (Fig. 6). The animal was fixed with adhesive tape on the base plate to prevent undesired movements due to electrical stimulation. The whole setup was placed over a warm flat streamer controlled by a hot-water circulating pump, and the hind paw skin temperature was monitored and maintained above 32 °C.

E. Stimulation Experiments

In acute stimulation experiments, monophasic rectangular current pulses (GRASS S88 with voltage to current converter) with a pulse width of 10 μs were applied. Compound muscle action potentials (CMAPs) were recorded simultaneously from the tibialis anterior and gastrocnemius medialis muscles with small needle electrodes inserted in each muscle. Additionally, the torque development in the foot was measured with the newly developed device. The evoked potentials and the torque were displayed on a storage oscilloscope (Tektronix TD421) at settings appropriate to measure the amplitude from baseline to peak and the latency to the onset. Each of the four tripoles of the cuff with the middle electrode as cathode and outer electrodes short-circuited to be anodes were tested from threshold up to supramaximal excitation. Tripoles that excited antagonistic

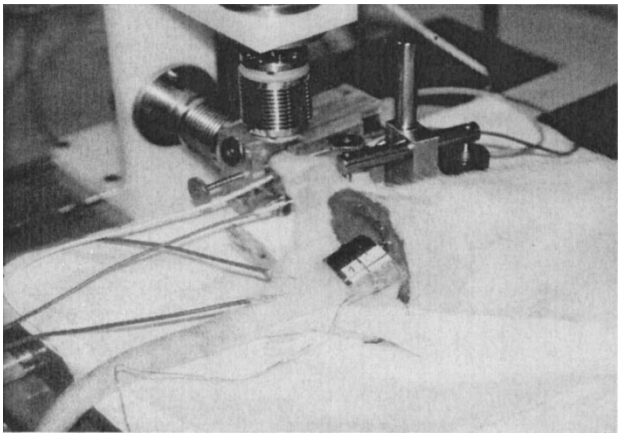

Fig. 6. Experimental setup. Polyimide-based cuff implanted around the sciatic nerve of the rat. Needle electrodes inserted to record CMAPs and rat foot in torque measurement setup.

muscles were chosen for subsequent, repeated stimulation to investigate the reproducibility of the stimulation effects with regard to the torque development.

III. Results

First investigations were performed on eight rats using the electrodes and the measurement setup described above. The data presented here are from single measurements being representative for all measurements. Detailed results with regard to the selectivity during stimulation can be found elsewhere [24].

A. Measurement Setup and Calibration of Torque Transducers

The torque transducers with the amplifier circuitry were calibrated outside the setup using a cantilever at which probes with a known mass were fixed. The fixations were 99.1 and 100.4 mm apart from the axis of the transducer (Fig. 7). Using test masses with $m = 970$ g, the resulting torques were $M_1 = 943$ and $M_2 = -955$ N·mm. The inclination of the graph comparing the calculated and the measured value of the torque was calculated via the maximal voltage values at the torque transducer outputs

$$m_M = \frac{\Delta M}{\Delta U} = \frac{M_2 - M_1}{U_2 - U_1}.$$

The calibration constant was calculated to $m_M = 400$ N·mm/V with the measured voltages of $U_1 = 2.279$ V and $U_2 = -2.469$ V. A calibration curve was determined applying different forces on the cantilever (Fig. 8). The mean relative deviation between the measured value and the nominal value that is a measure for the systematic error was 0.3%. The mean square deviation that is a measure for the variance of the measurements was 2.7%.

The transducers and the bellows couplings were assembled with the shoe in the mount onto the base plate (Fig. 9). The bellows couplings and the shoe were easy to assemble, exchange of the torque transducer was fast and easy to do. The clamp for the knee fixation could be moved with the magnet that adhered very strong to the magnetic base plate.

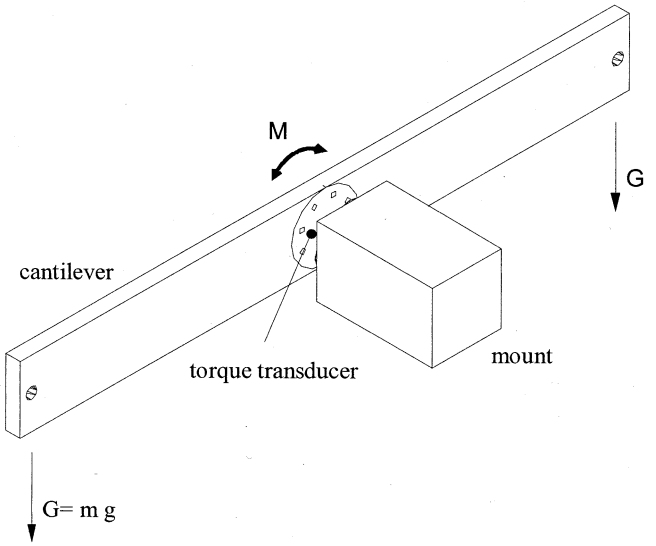

Fig. 7. Setup for the calibration of the torque transducer with test masses. The test masses exert a force on the cantilever due to gravity. The resulting torque was measured with the transducer.

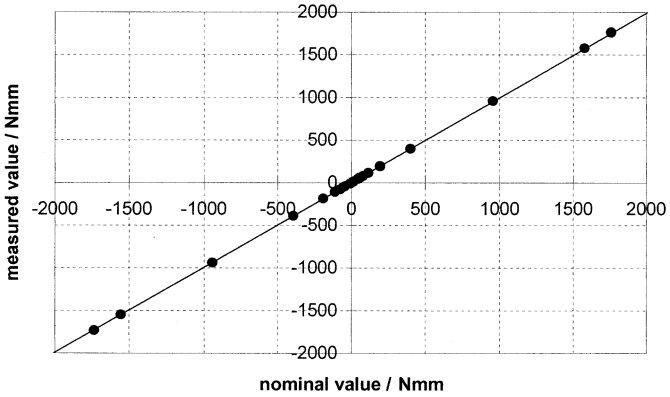

Fig. 8. Calibration graph of the torque transducer. The dots represent measurement results, the line represents the calculated values.

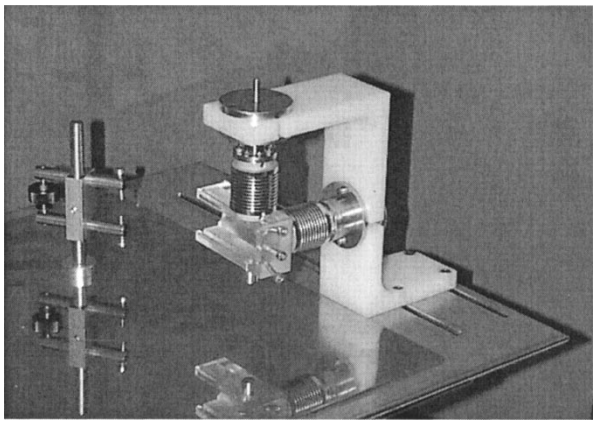

Fig. 9. Photograph of the assembled torque measurement setup. Adjustable clamp (foreground, left). Mount (background) with "shoe," bellows couplings and torque transducer.

B. Nerve-Cuff Electrodes

Nerve-cuff electrodes with 12 stimulation sites were implanted around the sciatic nerve at the midthigh without particular alignment of the stimulation sites to distinct fascicles.

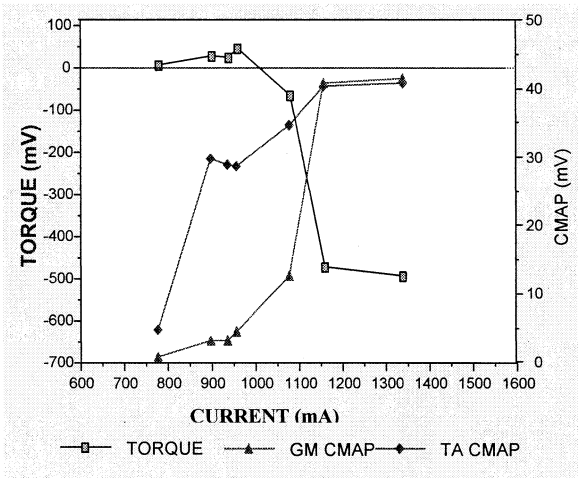

Fig. 10. Muscle excitation (CMAPs) and torque development with increasing stimulation amplitude: dorsiflexion (positive values of torque) at low amplitudes, plantarflexion (negative values of torque) at higher amplitudes. A cuff was placed around the sciatic nerve, tripole active at 0°. (Torque: 100 mV = 2.5 N·mm. GM: gastrocnemius muscle. TA: tibialis anterior muscle.)

The nerve cuff closed properly after release during implantation covering the whole nerve perimeter, to which it adhered well [25]. The wound was closed at the muscle plane with surgical sutures.

C. Torque Development Due to Electrical Stimulation

Stimulation through different tripoles with the cathode in the middle and outer electrodes short circuited to anodes was performed. The CMAPs of tibialis anterior muscle and gastrocnemius muscle were measured. Additionally, the torque development of the plantarflexion/dorsiflexion plane was monitored.

Stimulation was delivered by single pulses through the cuff electrode. By increasing the stimulus amplitude on one selected tripole, we obtained CMAPs of graded recruitment curves from gastrocnemius medialis and tibialis anterior muscles (Fig. 10). Positive and negative torques were measured indicating plantarflexion and dorsiflexion, respectively. Stimulation through the tripole that was located near the peroneal fascicle innervating the tibialis anterior muscle produced dorsiflexion at low stimulation amplitudes. As shown in Fig. 11, stimulation through one tripole of the cuff electrode evoked CMAPs from gastrocnemius muscle with a lower amplitude (8 mV) than from tibialis anterior muscle (24 mV). Thereby, we obtained an initial dorsiflexion of 2.5 N·mm, followed by smaller plantarflexion of 1.25 N·mm (Fig. 11). Increasing the amplitude of pulse stimulation to a supramaximal response level we obtained CMAPs of maximal amplitude (about 26 mV, Fig. 11) for both gastrocnemius and tibialis anterior muscles. The stronger output of the plantarflexor muscles resulted in net plantarflexion of the foot (34 N·mm) despite coactivation of the two muscle groups (Fig. 12). On the other hand, stimulation applied through tripoles near the tibial fascicle produced plantarflexion with increasing degree of force from low to high stimulus intensity (Fig. 13). In all the animals, similar results were obtained without an *a priori* alignment of the nerve cuff electrode. At least one tripole of the cuff electrode was able to excite first the peroneal fascicle to obtain dorsiflexion at low stimulation amplitude, whereas at least two tripoles induced plantarflexion from low stimulation amplitude.

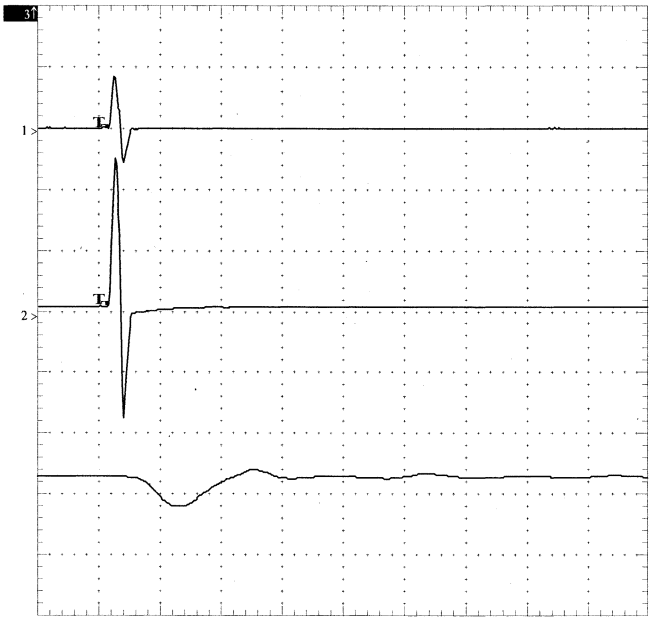

Fig. 11. Simultaneous recordings of CMAP from (top channel) gastrocnemius muscle, (middle) tibialis anterior muscle, and (bottom) ankle torque. Stimulation was delivered by a single pulse through the cuff electrode. Note the initial dorsiflexion (negative torque), followed by smaller plantarflexion (positive torque). Vertical scale: 10 mV/square for CMAPs and 6.25 N·mm/square for torque. Horizontal scale: 10 ms/square.

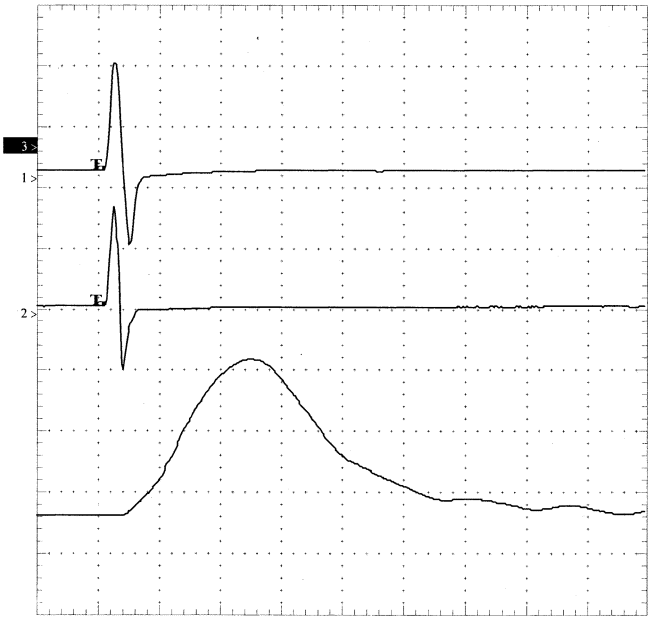

Fig. 12. Simultaneous recordings of CMAP from (top channel) gastrocnemius muscle, (middle) tibialis anterior muscle, and (bottom) ankle torque. Stimulation was delivered by a single pulse through the cuff electrode. The stronger output of the plantarflexor muscles resulted in net plantarflexion (positive torque) despite coactivation of the two muscle groups. Vertical scale: 20 mV/square for CMAPs and 12.5 N·mm/square for torque. Horizontal scale: 10 ms/square.

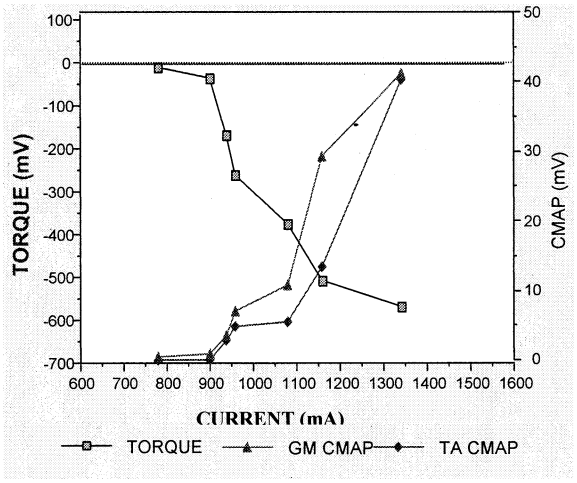

Fig. 13. Muscle excitation (CMAPs) and torque development with increasing stimulation amplitude: plantarflexion (negative values of torque) during the complete stimulation range. A cuff was placed around the sciatic nerve, tripole active at 180°. (Torque: 100 mV = 2.5 N·mm. GM: gastrocnemius muscle. TA: tibialis anterior muscle.)

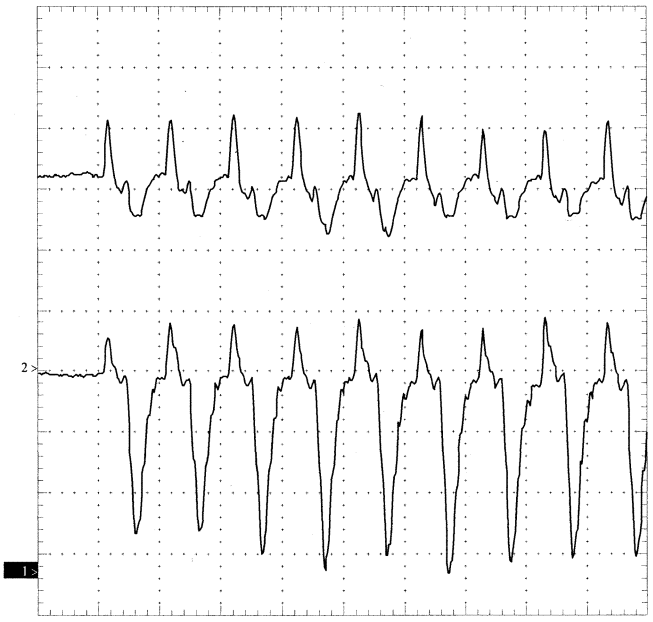

Fig. 14. Recordings of torque. Stimulation was delivered by pulses at 10 pps through two different tripoles of the cuff electrode, producing alternating excitation of the tibial fascicle and plantarflexion (positive torque) and of the peroneal fascicle and dorsiflexion (negative torque). Two recordings from the same preparation are shown; the stimulus amplitude delivered to the tripole inducing plantarflexion was maintained constant, while the stimulus amplitude to the tripole causing dorsiflexion was changed from low level (top recording) to a higher level (bottom recording). Vertical scale: 2.5 N·mm/square. Horizontal scale: 100 ms/square.

Delivering repeated pulses at 10 pps through two different tripoles of the cuff electrode at a fixed amplitude, alternating excitation of the tibial fascicle and of the peroneal fascicle was produced. It resulted in alternating plantar flexion (positive torque) and dorsiflexion (negative torque), respectively, (Fig. 14) that was monitored with the torque measurement device indicating its applicability not only for static experiments but also for dynamic applications.

IV. Discussion

For functional electrical stimulation with multichannel nerve electrodes in neural prostheses, the selective excitation of different nerve fascicles is a prerequisite for functional recruitment of muscles for different movements within a control task like grasping, standing up or walking. Electrode performance should be tested in animal models that allow assessment of the electrode-nerve interface and of performance in the intact musculoskeletal system.

In this study, we present the setup of a measurement device useful to monitor the torque development of the rat foot on the ankle joint with a torque transducer. Noninvasive measurement of joint torques overcomes some of the problems associated with invasive methods. First of all, multiple measurements over the course of an implantation can be performed. No dissection of tendons to prepare single muscles in clamps with attached force transducers [2], [29] is necessary. Isolated muscle force recordings do not provide information on the contribution of the different muscles to a physiologically complex movement or force development in the intact limb. The torque measurement allows the assessment of net motor output by specific stimulation protocols in adequate dimensions. Grill and Mortimer [10], [12] described a system that allowed repeated ankle torque measurement in the cat by means of a transducer device (JR3, Woodland, CA) with three force sensors and three torque sensors. Measurement of the three movement directions and torques is necessary in the cat [18]. The maximum torque measured in the plantar-/dorsiflexion plane was about 3 N·m [12] using 12 polar-cuff electrodes for recruitment studies of different muscles. A setup for custom made single axis torque transducer has also been described for the mouse model [2] where a maximum torque of 25 N·mm was recorded in the plantar-/dorsiflexion plane. We developed a measurement setup for the rat with capabilities to monitor the plantar-/dorsiflexion and the medial/lateral rotation plane. Two different axis of rotation may be decoupled using two single axis transducers and bellows couplings. The measurement setup is quite small and the cost is in a range that allows the purchase even in laboratories with a low number of experiments per year.

We chose the rat animal model with the sciatic nerve as implantation site for cuff electrodes for several reasons. First, the rat allows easy access to the sciatic nerve which has a diameter of about 1.2 mm. It may serve as a good model for a small nerve, comparable to the very distal parts of nerves in humans, e.g., distal branches of the ulnar, medial, and radial nerve that might be interfaced in grasp neural prostheses or the human sacral roots for bladder stimulation, respectively.

Using only single pulses during stimulation experiments with long interpulse intervals in between, we did not induce fatigue of the stimulated muscles. The torque development remained stable and was reproducible over the experimentation time indicating (also) no muscle fatigue and a stable fit of the rat "shoe" device. In order to better distinguish the contribution of the different antagonistic muscles to the isometric torque development, we made EMG recordings of the gastrocnemius and the tibialis anterior muscles. The recordings of the evoked CMAPs were useful in order to correlate the dorsi-/plantarflexion torque with the degree of muscle activation under the different stimu-

lation conditions. The complete setup proved to be easy to use, stable, and provided reproducible data in the acute experiments performed on eight animals.

Cuff-electrodes have been used for a long time to investigate selective stimulation on peripheral nerve trunks. McNeal and Bowman [21] obtained selective activation of foot flexors and extensors in dogs with a rigid cuff electrode having an inner eliptical cross section of 3 $\times$ 4 mm. Other rigid cuffs achieved selective stimulation when snuggly fitting to the nerve, as well as spiral cuffs with a diameter of 2.0 mm for the rabbit sciatic nerve [31] and the cat sciatic nerve [32]. The cuff diameter has been chosen approximately 25% smaller than the nerve diameter [31] to obtain low-excitation thresholds and good selectivity. Otherwise, different studies have shown that snug fitting cuff electrodes produce nerve damage [22]. Self sizing spiral cuffs with a loose fitting made of silicone sheets have reduced that problem [23] but still might produce damage when implanted [19]. Most spiral cuffs were made with means of precision mechanics from monopoles [22] over tripoles [3] for orderly stimulation of skeletal muscle up to 14 electrodes (area of 2 mm^2) [26] for nerve fascicle selective stimulation. Most cuffs assessed for selective stimulation have 12 electrodes arranged to four tripoles around the circumference of the cuff ([10]–[14], [23], [31]–[33]). Their diameters were all above 2 mm, mostly between 3 and 4 mm. A rigid-cuff electrode was designed for smaller nerves with a diameter of 1 mm [17] using also the precision mechanics approach. A thin-film peripheral nerve-cuff electrode was presented by Walter *et al.* [34] in 1995 using a sheet of fluoroethylenepropylene (FEP) with a thickness of 500 μm as substrate and a diameter of about 3.5 mm in an investigation concerning the mechanical biocompatibility of the approach. Small cuff electrodes with 12 stimulation sites are hard to fabricate by means of precision mechanics. A micromachining-based approach allowed the reproducible fabrication of 12 polar cuffs with diameters between 0.7 and 2.0 mm [30]. These polyimide-based nerve cuffs have been tested in chronic experiments on the sciatic nerve of rats previously [25] with an arrangement of three ring electrodes. They remained stable in the site of implantation and caused only a mild foreign body reaction over a 2–6-month implantation period. There were no changes in motor and sensory nerve conduction tests, nociceptive responses and walking track pattern over followup, and no morphological evidence of axonal loss or demyelination. No delamination or stress cracking was observed, as it was reported in another study with "older" types of polyimides [19]. With the experience of that study, the implantation of the slightly modified cuffs with 12 electrodes caused no problems during acute implantation for selectivity investigations. Excitation thresholds of the tibialis anterior and gastrocnemius muscles were below 800 μA at a pulse width of 10 μs with tripolar stimulation. CMAPs increased gradually with increasing stimulation amplitudes. Due to the larger muscle mass of the gastrocnemius muscle in comparison to the tibialis anterior muscle, there is a large imbalance in torque development during stimulation. Even if the levels of the CMAPs are higher in the tibialis anterior (Fig. 12) than those of the gastrocnemius, the latter revailed at high stimulation amplitudes and changed dorsiflexion to plantarflexion. On another stimulation tripole, CMAPs of gastrocnemius muscle were slightly higher than CMAPs of the tibialis anterior and led to a clear plantarflexion from low stimulation amplitudes on (Fig. 13). With the measurement setup, graded recruitment could be observed in the time course of the torque development (Fig. 12). A detailed description of the investigations with regard to the selectivity of the muscle recruitment and the torque development has been published [24].

So far, we have investigated the selective stimulation for foot extension and flexion without and with steering currents [24]. The temporal resolution of the measurement system was sufficiently high to resolve muscle recruitment (Fig. 12) as well as simultaneous repeated stimulation via two tripoles that resulted in alternating torques in plantar- and dorsiflexion (Fig. 14).

The co-contraction of antagonistic muscles during stimulation is a point for further investigations. Detailed investigations of transverse steering currents [7] [14], or prepulses [13], [32], [33] to activate distant nerve fibers should be done to adapt the stimulation protocols to nerves with diameters below 2.0 mm. In order to decrease reverse recruitment, the stimulation protocols should be evaluated with regard to appropriate pulse widths [8], blocking of large nerve fibers [6], and applying quasi-trapezoidal current pulses [5]. These paradigms will be taken into account in future experiments to obtain higher selectivity and hopefully single muscle responses on different tripoles in distinct stimulus amplitude ranges.

The measurement setup presented here is suitable for multiple measurements directly after implantation of cuff electrodes up to chronic values of several months postimplantation using percutaneous leads or a fully implantable telemetric system. The lateral/medial rotation plane will be taken into consideration by a second torque transducer in the measurement setup. Therefore, only a distance ring between the bellows coupling and the mount has to be replaced by the transducer. The system for a noninvasive torque measurement would be also an interesting tool for the evaluation of nerve regeneration and muscle reinnervation in the presence of sieve-like microsystems.

V. Conclusion

We developed a noninvasive measurement setup to monitor the torque development of the intact rat foot after electrode implantation in the plantar-/dorsiflexion and medial/lateral rotation plane. Pilot experiments were performed only in the plantar-/dorsiflexion plane. Static and dynamic torque measurements during electrically induced isometric contractions were performed. The measurement setup delivered stable values over time and interindividually comparable results from different animals. At the end of the first set of experimentation, we conclude that polyimide-based nerve cuffs on small peripheral nerves were suitable for selective control of different muscles during electrical stimulation [24]. In chronic implantations, these nerve cuffs were mechanically harmless for the nerve and allowed stimulation within a wide margin of safety [25]. Future studies will be addressed to the torque development over implantation time to monitor changes of excitation thresholds and torque over time after electrode implantation in chronic experiments with multipolar cuffs for functional selective stimulation.

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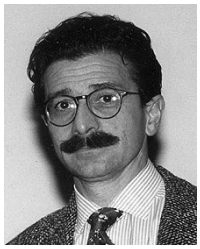

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