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Electromechanical film sensor device for dynamic force recordings from canine limbs

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Introduction

Biomechanical changes in weight-bearing of the limbs of experimental animals, e.g., dogs, can be assessed accurately in vitro with strain-gauge rosette measurements (*Finlay et al.* 1982). The invasive measuring techniques have usually an adverse effect on the biomechanics of the limb in question. The use of the non-invasive force plate method of Kistler is not simple either (*Budsberg et al.* 1987, *Jevens et al.* 1993). The experimental animals should be co-operative and pre-trained. This requirement cannot always be met. An additional difficulty with the force plate measurement is the determination of the walking or running speed of the animal. On account of these reasons, repeated measurements are seldom comparable with each other. However, the majority of in vivo studies have been performed by utilizing the Kistler force plate technique.

In this work we introduce a transducer equipment for the non-invasive measurement of the forces acting upon the limbs of the beagle dogs. The method is based on the electromechanical film (EMF), which is a new type of sensor material (*Kirjavainen* 1987, *Savolainen and Kirjavainen* 1989). EMF has already proved to be versatile in physical activity measurements (*Räisänen et al.* 1992) and in human respiration control (*Siivola et al.* 1993). EMF has some physical properties similar to pietzoelectric polyvinylidene (PVDF), which has been used in many different applications (*Siivola* 1989, *Kobayashi and Yasuda* 1981). The goal of this study was to develop an EMF measuring technique to obtain data on forces acting upon the limbs. To the best of our knowledge EMF sensors have not been previously applied for quantitative dynamic force measurements. Data were obtained from normal dogs, from dogs with their right hindlimb immobilized, and from dogs which had osteotomy at their right hindlimb.

Materials and Methods

Equipment based on the EMF sensors. The forces acting on the limbs of the beagles were measured by attaching an EMF sensor to each limb. EMF consists of a foamized, permanently polarized plastic film with conductive electrodes on its surface (*Kirjavainen* 1987). It forms an elastic electret, which generates on its surface an electric charge proportional to the dynamic force applied on it. In this work the EMF sensors had an active area of approximately 20 mm x 20 mm sealed between two plastic layers (width 25 mm,length 100 mm) with a thickness of the whole structure about 1 mm (Figure 1). The sensors were manufactured in co-operation with Messet Ltd, Kuopio, Finland. The current signals yielded by the EMF sensors were measured

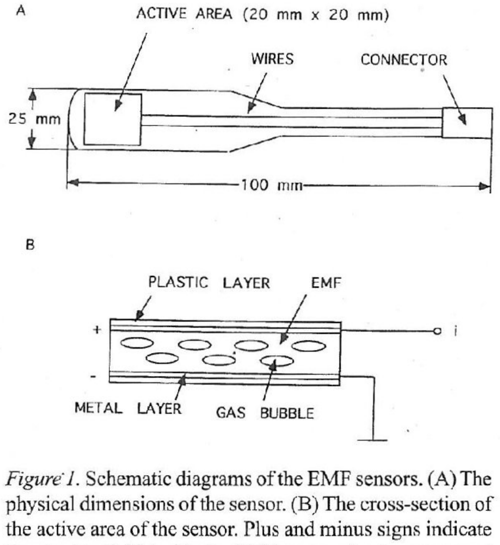

Figure 1. Schematic diagrams of the EMF sensors. (A) The physical dimensions of the sensor. (B) The cross-section of the active area of the sensor. Plus and minus signs indicate the electric field inside the EMF sensor due to permanent polarization. i denotes the output current of the sensor.

using an sampling integrator amplifiers (ACF-2101, Burr-Brown, Tucson, AZ, USA). The signals from the amplifiers were led to a multiplexer and an analog-to-digital converter (DT-2831, Data Translation, Marlboro, MA, USA), which was installed in a 386 AT computer. Measurements were carried out using a Global Lab data acquisition software (Data Translation, Marlboro, MA, USA). To generate the dynamic force the signals were numerically integrated with the Matlab software (Mathworks, Natick, MA, USA). Using a conventional material testing device with cyclic loads (*Parkkinen et al.* 1989), each sensor was calibrated with a series of known weights ranging between 1 kg and 4.5 kg. The calibration curve was computed from the least square fitting of the experimental data. Repeatability of the force measurements was determined using the method based on the intraclass correlation coefficients (ICC) (*Selkäinaho* 1983).

Magnetoresistive sensor for the knee joint angle measurements. The joint angle of the right hindlimb of the beagles was measured by a magnetoresistive sensor (KM110BH/2190, Philips, Eindhoven, The Netherlands). The sensor system consisted of a rotating magnet to measure the magnetic field strength as a function of the angle. The sensor module and magnet was fixed to the angle region of the device. The sensor gives the output voltage within the range +/- 45 degrees. The sensor was covered with dressing gauze and attached to the outside of the knee joint by elastic bandage (Figure 2).

Figure 2. The experimental arrangement. The magnetoresistive sensor was attached to the right hindlimb. EMF transducers were attached to the paws by elastic bandage.

Animals and experimental design. Female beagles (n=11) from the National Laboratory Animal Center (*Kuopio,* Finland) and from the Marshall Farms (North Rose, NY, USA) were used for the experiments. The EMF sensors covered with dressing gauze (Tubinette, Seton, Oldham, England) were fixed under each beagle paw with elastic bandage (Tensoplast, Smith &Nephew, Hull, England) (Figure 2). At the beginning of the experiment, the dogs were 15 weeks and at the end 44 weeks old. The weight of the animals varied between 3.6 kg and 10.8 kg. The animals were weighed before each individual experiment. The force measurements were carried out when the beagles were walking or running on the treadmill (*Arokoski et al.* 1991).The animals were trained on a treadmill belt that was operated at horizontal position, and also at 15° uphill or downhill inclination. Three of the dogs were controls, the right hindlimb of two other dogs were casted with the knee joint immobilized at about right angles according to *Kiviranta et al.* (1987). For three dogs, a 0° right hindlimb osteotomy and for another three dogs, a 30° osteotomy of right hindlimb was carried out. The osteotomy was performed at the age of three months at the proximal part of the right tibia, below the growth plate, according to the principles given by *Johnson and Poole* (1988). The osteotomy was fixed with an AO-T-plate (Stratec Medical, Waldenburg, Switzerland), which was either unbent or bent to 30° angle. Details of the operation technique have been presented elsewhere (*Panula et al.* 1997). X-ray pictures were taken immediately, after one week and four weeks after the surgery. During the experiment the dogs were kept under kennel conditions in the National Laboratory Animal Center (Karttula, Finland) where the dogs were lived in three-dog-fences. The design of the experiment was approved by the Animal Care and Use Committee of the University of Kuopio.

Results

The calibration curve of the EMF sensors obeyed the equation $F = -0.740\ V^2 + 13.540\ V + 1.410$ ($r = 0.999$), where V is the output voltage of the EMF sensor and F is the dynamic force yielded by the material testing device (Figure 3). To assess the repeatability of the measuring system, the value of the intraclass correlation coefficient (ICC) was

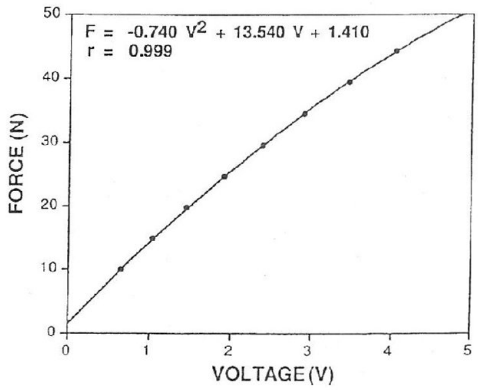

Figure 3. Calibration curve of the EMF force transducer.

Table 1. Repeatability of the EMF force measurement. Five consecutive measurements from one dog were used to calculate the intraclass correlation coefficient (ICC) which indicates the degree of the internal consistency of the measurements.

Treadmill position and speed	Forelimbs	Hindlimbs
Horizontal		
2.5 km/h	0.86[a]	0.91
5.0 km/h	0.87	0.89
7.5 km/h	0.88	0.89
Uphill		
2.5 km/h	0.85	0.91
5.0 km/h	0.86	0.89
7.5 km/h	0.89	0.90
Downhill		
2.5 km/h	0.87	0.88
5.0 km/h	0.86	0.89
7.5 km/h	0.88	0.88

a) 0.71-0.80 moderate, 0.81-0.90 substantial, and 0.91-1.00 almost perfect repeatability.

Table 2. Force (N/kg body weight) measured under the paw of forelimbs and hindlimbs of beagle dogs (n = 2 - 3) during one walking cycle on the treadmill at the speed of 5.0 km/h.

Treadmill position	Forelimbs Force (N/kg body weight)	Forelimbs % of body weight	Forelimbs Change[a)] (%)	Hindlimbs Force (N/kg body weight)	Hindlimbs % of body weight	Hindlimbs Change (%)	Fore-hind ratio
Horizontal	3.74±0.33[b)] n = 10c)	37	–	4.18±0.81 n = 12	42	–	0.90
Uphill	3.74±0.21 n = 10	37	–	4.69±0.70 n = 12	47	11	0.80
Downhill	4.06±0.30 n = 10	41	10	4.03±0.75 n = 12	40	-5	1.01

a) Change related to horizontal treadmill position. b) Mean ± SD. c) n = number of force measurements.

computed. The ICC was calculated both for the fore- and hindlimbs and when the treadmill was operated at different positions, i.e., at horizontal position, or with either 15° uphill or 15° downhill inclination, and at varying velocities (2.5 km/h – 7.5 km/h). The ICC values (n = 18) varied between 0.85 and 0.91. This was a proof of either substantial or almost perfect intraclass correlation (Table 1). During running at the horizontal treadmill position, the load transmission of the total body weight of the control animals was 37% for the fore- and 42% for the hindlimbs, respectively (Table 2). The forces under the hindlimbs increased 11% during uphill running at the same time as the force under the forelimbs did not change. During downhill running, the forces decreased 5% under the hindlimbs and increased 10% under the forelimbs.The +/- 15° inclination of the treadmill changed the ratio of the forces under the fore- and hindlimbs about 10%.

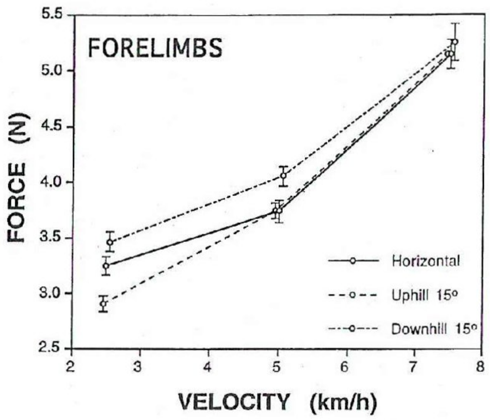

Figure 4. Dynamic forces measured under the paws of forelimbs as a function of velocity. The vertical bars represent the standard error of the mean (SEM).

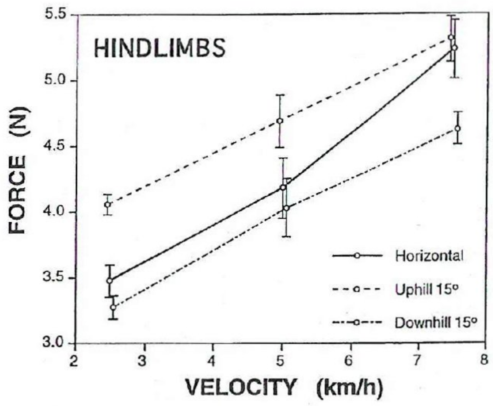

Figure 5. Dynamic forces measured under the paws of hindlimbs as a function of velocity. The vertical bars represent the standard error of the mean (SEM).

The increase of the treadmill (horizontal position) belt speed from 2.5 km/h to 5.0 km/h and from 5.0 km/h to 7.5 km/h, caused a 13% and 27% force increments at the forelimbs (Figure 4) and at the hindlimbs the dynamic force increments were 16% and 20%, respectively (Figure 5). During 15° uphill running, the forces under the forelimbs increased 22% and 27%, and under the hindlimbs 13% and 12%, when the treadmill belt speed increased. During 15° downhill running, the corresponding force increments for the forelimbs were 15% and 23% and for hindlimbs 19% and 13%. The increase of the running speed increased the force under the paws about 10% per 1 km/h.
The dynamic force under the paws of a splinted dog during treadmill exercise was shared between the three weight-bearing limbs. At the speed of 2.5 km/h, the force under the left hindlimb increased 56% (Table 3). Under the right and left forelimbs the force increased 7% and 24%, respectively. The corresponding values with a running speed of 5.0 km/h increased the forces of the left hindlimb by 44%, right forelimb by 16%, and left forelimb by 25%. One month after the removal of the splint, the force measured under the previously immobilized limb was about 90% of that recorded from the control limb.

Table 3. Ratios between the forces measured under the paws of a splinted dog (right hindlimb splinted) and the forces from the same beagle without the cast during one walking cycle on treadmill. Treadmill at horizontal position.

Speed of treadmill		**Forelimb (n = 7)[a]**	**Hindlimb (n = 7)**
2.5 km/h	Right	1.07≠0.06[b]	[c]
	Left	1.24±0.06	1.56±0.23
5.0 km/h	Right	1.16±0.08	
	Left	1.25±0.07	1.44±0.08

a) n = number of observations.
b) Mean ± SD.
c) One month after the removal of the splint, the forces measured under the right hind paw were at the 90% level as compared to the normal animal.

The dynamic forces measured under the paws of control dogs were equal between the right and left hindlimbs (Table 4). One month after the 0° osteotomy, we did not observed any significant difference in weight-bearing between the operated and the unoperated hindlimbs. On the other hand, one month after operation in the dogs that underwent 30° valgus osteotomy, the force under the op-

Table 4. Forces measured under the paws of the hindlimb after osteotomy. The values were related to forces measured under the left (unoperated) paw. Treadmill at horizontal position.

Speed Time after operation	Control (n = 3) Left	Control (n = 3) Right	Fore-hind ratio	Osteotomy 0° (n = 3) Left	Osteotomy 0° (n = 3) Right	Fore-hind ratio	Osteotomy 30° (n = 3) Left	Osteotomy 30° (n = 3) Right	Fore-hind ratio
2.5 km/h									
1 month	1.00	0.89	0.87a)	1.00	0.82	0.95	1.00	0.69	0.97
3 months	1.00	1.04	0.89	1.00	1.08	0.90	1.00	0.93	0.89
10 months	1.00	0.99	0.89	1.00	0.94	0.87	1.00	0.89	0.87
5.0 km/h									
1 month	1.00	0.89	0.87	1.00	1.00	0.91	1.00	0.77	0.93
3 months	1.00	0.93	0.86	1.00	0.93	0.89	1.00	0.94	0.90
10 months	1.00	0.99	0.88	1.00	1.04	0.86	1.00	1.03	0.89

a) Ratios were calculated with the same method as in Table 2.

erated hindlimb was only 69% of the force measured from the unoperated hindlimb at the treadmill belt speed of 2.5 km/h, and 77% at the speed of 5.0 km/h. The decrease in weight-bearing after valgus osteotomy of 30° was significant only one month after the operation. Three and 10 months after the operation, the weight-bearing was about equal in both the operated and the unoperated limbs. The increase of running speed increased the forces under the paws of operated limbs by about 5% per km/h, which was one-half of the value observed in control animals.

The signal from the angle sensor attached to the right hind knee joint was well in phase with the force signal from the right hind limb (Figure 6). The dynamic force signals recorded during one step cycle from the right hindlimb at three different treadmill speeds are given in Figure 7. The curves represent an average of ten consecutive cycles.

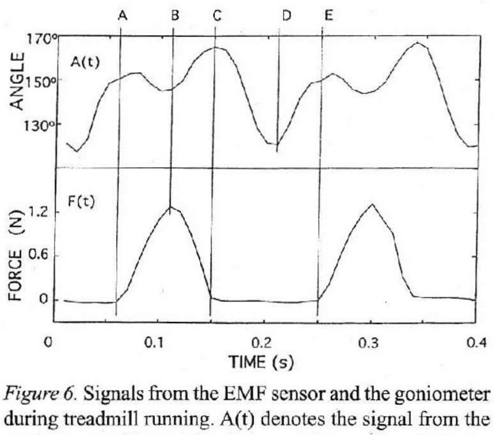

Figure 6. Signals from the EMF sensor and the goniometer during treadmill running. A(t) denotes the signal from the goniometer calibrated for the knee joint angle and F(t) denotes the force from the EMF sensor under the paw. (A) The paw hits the floor, sensor starts to compress and the knee joint undergoes extension. (B) Maximum force is reached and the sensor starts to relax. (C) The sensor is fully normalized and the knee joint reaches its utmost extension. (D) Knee joint is flexed. (E) A new step cycle starts.

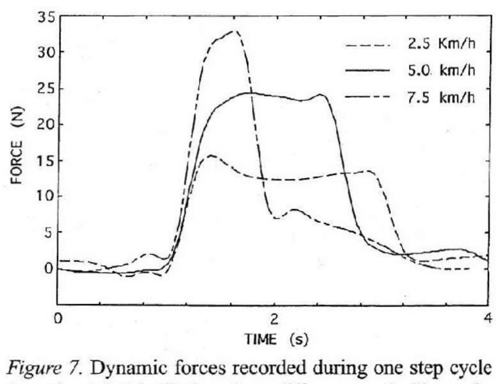

Figure 7. Dynamic forces recorded during one step cycle from the right hindlimb at three different treadmill speeds. Each curve represents the average of ten consecutive cycles.

Discussion

In previous studies, biomechanical forces falling on limbs of experimental animals during walking and running were registered with the Kistler force plate (*Budsberg et al.* 1987, *Dueland et al.* 1977, *Page et al.* 1993). In this study, the forces were measured under the paws of the beagle dogs under different experimental conditions using an equipment based on the EMF sensors. To our knowledge, this kind of technique has not been used earlier to assess dynamic forces quantitatively. The force registration equipment was readily portable. The sensor was attached to each limb by elastic adhesive bandage. The EMF sensors proved to be reliable, enduring and easy to use. The sensors can be designed to have any physical shape or form. The relationship between the applied force and the measured sensor output voltage obeyed the second order polynomial equation up to 45 N force. The measurement system showed a good repeatability in laboratory conditions. Five consecutive force measurements from one dog were done in vivo and we found them highly repeatable (Table 1).

The EMF sensor measurements could be carried out under controlled circumstances, i.e., the speed and inclination of the treadmill belt could be varied deliberately. The running velocity of the animals affected substantially the load put on the sensor. In this study, the running velocity could be controlled accurately. This is probably more difficult with the force plate method. The force plate is usually embedded onto the floor of the laboratory and the animals are trained to walk or run over the plate. In the earlier force plate studies, the determination of velocity of the animals was performed with photoelectric cells attached both to the start and to the end points of the measuring segment. The velocity recorded in this way is not necessarily an accurate representation of the actual limb speed, however, because maintenance of a constant velocity is almost impossible (*Budsberg et al.* 1987). It is also difficult to evaluate separately the force laid on each individual limb. This can be realized, however, with the aid of video camera technique. The force plate measurement gives the vertical force component, which is more consistent than the horizontal force component (*Dueland et al.* 1977).

One disadvantage of the EMF sensor used in this study was that it measures only the dynamic forces while the force plate responds also to static force. The sum of dynamic forces under the weight-bearing limbs of the dog, e.g., under the left forelimb and the right hindlimb, reached about 80% of the body weight of the dog. This is probably due to the fact that the EMF sensor measures only the force perpendicular to the transducer plane, which was not always horizontally situated during the gait cycle. Another reasonable explanation is that the site of the paw, to where the sensor was attached, did not carry the full weight of the animal. Also the bandages might have an effect on the sensitivity of the sensors. The EMF sensors were always identically attached to the paws and the reproducibility of the measurements was good. However, it must be emphasized that data represent relative, not absolute force values. In the study of Page et al. (1993) the sum of forces on the forelimb and the contralateral hindlimb (left and right) did not match for the body weight either. They interpreted this to indicate that the dog had had three limbs in contact with the ground during a major portion of the gait cycle (at walking speed 1m/s). Also our results show that during the gait three limbs are in contact with the ground simultaneously. Still there are phases during which only two limbs contact the ground, i.e., the weight-bearing of these limbs should approach the body weight.

We observed at the horizontal position of the treadmill (belt speed ca. 1.4 m/s) that the load transmission of the body weight of running control beagles was higher through the hindlimbs (42% of the body weight) than through the forelimbs (37%). The body weight of the dogs ranged from 5.6 kg to 10.7 kg. The weight-bearing pattern of a greyhound dog appears to be different. In greyhounds, more weight was laid, at a walking speed 1 m/s, on the forelimbs than on the hindlimbs, i.e., forelimbs carry 53 to 65% and hindlimbs 24 to 41% of the body weight, respectively. The body weights of greyhounds ranged from 25 kg to 35 kg (*Dueland et al.* 1977). In the study of Budsberg et al. (1987) forelimbs carried 60% and hindlimbs 40% of the body weight which ranged from 8.6 kg to 58.5 kg. The breed of the dogs was not indicated. The differences compared to our observations can be explained by the different experimental conditions (running vs. walking) and by different anatomy and the distribution of weight over the dog bodies.

The force measurement during right hindlimb immobilization gave very much predictable results. The animals laid more weight on the uncasted left hindlimb. The forces under the left hindlimb increased by 44% to 56% (Table 3). At the same time the weight-bearing increased on the forelimbs by 7% to 25%.

At all time points, the relative force measured from beneath the 30° v algus osteotomy limb, was lower than the force recorded from the control animal or the sham operated dog (treadmill belt speed 2.5 km/h) (Table 4). One month after the 30° v algus osteotomy, the weight-bearing on the right hindlimb was 31% less than in the control dogs. On the other hand, the weight-bearing of the operated limb gradually approached the normal level later on during the experiment.

Egger et al. (1993) studied after a bilateral osteotomy of the canine tibia the effects of axial dynamization on bone healing. They measured pre- and postoperatively the vertical forces under the hindlimbs with the force plate method. In three weeks, the postoperative forces matched the values measured preoperatively. This was an indirect evidence of the bone healing which allowed the animal to load the limb normally. Keeping this in mind, therefore, in this study the possible role of the postoperative pain at the osteotomy site in weight-bearing deserves a comment. As discussed above, one month after the operation we found a significant difference in weight-bearing between the 30° v algus operated animals and the control dogs. It is possible that because of soreness at the osteotomy area, the dog was not willing to load the operated limb normally. At the same time interval, a less evident difference was observed between the 0° valgus operated (sham) animals (Table 4). Also, when the speed of the treadmill belt was increased from 2.5 km/h to 5.0 km/h, the weight-bearing of the operated limbs increased in the same way as in the controls. Further, after one month, we found that all the osteotomies were healed both clinically and radiologically. Therefore we think that the postoperative pain did not explain the difference observed between the operated dogs (i.e. 30° v algus osteotomy) and control dogs. Instead, we think that the reason for altered weight-bearing between the operated and the control dogs was the initial shortening of the operated limb after 30° v algus osteotomy. At the operation, we removed a wedge-shaped bone block from the tibia. The lateral base of the block was 7 mm. The resulting 30° degree valgus position of the tibia shortened the limb. Apparently one month after the osteotomy there still was an imbalance in the pelvis-hindlimb region which the animal could not compensate during gait. Later during the experiment, the dogs appeared to adapt to altered loading because the weight carried by the operated limb approached the load carried by the unoperated limb (Table 4). The length of the operated hindlimb gradually reached the length of the control hindlimb (*Panula et al.* 1997)

According to our experience, the EMF sensor device was a reliable and convenient method for the measurement of dynamic forces acting on the canine limbs. Effects of running, immobilization and osteotomy on weight-bearing limbs could be readily monitored.

Summary

An equipment based on the electromechanical film (EMF) sensors was designed for the measurement of forces acting upon canine limbs. EMF forms an elastic electret, which generates on its surface an electric charge proportional to the the force applied on it. The EMF sensors were calibrated using a conventional material testing device with cyclic loads. The beagles were trained on a treadmill working at horizontal position or with either 15° uphill or downhill inclination. The treadmill belt speed varied from 2.5 km/h to 7.5 km/h. The force under the canine paws varied depending on the inclination of the treadmill. When the dogs ran uphill, weight-bearing on hindlimbs increased 11% but the weight-bearing on forelimbs did not change. Downhill running increased weight-bearing on forelimbs by 8% and decreased weight-bearing of the hindlimbs by 5%. Immobilization of the right hindlimb increased weight-bearing on both forelimbs by 7-25% and on the left hindlimb by 56%. One month after a 30° valgus osteotomy operation at the right tibia, the dynamic force recorded from the operated hindlimb was 69% of the control value. Three months after osteotomy, the weight-bearing of the operated limb approached normal situation. Our results suggest that the EMF sensor is a reliable method for the measurement of dynamic forces acting on the weight-bearing limbs of the dogs.

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