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Key Words

Strain gage
Goat
Force
Tibia

Abstract

In vivo strain measurements at 8 locations on the tibia of the goat were performed. Successive in vitro calibrations were used to determine the assumed linear relationship between the measured strain signals and the external loads (3 forces and 3 moments) at the tibia. For the reconstruction of the in vivo external loads from the strain data, a transformation matrix was created from the calibration experiments, using 'singular-value decomposition'. The method is a reliable technique for measuring in vivo loads during functional gait and gives reproducible results.

Introduction

To assess the forces and moments acting on long bones during activity, different methods can be used. A possible way is to measure bone strains directly under in vivo conditions with strain gages attached to the surface of the bone [1]. From the strain data and the elastic properties of the bone, the stresses can be calculated, and subsequently, employing beam theory, the forces on the bone can be determined [2]. The accuracy of this method depends on the reliability of the values for the elastic properties of the bone tissue used and the accuracy of geometrical measurements of the bone. In addition, a simplification with respect to the load configuration is assumed with this method, since transverse forces are neglected.

Biewener et al. [3] presented a comparison between the method described above and a load reconstruction from a dynamic model of the bone, employing film camera assessment of limb motion as it passed over a force plate. The results of this indirect method were significantly different

from the reconstruction of a simultaneous strain measurement.

Rybicki et al. [2] used strain recordings and beam theory to reconstruct the external loads on some long bones of the horse, using 4 single strain gages. With strain rosettes (3 gages at 1 location oriented at different angles) more information with respect to the direction and magnitude of the principal strains can be obtained. Carter et al. [4] used 1 rosette and 2 single gages to measure the strains at 3 circumferential locations of the femoral shaft of a dog. Using beam theory and a finite element model, the axial load, the 2 bending moments and the torsional moment were reconstructed from the strain recordings.

In this paper a method is introduced, whereby the full three-dimensional external loads on a bone can be reconstructed from in vivo strain measurements combined with a successive in vitro calibration technique. The method is suitable to reconstruct all 6 load components and is not dependent on the reliability of bone elastic properties or geometry measurements.

Materials and Methods

Strain Gage Unit Preparation

T rosette strain gages (FCA-1; Tokyo Sokki Kenkyujo Co., Japan) were used for the *in vivo* strain measurements. These gages consist of 2 perpendicular strain gage elements. The grid size of the gage element was 1.0 mm long and 0.7 mm wide. The elements have Cu-Ni foil grids sealed within an epoxy carrier with a base of 4.5 mm. The T rosette strain gage was supplied by the manufacturer with 4 fine lead wires.

The first step in preparing the strain gage units was to modify the lead wires such that both gage elements were connected to each other (fig. 1), and the whole unit of 2 gages was provided with 3 Teflon-coated wires (UT3607; Habia, Breda, The Netherlands). For this purpose the Teflon-coated wires were cut to a length of 120 cm, and the tips were stripped and tinned. The Teflon-coated wires were aligned on the fine lead wires near the base of the strain gage. The fine lead wires were carefully bent, soldered and excess lead wires were trimmed off. A minute amount of soldering flux (Rosin soldering flux; Micro-Measurements, Raleigh, NC, USA) was applied for good flux and solder connection. The Teflon-coated wires used in one and the same strain gage unit were of exactly the same length and twisting.

Mechanical and chemical protection of the strain gage elements and the solder connection were secured by covering the area with a coating. First the gage surface, the solder connection and about 0.5 cm of the Teflon-coated wires were primed with RTV primer (RTV primer No. 1; Micro-Measurements) and then coated with a thin layer of silicone rubber (3145 RTV coating; Dow Corning Corporation, Midland, Mich., USA). The overall size of the strain gage unit was approximately 70–80 mm² (fig. 2). Finally each unit was tested, individually packaged and gas sterilized.

Operation

The strain gage units were attached to the left tibia of a goat of 41 kg weight under general anaesthesia. An anterior/medial incision was made, starting 10 cm under the stifle joint. Correct homeostasis of the soft tissue was achieved by electrocautery. The periosteum was exposed, and an incision of 8 cm was made for the reflection of the periosteum and the muscles. Thus by elevating the soft tissues, the strain gage units could be placed on the anterior, medial and posterior surfaces of the cortex. The bone surface was scraped, dried and cleaned with alcohol. A small amount of adhesive (n-butyl-2-cyanoacrylate; Braun Melsungen, FRG) was applied to the bonding surface of the strain gage units and the cortex. The units were held for 1 min under finger pressure. Altogether 2 units were glued to the anterior, 4 to the posterior and 3 to the medial surfaces of the mid/distal part of the diaphysis. After bonding the units were again tested.

The Teflon lead wires were secured and routed subcutaneously and departed percutaneously high in the flank of the goat. The ends of the wires were coiled and packed up in a small pocket fixed to the goat. All incisions were sutured and closed.

In vivo Measurements

After a recovery period of 3 days, the *in vivo* strains were recorded. The experimental set-up consisted of a connector box with Wheatstone bridges in a half-bridge configuration and provided with 8-metre-long shielded cables, a signal conditioning equipment and a computerized data acquisition system. A personal computer was fed with the strain data sampled by an A/D converter with a sampling frequency of 10 Hz during the measurements. The personal computer

Fig. 1. Adjustment of the strain gage. The gages were connected to each other and provided with 3 Teflon-coated wires.

Fig. 2. The strain gage unit (right side) as it was attached to the tibia of the goat.

provided further storage and processing of the data. The connector box was tied to the back of the goat using a small jacket. The goat was led, and strains were recorded during several functional gait patterns (slow walking, fast walking on a rough rubber floor and jumping on a bench 50 cm high).

Immediately after the *in vivo* measurements the goat was killed, and the zero strain levels of the 8 strain signals were recorded for 0.5 h. During this recording the linear drift of the zero strain level appeared

Fig. 3. The coordinate system used for the tibia.

Fig. 4. a, b Transverse forces ($\circ = F_x$; $\Delta = F_y$), axial force ($\square = F_z$), bending moments ($\bullet = M_x$; $\blacktriangle = M_y$) and torsional moment ($\blacksquare = M_z$) during two gait cycles in slow walk.

to be less than 100 microstrain/h. All in vivo strain data were corrected for zero strain and drift.

In vitro Calibration

Within 24 h the in vitro calibration was performed. For this purpose the tibia was isolated and fixed with acrylic cement in two fixtures. In order to reconstruct the in vivo loads, a coordinate system was defined relative to the geometry of the tibia with the origin in the intercondylar fossa of the tibial plateau and the z-axis in the longitudinal direction of the bone (fig. 3). In this coordinate system we defined two transverse forces F_x and F_y , an axial force F_z , two bending moments M_x and M_y and a torsional moment M_z .

An MTS loading machine (MTS Systems GmbH, Berlin, FRG) was used to apply the axial force and the two bending moments. Both transverse forces and the torsional moment were applied with pulleys

and weights. The loads were adjusted for about 80 calibration experiments, whereby the strains as a result of 1 or 2 load components were measured for each calibration.

All strains were linearly dependent on the external loads, so a linear transformation (T) between the 8 strains and the 6 load components should exist:

$$\begin{bmatrix} T_{11} & \cdot & \cdot & \cdot & \cdot & T_{16} \\ \cdot & & & & & \cdot \\ \cdot & & & & & \cdot \\ \cdot & & & & & \cdot \\ T_{81} & \cdot & \cdot & \cdot & \cdot & T_{86} \end{bmatrix} \begin{bmatrix} S_1 \\ \cdot \\ \cdot \\ \cdot \\ S_8 \end{bmatrix} = \begin{bmatrix} F_x \\ F_y \\ F_z \\ M_x \\ M_y \\ M_z \end{bmatrix}, \quad (1)$$

with T_{ij} the components of the transformation matrix, S_1 – S_8 the 8 strain signals and F_x – M_z , the 6 load components.

Fig. 5. a-f Average peak components ($\pm$ standard deviation) of the loads during different activities.

So the calibration experiments resulted in 80 sets of 6 load components and 80 corresponding sets of 8 strains. The matrix which fits best all these experiments can be determined by using 'singular-value decomposition' [5]. The method works out such that a transformation matrix T is constructed which minimizes the expression

$$\sum ||T_{11}S_1 + T_{12}S_2 + T_{13}S_3 + \dots + T_{18}S_8 - L_1||, \quad (2)$$

where T_{ij} are the components of the transformation matrix, S_j the

strain values and L_i the load components corresponding to F_x - M_z and the summation takes place over all calibration experiments. In other words, the procedure gives a transformation matrix with a least-square fit through all calibration experiments.

The in vivo strains were successively multiplied by the transformation matrix T (equation 1) to reconstruct the in vivo loads F_x , F_y , F_z , M_x , M_y and M_z .

Results

The reconstruction of the six load components during two gait cycles of normal (slow) walking is shown in figure 4. The peak force value in the z-direction during the stance phase was 350 N, which is approximately 85% of the weight of the goat. The minimal force during the swing phase was 60–70 N (15–17% body weight) for F_z . The two transverse forces reached peak levels during the stance phase of about 20 N (5% body weight) for F_x and 4 N (1% body weight) for F_y .

The reconstructed bending moment around the y-axis ranged from zero during the swing phase to a peak value of 17 Nm during the stance phase. Around the negative x-axis this range was from 2–12 Nm. The torsional moment was much lower and reached peak values up to 2.5 Nm during the stance phase.

In more strenuous functional activities, like fast walking or jumping, the peak values reached higher levels. Figure 5 shows a comparison of the peak values for four different activities: standing, walking, fast walking and jumping. Fast walking did not increase the loads significantly. For jumping, the average peak values reached 2–3 times those of normal or fast walking. The variability in the transverse forces F_x and F_y and also in the bending moment M_y and torsional moment M_z increased drastically with jumping (fig. 5a, b, e and f).

Discussion

The present method, using singular-value decomposition to find the best fit through a set of calibration experiments, proved to be a reliable technique for a direct load

reconstruction from strain measurements. The advantages are that no further information is needed with respect to the precise geometry of the bone and its elastic properties. Furthermore, the method is not extremely sensitive for the exact location of the strain gage attachments.

A disadvantage is that the number of strain signals, needed for a complete reconstruction of all 6 load components, is quite large. In theory at least 6 strain signals are needed, but in practice a redundant system of 8 or 9 signals gives better accuracy.

A recovery period of 3 days was relatively short, and it was observed that the goat did not walk soundly. A second experiment (of which the data are not available yet) showed that a recovery period of approximately 12 days is long enough to restore normal gait.

The present results indicate that the longitudinal force during walking comes close to body weight. The transverse forces are relatively small. The bending moments are in the range of 10–20 Nm during slow and fast walking which is high when compared with in vivo studies of other bones [2, 4]. During jumping, not only the average peak load components increased by a factor of 2–3, but also the variation of the peaks of F_x , F_y , M_y and M_z increased drastically.

In normal walking or standing the variation in different sets of measurements is small, indicating a good reproducibility of the reconstruction experiment.

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