EFFECT OF CANE USE ON TIBIAL STRAIN AND STRAIN RATES[1]

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The effect of cane ambulation on hip biomechanics has been well studied, but its effect on tibial strains and strain rates is not known. To test the hypothesis that cane use may lower tibial strain and strain rates during walking, percutaneous axial extensometers were mounted on the right medial cortex of the midtibial diaphysis in seven male volunteers. In vivo peak-to-peak axial tibial strains and strain rates were measured for ipsilateral and contralateral cane usage and compared with a no cane control. Cane-assisted ambulation was not found to significantly lower strain magnitudes; however, tibial strain rates were significantly lowered by both ipsilateral and contralateral cane usage. We conclude that either ipsilateral or contralateral cane usage may be beneficial when lowering tibial strain rate is desired, such as in the treatment of tibia stress fracture or osteoarthrosis of the knee.

Key Words: Strain, Stress Fractures, Bone, Biomechanics, Ambulatory Aids, Canes

Bone undergoes deformation during physical activity. Because these deformations are small, they are expressed in units of microstrain (10^{-6}). During walking, tibial strains are in the range of 250–600 microstrains in compression and tension.[1,2] A strain of 25,000 microstrains (2.5% deformation), is sufficient to fracture the tibia. High bone strains or strain rate have been implicated in the cause of stress fracture[1] and osteoarthrosis of the knee.[3] Lowering the strain or strain rate plays a role in the treatment of both.[3,4]

Objectives: Upon completion of this article, the reader should be able to (1) recognize the biomechanical role of canes in treatment of arthritic diseases of the hip and problems below the hip; (2) differentiate between contralateral and ipsilateral cane use in unilateral lower limb problems; (3) describe the effect of cane use on tibial strains and strain rates. **Level:** Update

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To date, most measurements of ambulatory aid effects on human gait have been made by externally placed devices such as force plates and accelerometers. In vivo internal strain measurements of bones in humans has been performed only four previous times, and the effect of an ambulatory aid on the internal strain environment of the human tibia has never been examined.[1,2,5,6]

The purpose of this study was to measure the effect of cane ambulation on in vivo human tibial strains and to evaluate whether canes may play a role in the treatment of the most common stress fracture among runners and military recruits, the tibial stress fracture,[7] and in the treatment of osteoarthrosis of the knee.

MATERIALS AND METHODS

The Extensometer

Before this study, the percutaneous extensometer used in this experiment to measure in vivo, the range of the peak compression to the peak tension

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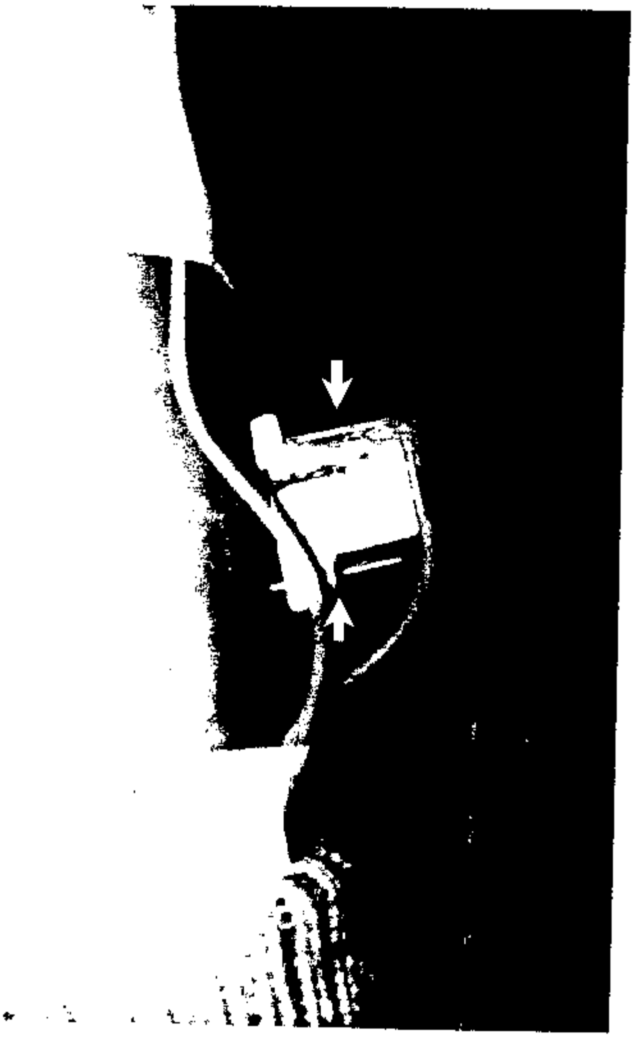

Figure 1. The percutaneous extensometer mounted on two K-wires (arrows), inserted to a depth of 4 mm in the medial cortex of the tibia.

human tibial axial strains and maximum compression strain rates was developed at the Bone and Joint Center, Henry Ford Hospital, Detroit, Michigan. In vitro mechanical testing of the device and in vivo animal tests were carried out there before human studies.

The percutaneous extensometer (17.6 g) was designed to be mounted on two trocar tipped treaded 2-mm K-wires inserted 19.1 mm apart, to a depth of 4 mm within cortical bone (Fig. 1). The extensometer consisted of a phosphor bronze beam (0.254 mm thick) with two pairs of uniaxial strain gages (CEA-09–062UW-350), one pair bonded to the top and the other bonded to the bottom surface of the beam. The gages were wired as a full Wheatstone bridge. The beam was supported by an aluminum block (19.1 × 19.1 × 9.53 mm).

The in vitro relationship between the percutaneous extensometer and bone strain was determined in two calibration experiments. The first was placement of the extensometer on a dog tibia in vitro with a strain gage (CEA-13–06UW-350, Micro-Measurements Group, Raleigh, NC) mounted between the extensometer mounting pins. To determine the relationship between the two methods of estimating strains, the extensometer and bone construct was loaded axially for 200 cycles in an INSTRON 8501 Servohydraulic loading system with a 0.25-Hz, 1500-N, haversine wave function. Extensometer and strain gage output voltages were collected at 100 Hz using a Labtech Notebook (Wilmington, MA). There was an $r^2 > 0.96$ correlation between the output of the strain gage and the extensometer. The second calibration experiment was placement of the extensometer on a composite bone (Pacific Research Laboratories, Vashon, WA) with a strain gage (CEA-06–062UW-350) mounted between the pins on which the extensometer was mounted. This construct was used to examine the effect of impact loading on the readings of the extensometer. Data were collected at 20,000 samples per second using a Nicolet 420 digital oscilloscope (Nicolet Instruments, Madison, WI). Two tests were performed: (1) The composite tibia was bent by hand as rapidly as possible (>2 Hz) in the anteroposterior direction for 100 repetitions. (2) The distal end of the composite tibia was manually struck 100 times against a solid object covered with a 1.25-cm layer of packing foam to produce shock accelerations in the construct similar to those which would be experienced in field applications. In the experiment of rapid manual bending of the composite tibia construct, the correlation between the strain gage and extensometer output was $r^2 = 0.998$. However, the impact experiment with the composite tibia showed only an $r^2 = 0.4$ relationship between the unfiltered output of the extensometer and the strain gage. After Fourier filtering at 3.7 Hz, the relationship between the two data sets was improved to $r^2 = 0.7$.

The relationship between the percutaneous extensometer and bone strain in vivo was determined on a skeletally mature foxhound. The percutaneous extensometer was mounted axially on the right radius of the dog under general anesthesia, with an unstacked rosette strain gage (EA-06–01-RJ-120, Micro-Measurements Group) bonded directly on the bone between the extensometer mounting pins which were placed to a depth of 4 mm in the bone. Strain recordings were made with the dog walking on a hard surface at a speed of 4 km/h. The extensometer mounted on the dog radius had an $r^2 > 0.8035$ between its output and the axial readings of the strain gage. There was no heel strike transient artifact present.

Study Design

Seven male subjects from the scientific staff (ages 23–50) volunteered to be the subjects. The anthropomorphic data of the subjects is summarized in Table 1. All signed informed consent. The experimental protocol was approved by the Human Rights Committee of the Hadassah University Hospital,

TABLE 1
Anthropomorphic data

Subject no.	Age	Height (cm)	Weight (kg)	Shoe size (EU)
1	35	170	73	44
2	28	180	70	44
3	50	183	85	45
4	23	176	63	43
5	23	182	78	43
6	43	169	74	43
7	29	185	93	40

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Surgical implantation of the percutaneous extensometer was performed on an outpatient basis at the Hadassah University Hospital on each of the subjects as follows. Surgery was performed in the morning and removal of the percutaneous extensometer was performed on the same day after completion of the data collection. Prophylactic intravenous cefonicid was given before each surgery. The right leg was prepared and draped at the level of the midshaft of the tibia to facilitate placement of two K-wires along the medial border. Surgical anesthesia was achieved by a local injection of 2.5 ml of 1% lidocaine and 2.5 ml of 0.25% Marcaine to the skin, to subcutaneous tissue and to the periosteum of the tibia. Two K-wires with trocar tips (2.0 mm diameter) were inserted into predrilled holes (1.5 mm diameter) to a depth of 4 mm in the anteromedial tibial midshaft. The proximal and distal K-wires were placed parallel to each other and perpendicular to the surface of the tibia 19.1 mm apart along the long axis of the tibia. Consistent K-wire placement was achieved through the use of drill and wire placement guides. The extensometer was mounted on the two K-wires above the skin, at a measured distance of 1 cm from the cortical surface of the tibia and tightened with set screws. After the placement of the extensometer, the K-wires were cut so each wire protruded 2 mm above the extensometer.

Before beginning strain measurements, each of the seven subjects was fitted with a cane by a registered physiotherapist who also demonstrated its use. Subjects then did two supervised practice trials. Recordings were made while the subject walked alongside a pacer on a concrete surfaced course measuring 20 m. All subjects wore standard Israeli two-layer infantry boots (1600 g), with midsoles constructed from 45 durometer polyurethane and outer soles constructed from 95 durometer polyurethane. Results from three separate trials were recorded for each subject: (1) walk without cane; (2) walk with standard cane in right hand (ipsilateral to extensometer); (3) walk with standard cane in left hand (contalateral to extensometer).

With ipsilateral cane usage, a cane in the right hand was applied to the ground just prior to the time of right heel strike. With contralateral cane usage, a cane in the left hand was applied to the ground just before the time of right heel strike.

The duration of time required to complete each trial was recorded in seconds at the time of the experiment. For each subject, the range between the highest tibial axial compression and highest tension strains (peak to peak) during a gait cycle and the maximum tibial axial compression strain rate value were calculated for three steps for each treatment group. Interpersonal differences in gage readings were corrected by computing subject specific correction factors for both strain magnitudes and strain rates. This was determined by computing an overall experimental average based on the total of 54 steps sampled in the experiment. Subject average values were then computed based on the nine individual steps recorded for each subject. The correction value was determined by the following formula, which was used to correct individual trials:

Subject correction factor

= Experimental average value/subject average value

Duncan ANOVA with an alpha set to 0.05 was used to examine for statistically significant differences in the time to complete the walk, for the corrected range of the peak compression to peak tension axial strain values, and for the corrected maximum axial compression strain rates between the control and two treatment groups.

RESULTS

None of the subjects experienced pain from the percutaneous clipgage during the experimental recordings and none had an antalgic gait. The low pass Fourier filter at 2 Hz was successful in filtering out the heel strike transient in subjects 1 to 6. This transient occurred before the maximum axial compression strain reached during the gait cycle. In the seventh subject, the data were discarded because the low-pass Fourier filter at 2 Hz resulted in significant data loss.

Table 2 compares the time required to complete the 20-m course for the three trial groups. There was no statistically significant difference by ANOVA between any of the three groups for the time required to complete the course ($P = 0.1$).

Table 3 compares the range of the peak compression to the peak tension tibial axial strains for the three trial groups. Values are listed in microstrains (μs). There was no statistically significant difference by ANOVA between any of the groups ($P = 0.16$).

Table 4 compares the maximum tibial compression tibial strain rates for the three trial groups. Values are listed in microstrains per second (μs/s). By ANOVA, a statistically significant lower strain rate was seen when either ipsilateral or contralateral cane usage was compared with the no cane control ($P = 0.00002$). No significant difference was detected when ipsilateral usage was compared with contralateral usage.

TABLE 2
Time required to complete 20-m walk according to ambulation with and without a cane

Subject no.	Time to walk 20 m (s)		
	Without cane	Ipsilateral cane	Contralateral cane
1	5.6	8.4	9.4
2	5.7	7.1	6.5
3	7.9	8.9	8.0
4	5.9	7.4	7.5
5	7.1	11.8	14.6
6	6.1	6.9	7.7
Average	6.4	8.4	8.9
SD	0.9	1.8	2.9

$P = 0.1$.

TABLE 3
Range of peak compression to peak tension tibial axial strain according to walking with and without a cane

Subject no.	Range of the peak compression to peak tension axial strain (μs) while walking		
	Without cane	Ipsilateral cane	Contralateral cane
1	142.5	140.8	159.0
	171.7	125.3	197.4
	173.3	151.0	199.8
2	156.7	158.6	177.2
	184.5	147.9	152.9
	186.0	163.1	134.1
3	155.9	193.6	162.6
	182.8	162.0	169.4
	200.5	145.6	88.5
4	207.6	246.8	77.8
	185.2	245.2	92.7
	135.8	204.4	65.2
5	168.6	178.6	199.7
	169.0	152.3	165.8
	169.0	145.1	112.8
6	113.6	109.3	147.4
	119.4	181.0	178.3
	230.5	192.7	188.7
Average	169.59	169.07	148.29
SD	29.63	36.90	43.50

$P = 0.16$.

TABLE 4
Maximum tibial axial compression strain rates according to walking with and without a cane

Subject no.	Maximum tibial axial compression strain rate while walking (μs/s)		
	Without cane	Ipsilateral cane	Contralateral cane
1	1353.7	736.6	783.9
	1287.0	608.7	739.4
	1284.2	519.8	820.0
2	2515.3	1326.5	1122.5
	1978.1	1437.2	1499.2
	1797.5	1444.3	1248.1
3	714.3	269.3	394.8
	702.6	269.3	386.4
	714.3	281.0	392.2
4	733.6	638.6	825.3
	756.5	589.5	920.3
	694.3	995.6	789.3
5	629.0	368.2	290.0
	651.8	244.4	329.2
	469.3	280.7	413.9
6	879.3	1437.2	954.7
	784.1	1444.3	869.4
	918.6	1402.7	839.9
Average	1048.0	794.1	756.6
SD	554.1	491.5	338.0

$P = 0.00002$.

DISCUSSION

To evaluate the effect of cane assisted ambulation on tibial strains and strain rates in multiple subjects, it was first necessary to develop an in vivo strain measuring technique that was less invasive than direct bonding of a strain gage to the tibia used in prior experiments.[1,2] For this purpose, an extensometer externally mounted on two K-wires placed percutaneously to a depth of 4 mm into the tibial cortex was developed. The device was placed to measure axial strains on the medial cortex of the midtibial diaphysis, the most prevalent site of stress fractures among runners and military recruits.[7] Fourier filtering of the extensometer data was necessary because of the presence of a heel strike transient caused by the extensometer design. This transient occurred before the peak compression and peak tension strains during the gait cycle and therefore did not interfere with data analysis.

In this study there was no significant difference found between the range of the peak compression to the peak tibial axial strain of subjects walking with and without canes. The average value in the non-cane-ambulation group was approximately 170 microstrains, roughly one third of the range of the peak axial compression to peak axial tension tibial strain reported by Milgrom et al.[8] when recording were made from a strain gage directly bonded to the human tibia in one subject. This difference can be attributed to the fact that the current experiment used an extensometer mounted externally on two K-wires to measure strains and that in vivo calibration of the device was done on canine rather than human bone.

In contrast to strain magnitudes, the maximum axial compression strain rates in this experiment were found to be significantly reduced with either ipsilateral or contralateral cane usage. A possible explanation for this can be found in the timing of cane contact with the ground observed in this experiment. During both ipsilateral and contralateral cane use, cane contact with the ground was observed to occur just before heel strike. Although this apparently did not significantly lower the force of impact, it may have spread the impact over a greater time period and thereby decreased the strain rate. This is analogous to a parachutist, who by using proper technique is able can spread the impact of landing out over time and thereby protect the bones from fracture.

In a previous study, Burr et al.[1] recorded tibial strains at the same anatomic site as in this experiment but used a rosette strain gage bonded directly to the bone. They found that changing the rate of ambulation from a walk at 5 km/h to a jog at 10 km/h caused tibial strain rates to more than double and strain magnitudes to also increase significantly. In our study, cane usage was not found to have a statistically significant effect on the time it took for the subjects to complete the walking course. Therefore, any effect of cane usage on the tibial strain environment in this experiment cannot be attributed to the cane functioning simply as a biologic brake.

It is important to understand the role canes may play in the management of other neuromuscular and musculoskeletal diseases. By adding a third point of contact with the ground, a cane widens the base of stance and increases stability.[9] For older people and individuals with problems of balance, this may add the extra degree of stability needed to maintain ambulatory freedom and prevent falls leading to serious injury. For these conditions, it makes little difference on which side of the body the cane is used, so long as the person has no preexisting upper limb deformities.[10]

Such is not the case in the treatment of arthritic diseases of the hip. In a 1956 presidential address to the American Academy of Orthopedic Surgeons entitled "Don't Throw Away the Cane," Dr. Walter Blount presented a series of works published by the German scientist Friedrich Pauwels.[11–13] In this work, Pauwels presents a rationale for utilizing a cane in the hand contralateral to the affected hip. In this explanation, joint forces at the hip are seen as the result of two opposing torques that if unbalanced tend to cause rotation at the hip in the frontal plane. The first torque is the body weight acting through the relatively long lever arm between the body's center of gravity and the hip joint. Because the lever arm multiplies the body weight so greatly, a tremendous torque is created. For no rotation to occur, this must be balanced by the pull of the abductor muscles ipsilateral to the pathologic hip. Because these abductors act through a relatively short lever arm, they must exert an enormous pull. As a result, the hip experiences a large joint force, and arthritis is worsened. Placing a cane in the contralateral hand gives it the longest lever arm and greatest mechanical advantage. This allows the person to easily offset the effect of the body weight, and thereby minimize joint forces. This is demonstrated clinically as contralateral cane usage for hip pathology is nearly uniformly seen.

Most authors also advocate contralateral cane usage for the treatment of unilateral lower limb problems below the hip, such as arthritis, acute fractures, and congenital deformities.[9,10,14] Here, the cane acts not to minimize forces in the frontal plane but rather as a load sharing strut to minimize vertical stress. The most compelling argument for contralateral usage is that this minimizes shifts in the body's center of gravity, which must lie between the cane and the weight bearing foot.[9] Contralateral cane usage ensures that the center of gravity remains medial to the weight-bearing foot and close to the midline of the body. In contrast, ipsilateral usage causes the center of gravity to swing lateral to the weight bearing foot, and then over to the opposite foot on the next step, and this is felt to require more energy. In clinical practice, however, contralateral use is not uniformly seen. Many patients frequently feel more comfortable utilizing a cane ipsilaterally and physicians generally allow them to use whichever side is more comfortable. This is especially true in young healthy patients for whom fatigue is generally not a problem when canes are required for short-term intervention of acute conditions. High strain rates have been implicated in the origin of both stress fracture and osteoarthrosis of the knee.[3,4] The treatment concept of stress fracture is based on a temporary reduction of the bone strain environment, allowing bone to carry out a local remodeling response and strengthen itself.[4] Similarly in knee osteoarthrosis, lowering of subchondral bone strains rates can allow for biological healing of subchondral microdamage and strengthen cartilage support. On the basis of this study, we conclude that a cane may play a beneficial role in the treatment of both tibial stress fractures and osteoarthrosis of the knee by lowering the tibial strain rates. Either ipsilateral or contralateral usage can achieve the same effect, and we recommend letting patient comfort be the deciding factor.

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