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A modular six-directional force sensor for prosthetic assessment: A technical note

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Abstract—A device designed to measure the forces and moments transmitted through prostheses of persons with lower limb amputation is presented. The sensing unit design is an advancement over previous prosthesis force measurement devices in that it is very thin (19 mm) and lightweight (527.5 g, including signal-conditioning instrumentation). The disk-shaped transducer fits between the socket and socket adapter of a standard modular prosthesis, measuring all six force and moment components at this location. Twelve strain gages were used, configured into six two-arm active Wheatstone bridge circuits. A 6×6 matrix was constructed from calibration data to relate the 6-component bridge-output vector to a 6-component force and moment vector. In a bench-test setting, the sensor was evaluated under typical load combinations encountered during the walking of a person with transtibial amputation (TTA) and shown to have errors less than 7.2% of the full-scale output for each direction. Data collected on a subject with TTA walking at different speeds are presented.

Key words: *force sensor, lower limb amputation, prosthetic alignment, prosthetic fitting, prosthetics.*

INTRODUCTION

A modular six-directional force sensor has application to prosthetic design and research in three principal ways:

1. *Componentry Design and Evaluation.* Designing commercial componentry (e.g., feet, ankles) is a challenge in prosthetics. Because data on forces and moments encountered during component-threatening (material failure and fatigue) activities such as playing tennis or basketball are minimal, engineers designing prosthetic componentry do not have information on the combinations of forces and moments their designs must tolerate. If over-designed, the componentry is inefficient. If under-designed, it is unsafe. A lightweight, robust, and modular six-directional force sensor would allow measurement of forces and moments during strenuous activities, providing data that engineers could then use in componentry design. Prosthesis loading data are also critically needed for committees that develop safety standards for prosthetics (1). Standards committees establish criteria to ensure that componentry is safe for the ranges of activities undertaken by persons with amputation.
2. *Prosthetic Finite Element Analysis.* Finite element (FE) models of the residual limb and prosthetic socket potentially provide insight into prosthesis design features that strongly influence the normal and shear stresses at the residual limb-socket interface. Interface stresses are of strong clinical relevance because of their role in skin breakdown. Several investigators have developed residual limb-socket FE models (e.g., 2–4), and their enhancement into useful prosthetic tools is an area of active research. However, FE models rely on accurate input data, data that typically include the forces and moments applied through the pylon to the bottom of

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the socket. Six-axis prosthetic force sensors provide these data, assisting in enhancement of FE models as useful tools in prosthetic design.

3. *Prosthetic-Fitting Tool.* Highly subjective practices are used in the assessment of prosthetic fit, partly because those methods have been effective, but also because limited quantitative tools are available in prosthetics. With increased fiscal constraints and demands on a prosthetist's time, devices that improve his or her efficiency and the quality of fit are of strong utility. A device worth pursuing is one that, based on just a few walking steps by a patient with amputation, indicates to a clinician the necessary alignment adjustments. The basis that such a device might be possible comes from gait studies on a limited number of subjects with amputation, suggesting that relationships exist between gait force (and moment) versus time curves and fit parameters (5–10). To pursue and investigate the feasibility of such a device, there is a need for a convenient, portable, inexpensive measurement tool to conduct the necessary extensive force and moment data collection at different misalignments on many subjects.

There are essentially three techniques available to measure forces and moments in a prosthesis. They are:

1. *Gait Analysis Laboratory: Force Plate and Video System.* A gait analysis laboratory typically consists of a ground reaction force plate imbedded in the floor and a video system for tracking markers on the subject. Forces and moments in limb segments are determined by transforming the ground reaction force to limb coordinates based on kinematic data. A gait analysis laboratory is not a highly viable technology for the three purposes listed above, since it is expensive, not portable, and requires much postprocessing to convert the raw data to segment and joint force and moment data. In addition, unless more than two diagonally positioned force plates are used, data for only one step in a sequence of steps can be monitored, thus limiting the percentage of steps per session that can be collected.
2. *Instrumented Pylon.* Instrumented pylons are another approach to measurement of prosthesis forces and moments and have been developed by several researchers (10–13). A design similar to the unit described by Berme is currently being marketed by a prosthetics company on a design-to-customer basis. Instrumented pylons are more portable than force plates and not as expensive; however, they are difficult to use on large populations of subjects with amputation, because they are of a fixed length and typically replace the pylon section. Berme's design was 7 cm in length (11). Replacing the pylon section is a problem because the instrumented pylon length is likely to be different from the pylon section length it replaces. Thus, each prosthesis must be custom-adjusted at the socket or foot to achieve the proper leg length. For persons with long transtibial amputations (TTA), the instrumented pylon will make the leg too long and, therefore, cannot be used.
3. *Load Cell.* Commercial load cells that measure all six force and moment components exist (e.g., AMTI Corporation, Newton, MA; Astek Incorporated, New Burlington, WI; Sensotec, Columbus, OH; Assurance Technologies, Inc. [ATI], Garner, NC). However, most of these units and their signal processing instrumentation are large and relatively heavy, and are typically used in robotics and other industrial applications where a large mass is often not as much of a concern as in prosthetics applications. There does not appear to be a sufficiently lightweight commercial load cell possessing the required sensitivity in all channels to the ranges of forces and moments seen in human activities. A load cell custom-designed for prosthetic applications is needed.

In this research, a custom-designed sensor and portable signal-processing system were developed for measurement of the six force and moment components in a prosthesis during ambulation. The small and lightweight device, with on-board signal conditioning, overcomes many of the limitations of previous measurement systems (gait analysis laboratories, instrumented pylons, and industrial load cells). The described instrumentation fulfills an important need in prosthetics as an instrument for componentry design and evaluation, residual limb-prosthetic socket FE modeling, and also potentially as a prosthetic-fitting tool. The design of the sensor, its performance, and data from a pilot study are described.

METHODS

The instrumentation consisted of a titanium sensor unit and signal conditioners.

Sensor Unit

The sensing element was a titanium unit machined to a designed geometry and instrumented with foil strain gages (EA-13-062AQ-350, Measurements Group, Inc., Raleigh, NC) and consisted of three instrumented beams that attached an inner ring to an outer ring (**Figure 1**). The lower face of the inner ring connected to the prosthetic pylon, while the top face of the outer ring connected to a plate that attached to the prosthetic socket. Thus, all loads between the pylon and socket were transferred through the three beams. The practice of transmitting loads through instrumented beams is typical of commercial load cells. Each beam was instrumented with strain gages to measure bending in two orthogonal directions, a configuration from which all three forces and three moments (in an orthogonal reference frame) applied to the inner ring relative to the outer ring could be determined. Four strain gages were mounted on each beam, one on each side, and oriented to measure strains parallel to the long axis of the beam. Strain gages on opposite sides of a beam were configured in two-arm active Wheatstone bridges; thus, there were two active bridges per beam. Each bridge measured bending in the beam in a plane perpendicular to the gage surfaces. Completion resistors for all bridges (S2-350-01, Measurements Group, Inc.), two per bridge, were epoxied to the inner surface of the outer ring.

Strains measured with the six pairs of gages allowed all six force and moment components to be determined. Three pairs of strain gages responded predominantly to three of the force and moment components, while the other three pairs responded predominantly to the remaining three force and moment components. The bridges from gages on the tops and bottoms of the beams were sensitive to mainly bending about the x and y axes and force along z (**Figure 1**). Since there were three different measurement channels and three dominant loads (*Mx, My,* and *Fz*), the three measurements were sufficient to determine the three loads. A similar argument held for the three bridges configured with gages on the sides of the beams. *Fx, Fy,* and *Mz* could be determined from those three channels.

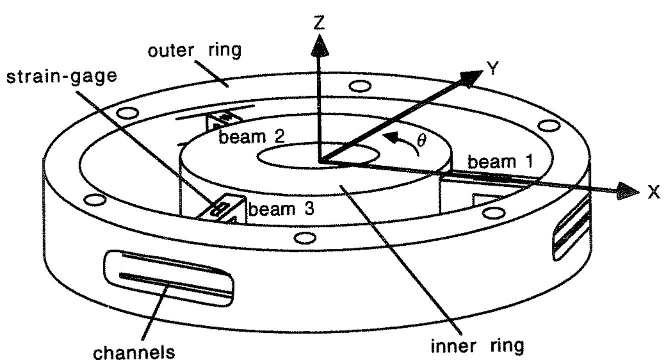

Figure 1.
Schematic of the titanium unit. The lower face of the inner ring connected to the prosthetic pylon, while the top face of the outer ring connected to a plate attached to the prosthetic socket. All loads from the pylon to the socket were transferred through the three instrumented beams.

The design of the beam, ring, and channel dimensions was challenging because of the force and moment combinations encountered by persons with amputation during ambulation. Typical forces of persons with TTA during walking, measured in the cross section of the socket axis (resultant of *Fx* and *Fy*), were 150 N to 275 N (as measured with an instrumented pylon), those along the socket axis (*Fz*) were 750 N to 950 N, and the bending moment about the bottom of the socket (resultant of *Mx* and *My*) was 55 N-m to 80 N-m (14). Torsional moments were typically 2 N-m to 5.5 N-m. These magnitudes in combination with a requirement that the sensor be no larger than 10 cm in diameter, so as not to interfere with the contralateral limb, resulted in moments *Mx* and *My* dominating the loading in the beams and potentially producing very high strains in the unit compared with the other loads (*Fx, Fy, Fz,* and *Mz*). To increase sensitivities to forces *Fx* and *Fy,* channels perpendicular to the beams were cut out of the outer ring (**Figure 1**). The cuts allowed a beam to translate relative to the outer ring when a force was applied along its axis, increasing the strain in the other two beams; thus increasing the signal when the sensor was loaded in *Fx* or *Fy.* The concept of cutting channels in the outer ring has been described previously by ATI; however, semi-conductor rather than foil strain gages were used in their load cells, improving the sensitivity but requiring heavy signal-processing instrumentation that was of unacceptable mass for this application.

Beam and ring dimensions were selected so that adequate sensitivities were induced without over-stressing the beams. The outer ring dimension was 10 cm, the designed maximal allowable diameter, so that the sensor did not protrude excessively far from the prosthesis and contact the contralateral limb. The thickness of the outer ring was 19 mm and the top connector plate that attached the sensor to the socket was 6.7 mm in thickness. The masses of the titanium unit and plate were 317 grams and 139 g, respectively.

Signal Conditioners

A small and lightweight (71.5 g) signal-conditioning unit, housed in a plastic casing, was positioned next to the sensor. We used two three-channel modular signal conditioners, described in detail elsewhere (15). Each unit allowed for bridge-balancing, amplification, and low-pass filtering for three channels. Two AD620 instrumentation amplifiers (Analog Devices, Norwood, MA) in series were used for each channel, allowing gains from 1 to 10,000 to be achieved. In this application, a gain of 660 was used for bridges with strain gages bonded to the tops and bottoms of the beams, while a gain of 5539 was used for bridges with strain gages bonded to the sides of the beams. A high-pass filter set at a 3 dB point of 100 Hz (the phase delay was linear in the passband) was used at the output of the second-stage amplifier. The unit was connected to the external surface of the outer ring (**Figure 2**). Positioning the signal conditioners close to the gages reduced noise levels compared with positioning the conditioners in a belt pack.

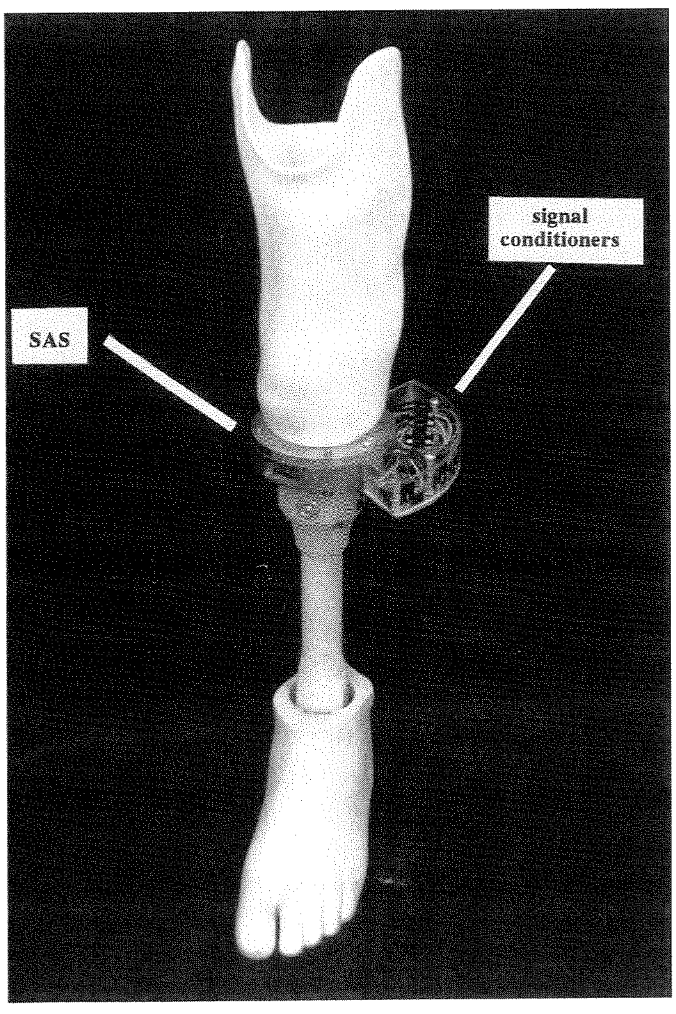

Figure 2.
The six-axis sensor (SAS) positioned between the socket and socket adapter sections of a modular prosthesis. The signal conditioners were located in the plastic box on the lateral side.

Performance

Calibration was conducted using a custom-designed device as illustrated in **Figure 3.** The device allowed different force and moment combinations to be applied to the sensor. By positioning the carriage unit at different positions along the rail, adjusting the tilt table angle α, rotational angle θ (θ is shown in **Figure 1**), and torsional moment arm length while hanging masses on the weight support, different combinations of forces and moments could be applied to the sensor. Actual forces and moments were calculated using the known moment arm lengths and applied masses.

For each force and moment direction, the magnitudes of the loads applied during calibration were in the ranges of those measured on subjects with amputation (14). Loads were applied in at least four increments (typically eight increments were used) to the maximal load. Measurements during unloading were also taken to evaluate hysteresis. A total of 523 different loading combinations were applied. It is important to note that the approach ensured that calibration performance was not dependent on the x-y orientation (θ angle) of the sen-

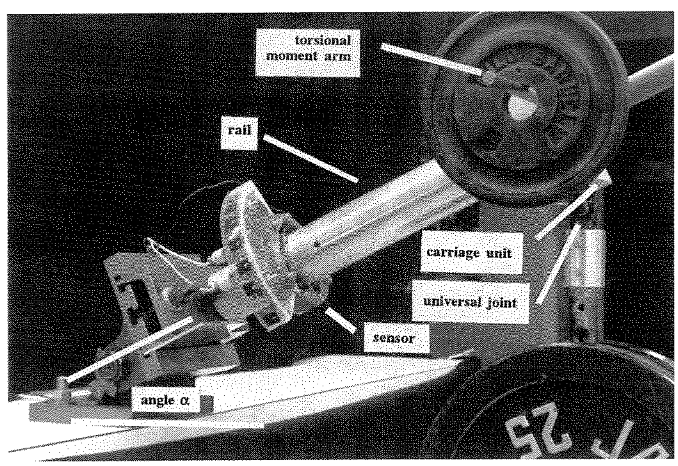

Figure 3.
Jig for calibration and bench testing. The carriage unit supported the weights and could be moved along the rail to change the length of the moment arm. A universal joint was used to ensure that no torsion (Mz) was applied by the lower weight. Only the weight in the upper right corner applied torsion. Different θ and α angles and applied masses were used to achieve different combinations of applied loads.

sor with respect to the direction of walking. Thus, the user could position the sensor in any θ angle in the prosthetic limb.

No hysteresis was evident from the data. This is expected due to the very linear performance of foil strain gages. Thus, in further analysis only the unloading (and not the loading) data were used.

From the data, a 6×6 matrix, **K**, was generated to relate the measured voltages to the applied loads:

$$\mathbf{F} = \mathbf{K}\mathbf{V}$$

where **F** was a 6×1 column matrix of the measured forces and moments (in three orthogonal directions) and **V** was a 6×1 column matrix of the voltages (**Figure 4**). **K** was a least-squares best fit to the calibration data. It was generated as follows: An $\mathbf{F}_{calib}$ matrix was created, made up of the column vectors from the calibration data. Each six-element column vector within $\mathbf{F}_{calib}$ represented an applied load from the calibration testing; thus, the $\mathbf{F}_{calib}$ matrix was of dimension 6×523. Similarly, a $\mathbf{V}_{calib}$ matrix (of dimension 6×523) was formulated where each six-element column vector in that matrix represented the output voltages corresponding to the applied force six-element column vector in $\mathbf{F}_{calib}$. Thus, **K** was determined:

$$\mathbf{K} = \mathbf{F}_{calib}\,\mathbf{V}_{calib}^{T}\,(\mathbf{V}_{calib}\,\mathbf{V}_{calib}^{T})^{-1}$$

Calibration tests produced the **K** matrix shown in **Figure 4.** Terms in small text italics are "crosstalk terms." As expected, they are of small magnitude relative to other terms in their row. Thus, the appropriate channels dominated the outputs for the given force or moment direction. Matrix terms in which a beam was aligned with a load axis were near to zero for that beam, (i.e., row 1 column 4: *Fx* with beam 1 as shown in **Figure 1**), and row 4 column 3: Mx about the beam 1 axis). The signs of the calibration terms (non-crosstalk terms) were appropriate relative to each other. For example, a force in *Fy* caused beam 3 to bend in a direction opposite to that produced by force *Fx*, and the terms (row 2 first term and row 1 first term) were opposite in sign. For typical maximal stance phase walking loads, the output voltages were similar for the three channels from gages mounted on the tops and bottoms of the beams. Output voltages were also similar for the three channels from gages mounted on the sides of the beams (**Table 1**). With the *y*-axis of the sensor aligned with the walking direction, voltages for the active channels for *Fz, Mx,* and *My* ranged from 0.29 V to 1.10 V; for *Fx, Fy,* and *Mz,* they ranged from 0.12 V to 0.22 V.

$$\begin{bmatrix} \mathrm{Fx} \\ \mathrm{Fy} \\ \mathrm{Fz} \\ \mathrm{Mx} \\ \mathrm{My} \\ \mathrm{Mz} \end{bmatrix} = \begin{bmatrix} 331 & \mathit{2.72} & \mathit{2.97} & 50.3 & \mathit{0.19} & -302 \\ -151 & \mathit{21.1} & \mathit{15.1} & -372 & \mathit{-16.1} & -189 \\ \mathit{-8.79} & -733 & -727 & \mathit{1.27} & 744 & \mathit{-11.7} \\ \mathit{-1.18} & 32.3 & -0.18 & \mathit{0.05} & 35.2 & \mathit{-0.42} \\ \mathit{-0.24} & -20.6 & 37.7 & \mathit{0.51} & 21.0 & \mathit{-0.36} \\ 8.00 & \mathit{-0.61} & \mathit{-0.54} & -3.96 & \mathit{0.43} & 6.18 \end{bmatrix} \begin{bmatrix} \text{V1 (bm 3, sides)} \\ \text{V2 (bm 3, top/bot)} \\ \text{V3 (bm 1, top/bot)} \\ \text{V4 (bm 1, sides)} \\ \text{V5 (bm 2, top/bot)} \\ \text{V6 (bm 2, sides)} \end{bmatrix}$$

Figure 4.
Calibration matrix for the sensor. Force vector (**F**) = calibration matrix (**K**) × voltage vector (**V**). The gage locations are in parentheses after each voltage component using the coordinate system convention shown in **Figure 1** (bm = beam, top = top of beam, bot = bottom of beam, sides = sides of beam).

Table 1.
Absolute values of voltages in each beam with typical walking force and moment levels.

Gage Locations	Direction	Force Level*	Voltages Beam 1	Beam 2	Beam 3
Tops and Bottoms	Fz	950 N	0.29	0.29	0.29
	Mx	80 N-m	—	1.10	1.10
	My	80 N-m	1.35	0.59	0.59
Sides	Fx	275 N	—	0.25	0.25
	Fy	275 N	0.27	0.12	0.12
	Mz	5.5 N-m	0.22	0.22	0.22

*Maximum forces and moments during walking: from a previous study (14). The voltages were calculated using a finite element model of the sensor, assuming loads were applied in one direction at a time. The outputs for Mx-Beam 1 and Fx-Beam 1 were zero because equal strains were induced in gage pairs for those load directions.

Quality of the matrix was tested by applying loads to the sensor in directions other than those used during calibration but still within the design ranges described above. Results (**Figures 5** and **6**) showed that for all six components for all of the tests, root-mean-square (RMS) errors were less than 7.2 percent of the full-scale values (**Table 2**). Bending and torsional channels showed mini-

Table 2.
Root-mean-square errors/full-scale output values for each channel.

Direction		Root-mean-square error full-scale output (%)
Force	Fx	6.6
	Fy	7.1
	Fz	6.2
Moment	Mx	2.0
	My	1.4
	Mz	5.5

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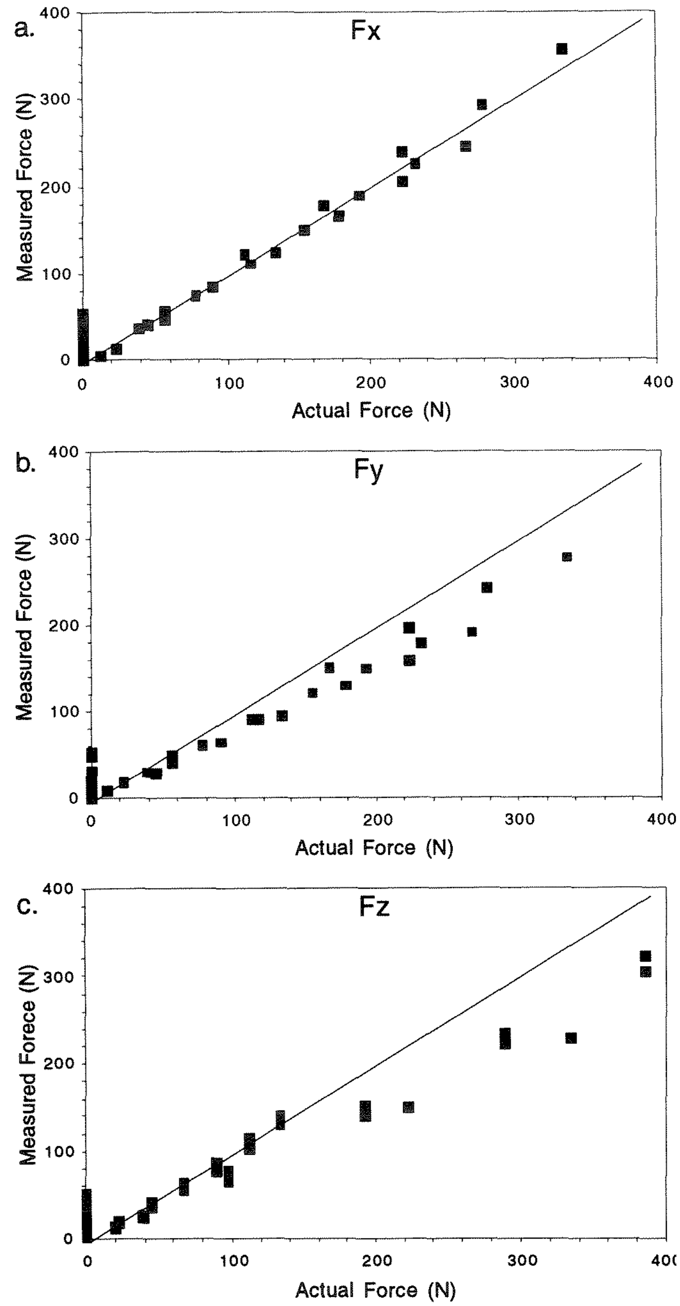

Figure 5.
Force test data in newtons. Actual forces (horizontal) vs. measured forces (vertical) are shown for each of the three force directions. The sensor was tested under simultaneous force and moment loading combinations different from those used for calibration. Each plot contains data from several tests and data from the same tests are shown in different plots.

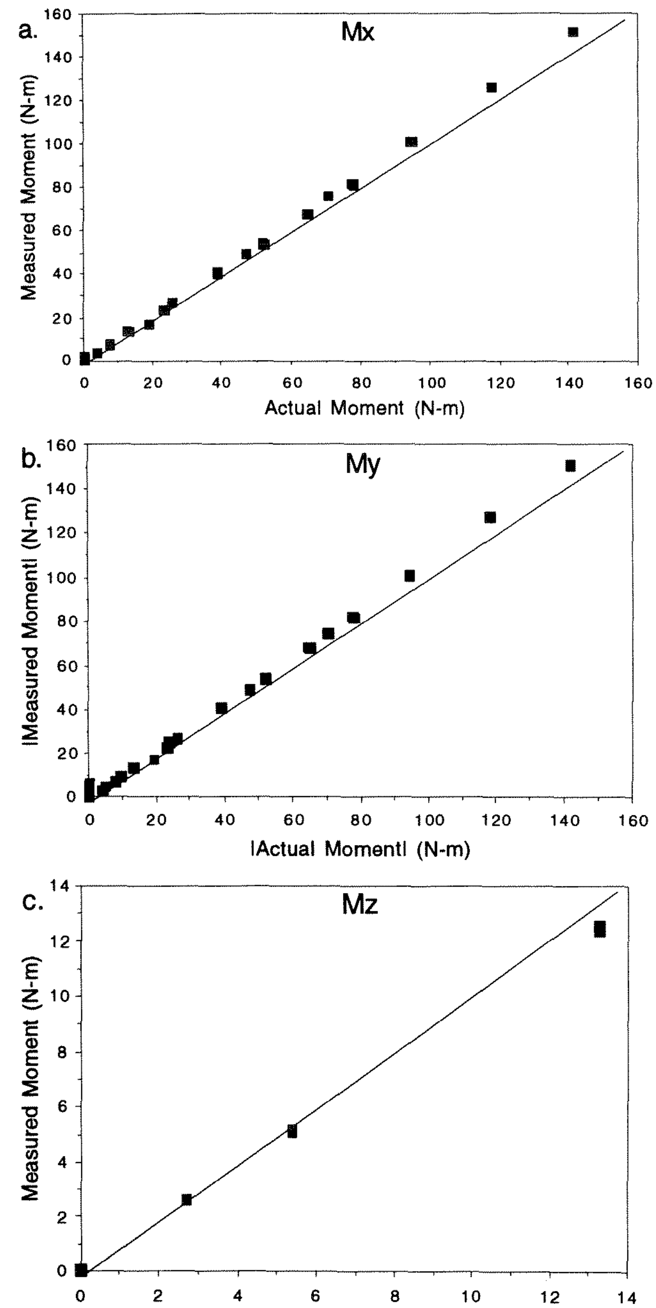

Figure 6.
Moment test data in newton-meters. Actual moments (horizontal) vs. measured moments (vertical) are shown for each of the three moment directions. The sensor was tested under simultaneous force and moment loading combinations different from those used for calibration. Each plot contains data from several tests and data from the same tests are shown in different plots.

mal errors. Axial force tended to be underestimated; shear forces were not as accurate as the other load directions.

Noise evaluation tests were also conducted. Noise in the signal under constant static loading conditions averaged 0.008 V peak-to-peak.

Standing and Walking Studies

Standing and walking studies were conducted on a subject with a left TTA to test the system in a clinical setting. With the sensor inserted between the socket and pylon of a Seattle™ system (M+IND, Seattle, WA) prosthesis, the 28-year-old male (mass 68 kg; weight 666 N) stood with equal weightbearing. He then walked the length of a vinyl floor hallway at approximately 69 steps/min and then at approximately 44 steps/min, with walking speed controlled by metronome.

Compared to an instrumented pylon, the new sensor was relatively easy to install into the prosthesis. The nylon pylon was reduced in length by the thickness of the sensor, and the sensor was inserted between the socket and socket (alignment) adapter. A cable extended to a stationary data acquisition system used for data storage (National Instruments NB-MIO-16 A/D board, Austin, TX; Centris 650 Apple Computer, Cupertino, CA).

Measurements with the subject standing with approximately equal weightbearing showed a measured axial force (Fz) of 282 N, which is approximately half (42.3 percent) of the subject's body weight. Analysis of walking data demonstrated that waveforms were of different shape at slow rates compared with fast rates. For example, axial force (Fz) showed a first peak magnitude:second peak magnitude ratio of 1.01 ($\pm$0.03) at 69 steps/min and 0.91 ($\pm$0.01) at 44 steps/min. At least 30 steps were collected at each speed. At the faster speed, the subject hit the ground more firmly at heel contact, inducing a greater axial force. At the higher speeds, bending in the sagittal plane (Mx) showed slightly greater absolute magnitudes during stance phase and a greater amplitude transient response during early swing phase (**Figure 7**). The higher magnitude transient response was probably due to dynamic effects from the increased push-off rate and force.

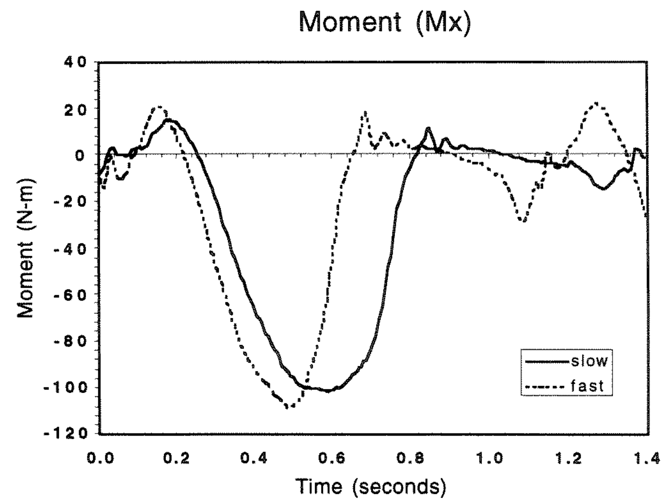

Figure 7.
Results from a data collection session on a subject with left-leg amputation. Bending in the sagittal plane (Mx) is compared for a slow (44 steps/min) and fast (69 steps/min) walking rate. The sensor was positioned in the prosthesis such that x was pointing laterally and y was pointing posteriorly.

DISCUSSION

The six-axis force sensor described in this technical note is a new design for prosthetic force and moment measurement. For the purpose of a small and lightweight force and moment measurement device with on-board signal conditioning, the design overcomes many of the limitations of previous measurement systems (gait analysis laboratories, instrumented pylons, and industrial load cells). The unit is thin and lightweight, and fits between the socket and pylon sections of a modular prosthesis without requiring removal of the entire pylon. Instead, the existing pylon is simply shortened by the thickness of the sensor, a relatively straightforward procedure. Data from consecutive steps can be collected. Signal conditioners are small and lightweight, contributing to the portability of the device.

The device has immediate use for the purposes of componentry design and residual limb-prosthetic socket FE modeling. Prosthesis force and moment data are strongly needed for those two applications. Use of the device as a prosthetic-fitting tool (e.g., dynamic alignment tool) is a longer term goal, since it must first be clearly established that misalignment is related to prosthetic force and moment data. The preliminary study presented here, prosthesis force data for different walking speeds, illustrates the capability of the device for evaluation of prosthesis load/gait feature relationships. Extensive data collection and subsequent identification and verification of prosthesis load/gait feature relationships will need to be carried out before the device can be considered a useful prosthetic-fitting tool.

In calibration testing, the sensor was shown to perform with RMS errors of less than 7.2 percent of the full-scale output. As shown in the evaluation data (**Figures 5** and **6**), errors were larger for the shear directions (*Fx* and *Fy*) than other directions, and axial force (*Fz*) tended to be underestimated. Slight gage misalignment and crosstalk with other high-strain producing load directions may have caused these errors. Modifications to the beam and back-support piece dimensions may help to reduce these limitations.

This sensor was designed and tested for loading ranges encountered during walking. The calibration jig and possibly the sensor design would need to be modified for loads of substantially higher magnitude, such as those encountered during extremely strenuous activities.

Future enhancements of the device include optimization of the geometry for reduced crosstalk with minimal mass but adequate strength. A portable data acquisition and processing system would also enhance the utility and versatility of the device.

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