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Transducers for the determination of the pressure and shear stress distribution at the stump–socket interface of trans-tibial amputees

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Abstract: Recent developments in prosthetic socket design have created renewed interest in monitoring the stress distribution at the socket–residual limb interface. Although a few devices for measuring pressure can be found in the literature, none are capable of measuring reliably in areas of high curvature, such as the important area at the patellar tendon bar. Furthermore, few devices can record shear stress, thought to be critical in causing tissue damage. In order to address these issues two new transducers have been designed and evaluated. One design allows the simultaneous recording of the normal and shear stresses at various points of the socket walls, while the other is capable of measuring the three components of the force applied on the patellar tendon. The latter design incorporates a feature that permits displacement of the patellar tendon bar, in order to study the effect of various amounts of indentation of the tendon on the stress distribution around the residual limb. Both transducers were calibrated using dead weights and special jigs to ensure accurate loading conditions. Under laboratory bench conditions the normal–shear force transducer showed: 2.03 per cent full scale output (FSO) hysteresis error for shear stress direction, 1.65 per cent FSO for normal direction; 99.56 per cent FSO overall accuracy for shear direction, and within 99.64 per cent FSO for normal direction; and for the patellar tendon transducer 1.53 per cent FSO hysteresis error for shear direction, 1.85 per cent FSO for shear stress direction; 99.65 per cent FSO overall accuracy for shear direction, and 99.58 per cent FSO for normal direction. During an amputee walking trial the transducers showed 92 per cent to 97 per cent repeatability. The two new transducers were used in conjunction with two other types of transducers, previously designed at the University of Strathclyde, in a series of tests on ten trans-tibial amputees. Sample results for walking activities and a summary of maximum stresses recorded are presented.

Keywords: prosthetic socket, trans-tibial socket, transducer, shear stresses, stump–socket interface

1 INTRODUCTION

In the UK, of the 5298 new referrals of lower limb amputees to limb centres in the year 2002, 52 per cent were at the trans-tibial level [**1**]. The next most common level of amputation was trans-femoral, accounting for 39 per cent. The overall total showed a 6 per cent increase in amputee referrals from the previous year, thus indicating an increasing trend in the number of amputations performed. This trend is reflected globally. Currently it is estimated that 25 per cent of prosthetic sockets manufactured in the UK are either rejected as ill fitting or are extensively modified subsequent to manufacture in order to achieve an acceptable 'fit', indicating a lack of consistency of the procedures in common practise.

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With the exception of prostheses fitted using the osseointegration techniques described by Branemark *et al.* [**2**], most lower limb prostheses must transmit the ground reaction forces through the soft tissues of the residual stump to the underlying skeleton. Prior to amputation these soft tissues would have been unaccustomed to being subjected to such loads, and they are therefore vulnerable to damage occurring from the use of the prosthesis. It is the task of the prosthetist to ensure that a prosthesis is designed in such a way as to transmit the loads associated with locomotion minimizing, if not obviating, the risk of damage to the tissues.

The understanding of the complex interplay of normal and shear stresses between the stump and the socket is seen as fundamental to developing a biomechanical model of the interface and therefore a way in which improvements in socket design can be acheived. Meaningful interface pressure measurements require a proper measurement technique, the use of accurate and reliable transducers, appropriate placement of the transducer at the stump–socket interface, and suitable data acquisition and analysis capabilities. An ideal system should be able to monitor both normal and shear stresses continually without imposing significant interference to the original interface conditions.

Many investigators have studied pressure at the stump–socket interface [**3–14**] as reviewed by Mak *et al.* [**15**], and several methods for measuring the pressure distribution have been developed. In comparison, few studies have been undertaken to investigate the shear stresses acting at the interface [**12**, **13**].

Methods of obtaining normal stress (pressure) at the stump–socket interface have included use of strain-gauged diaphragm type transducers, pneumatic devices, fluid-filled sensors, and strain-gauged cantilevers or beams.

Appoldt *et al.* [**4**] developed a strain-gauged beam transducer 11 mm in diameter and 27 mm in length, which could measure normal and shear stress. Sanders and co-workers published a series of articles [**12**, **16–19**] detailing the development of a triaxial transducer for interface stress measurement in trans-tibial sockets. Their transducer was 6.35 mm in diameter with a reported hysteresis error of 3.01 per cent full-scale output (FSO) for the normal direction and 0.31 per cent FSO for the shear direction. It was necessary for the subject to carry a back-pack box (21 × 18 × 12 cm, 3.2 kg) accommodating the data collection facility. No transducers were sited on the medial aspect of the socket in order to prevent interference with the normal walking pattern by contact with the other limb.

Williams *et al.* [**13**], by modifying a design of Tappin *et al.* [**20**], developed a triaxial force transducer 15.9 mm in diameter and 4.9 mm thick, using a strain-gauged diaphragm to detect normal force and using the Hall effect for measurement of shear force. Hysteresis error was reported as 2 per cent FSO for the normal direction and 8 per cent for the shear direction.

The present paper describes the design and development of transducers which are capable of measuring interface pressures and shear stresses at the prosthetic stump–socket interface. A description is also given of a specially built transducer which can apply and measure the three orthogonal components of the force on the patellar tendon for the purpose of studying various socket-fitting procedures currently used by prosthetists. Sample results obtained from a series of tests undertaken on ten trans-tibial subjects are presented.

2 DESIGN OF TRANSDUCERS

Four custom-made devices were used to determine the forces and stresses at the socket–stump interface. These were:

(a) patellar tendon (PT) transducer;
(b) normal pressure/shear stress transducers (BESTs);
(c) Entran® based device;
(d) electrohydraulic pressure sensor.

2.1 The PT transducer

This device can be attached on the PT bar of a trans-tibial socket for the purpose of measuring the normal force and the two components, i.e. longitudinal and tangential of the shear force acting on the tendon (Figs 1 and 2). The transducer is capable of displacing the PT bar radially towards or away from the tendon by ± 10 mm from the neutral position, thus simulating the variable amount of indentation applied by the

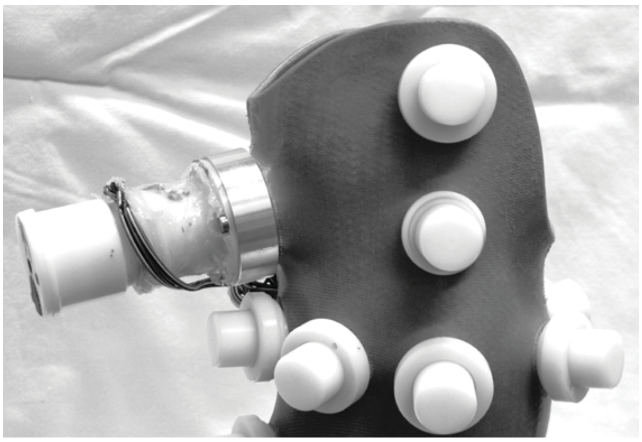


Fig. 1 The patellar tendon transducer fitted to a patellar-tendon-bearing (PTB) socket

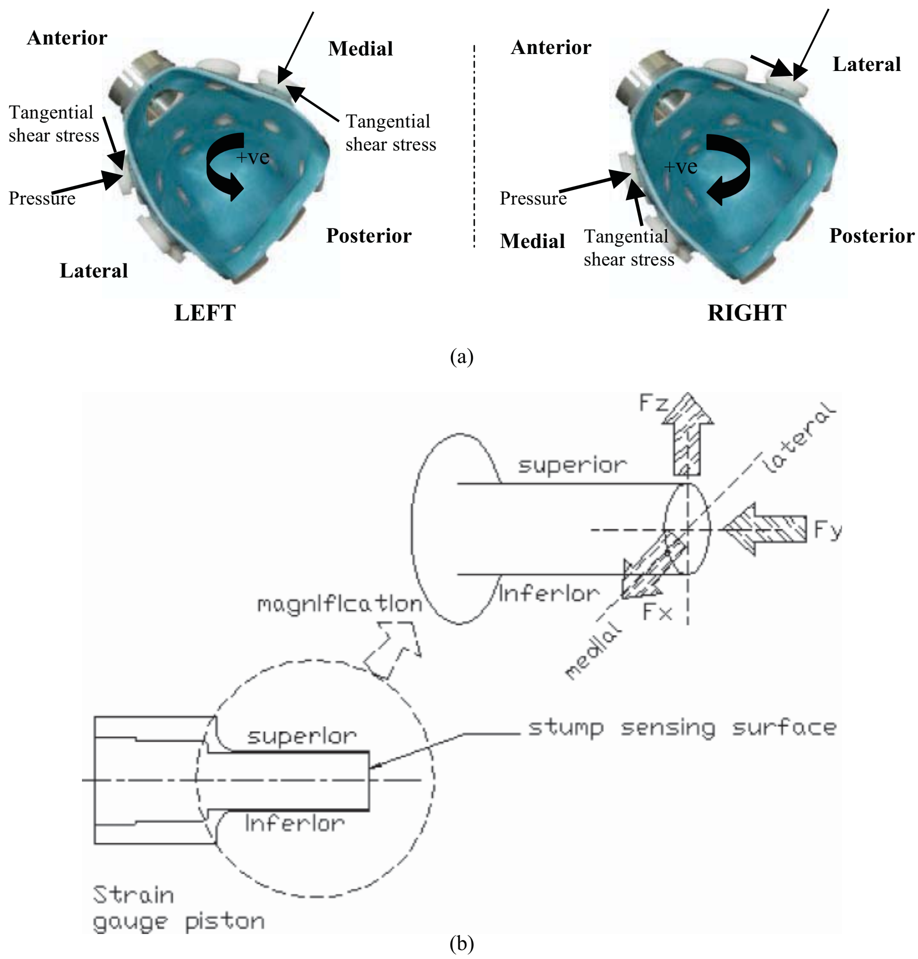


Fig. 2 (a) Sign conventions (by socket on stump): positive directions of forces as shown; (b) Transducer piston showing sign conventions adopted for the three forces measured at the interface F_y, normal force, F_x, horizontal shear force (tangential), and F_z, vertical shear force (longitudinal): positive directions of forces as shown

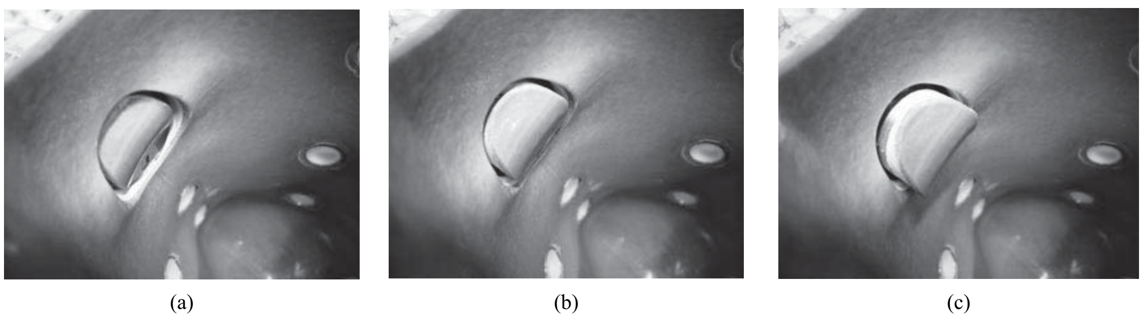
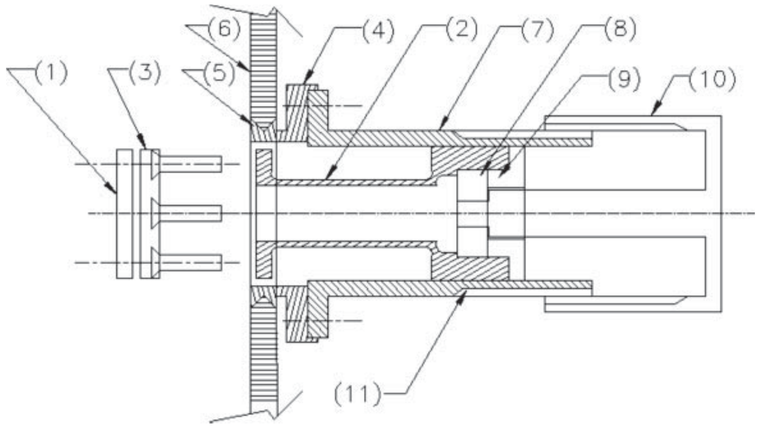
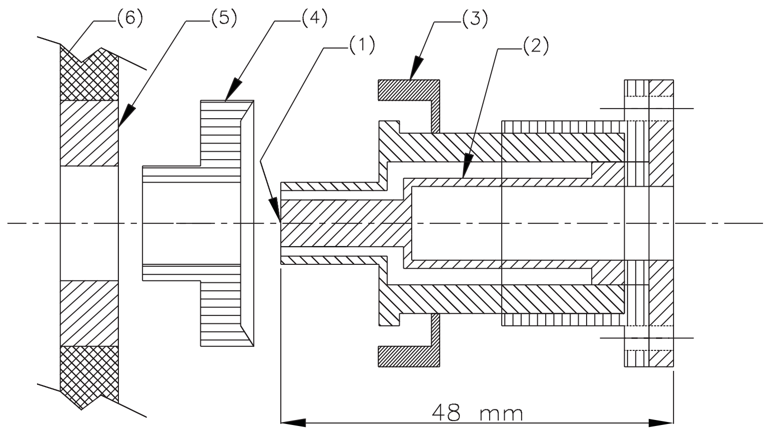


Fig. 3 Patellar tendon bar cut-out: (a) cut-out at -2 mm from neutral position; (b) cut-out at neutral position; and (c) cut-out at $+4$ mm from neutral position

prosthetist during the casting and rectification techniques. In order to achieve this movement, a section of socket containing the PT bar is cut out, as shown in Fig. 3. This was done by cutting a 25 mm diameter hole from the socket, using a jig designed to mate with a metal insert attached to the anterior aspect of the socket. The piece of socket removed formed the 'cut-out' or PT sensing surface and was bonded to the PT holder face using Akemi putty (Otto Bock Healthcare GmbH, Duderstadt, Germany) (Fig. 4(a)). This diameter of cut-out was chosen to achieve a representative area of the PT bar yet minimize dis-



(a)



(b)

Fig. 4 Schematic diagram of the custom designed transducers. (a) PT transducer (note: (1) 25.0 mm diameter PT bar sensing surface, (2) strain-gauged piston, (3) patellar tendon bar holder, (4) socket mount, (5) metal insert, (6) laminated socket (7) housing, (8) ball bearing, (9) piston push rod, (10) displacement control knob, (11) screw thread). (b) Pressure and shear transducer (BEST), exploded assembly drawing (note: (1) 5.6 mm diameter sensing surface, (2) strain-gauged piston, (3) lock nut, (4) socket mount, (5) metal insert, (6) laminated socket)

turbance to the socket contours. There was a radial clearance of 1.5 mm between the patellar bar sensing surface and the socket wall.

Attaching the assembled PT transducer, with the patellar sensing surface, to the socket via the metal insert ensured that the sensing surface was flush with the inner socket wall when the transducer was in the neutral position.

Rotational motion of the control knob was converted to linear translation of the strain gauge piston by means of a screw thread and single row radial ball bearing. By using a thread with a 1 mm pitch, each full turn of the control knob corresponded to 1 mm of linear translation. The components of the PT transducer were made from aluminium alloy and acetal bar to minimize the weight of the device, resulting in a mass of 102 g for the complete PT transducer. A schematic diagram of the PT transducer and its components is shown in Fig. 4(a).

The PT bar transducer comprised three full Wheatstone strain gauge bridges. The gauges were attached to the piston using a heat-curing adhesive (M-Bond 610, Micro-Measurements, Raleigh, NC) on a 12 mm diameter and 1 mm thickness of cross-section aluminium alloy cantilever beam. The gauges were supplied by Micro-Measurements with a gauge factor, $k = 2.0$ and resistance of $R = 120\ \Omega$ for the axial load measuring gauges (type: EA-06-062TT-120). For the shear gauges, the corresponding values were $k = 2.0$ and resistance of $R = 120\ \Omega$ (type: EA-12-062TH-120). Gauging was arranged in two bridge configuration as shown in Fig. 5: one bridge for the measurement of the normal force and one for the measurement of each of the shear forces, i.e. there is a bridge for tangential shear force F_x, for normal force F_y, and longitudinal shear force F_z. The normal force (F_y) was obtained using two 90° rosettes in each arm of the full bridge equi-spaced around the periphery of

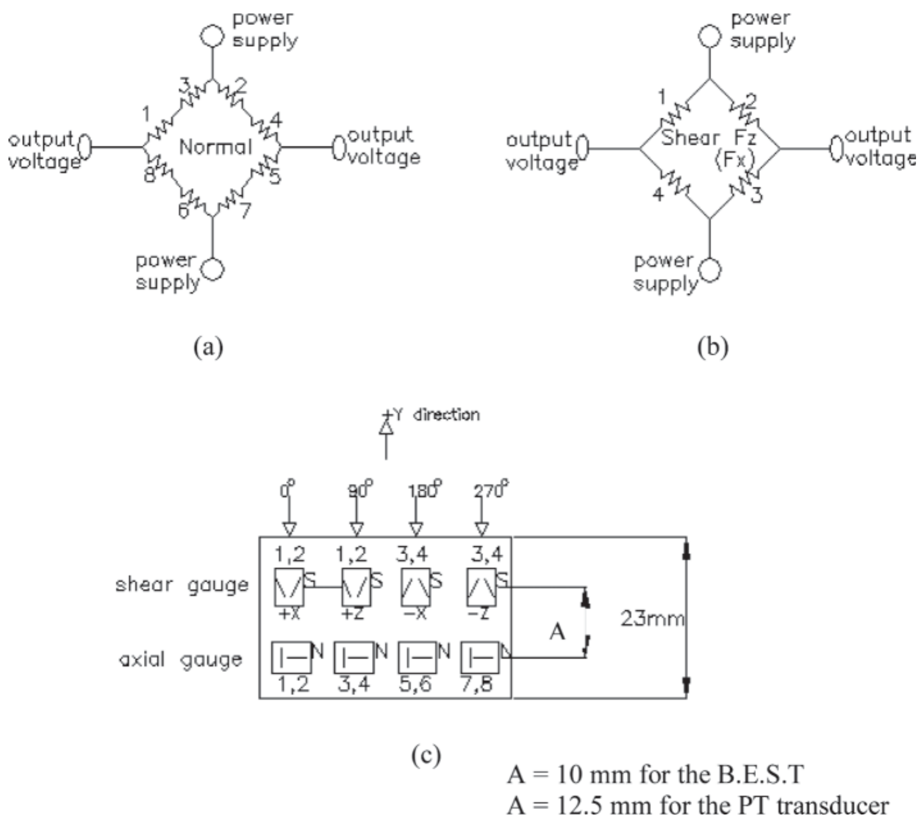


Fig. 5 Strain gauge orientation for BEST and PT transducer: (a) bridge for normal force measurement in the y direction; (b) bridge for shear force measurement in the z and x directions; (c) development of transducer piston cylindrical surface showing strain gauge positions

the piston, thus monitoring the direct compressive stress produced by the normal force, F_y. The shear forces, F_x and F_z were measured by rosette gauges aligned at 45° to the main axis of the piston. The distance between gauge centres was 12.5 mm.

2.2 The normal/shear or bioengineering unit shear transducer

The bioengineering unit shear transducer (BEST) shown in Fig. 6 is a strain-gauged type device capable of measuring the normal force (F_y) as well as two shear forces (F_x tangential shear; F_z longitudinal shear) simultaneously at any location of interest in a socket. The design of these transducers uses the same strain gauge types as the PT transducer, the only difference (beside the size), being that the BEST piston cannot be displaced inwards and outwards. The strain gauge arrangement is the same as that in the PT transducer. The size of the sensing surface is 5.60 mm in diameter and the mass of the assembly 89 g. The BEST was manufactured using three full Wheatstone bridge strain gauges, which were attached to the piston using the same heat-curing adhesive as the PT transducer on an aluminium alloy cantilever beam 11 mm in diameter with a thickness of 1 mm in cross-section. The distance between gauge centres was 10 mm (see

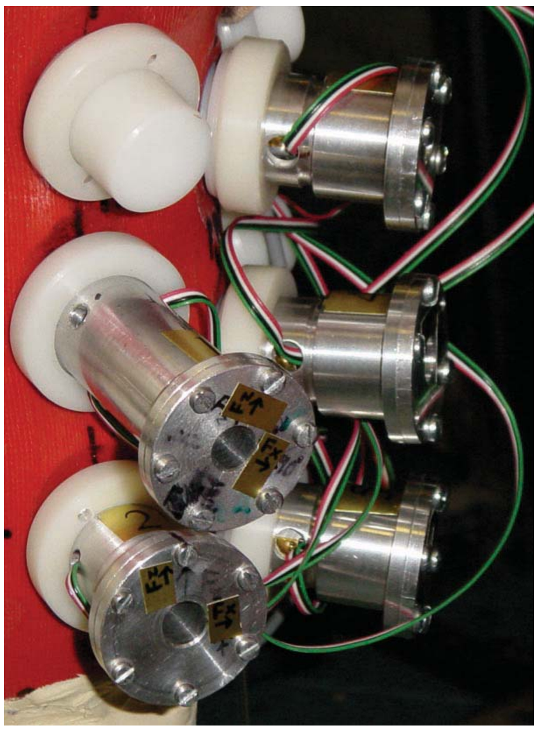


Fig. 6 Pressure/shear transducers (BESTs) mounted on socket

Fig. 5). All gauges were covered with a thin layer of silicone coating for mechanical protection and waterproofing, and to reduce noise. Details of the transducers are given in Fig. 4(b).

The PT transducer and BEST were machined from aluminium alloy (BS 1474: 1987 and tempered, T6). This material is commonly known as 'HE30'. The material has an ultimate tensile strength of 310 Mpa (minimum) and has 0.2 per cent proof stress of 260 Mpa (minimum). The mechanical properties of an aluminium alloy are: modulus of elasticity, $E = 68.9$ GPa, Poisson ratio $\nu = 0.26$, and shear modulus, $G = 27.3$ GPa.

2.3 Entran based device

This transducer is a piston in a housing device incorporating an Entran type ELFM-B1-5L load cell (Entran International, New Jersey, USA). This device has been described by Lee *et al* [**21**]. It is capable of measuring the normal interface pressure between the stump and the socket at any location of interest.

2.4 The electrohydraulic sensor

It is difficult to measure the distal end interface pressures owing to the shape of the stump end and it is impossible to use the BEST owing to the location and dimensional limitations. The electrohydraulic sensor consists of a sensor bag, a tube, and a commercial pressure sensor which can withstand up to 204 kPa [**22**]. The sensor bag, the tube, and the pressure sensor are filled with paraffin oil, care being taken to eliminate all air bubbles. Paraffin oil was chosen because of its viscosity and density, which are among the lowest values compared to other fluids, i.e. vegetable oils. The electrohydraulic sensor consists of a 28 mm diameter, 2 mm thick oil-filled polyvinyl chloride (PVC) bag, which is connected via a PVC tube to a strain-gauged diaphragm transducer. The tubing is 100 mm long, with an internal diameter of 2 mm and wall thickness of 0.5 mm (Fig. 7).

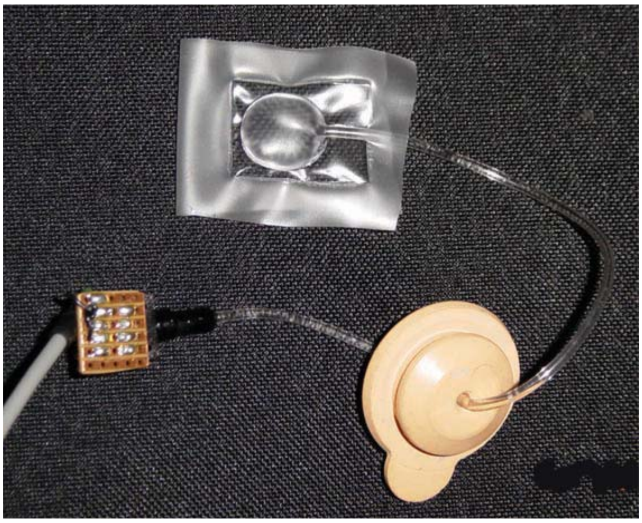


Fig. 7 Electrohydraulic sensor

3 EXPERIMENTAL SOCKETS AND DATA ACQUSITION

Fibre reinforced acrylic sockets containing machined metal inserts were manufactured using standard laminating techniques, for each subject. All subjects were fitted with check sockets prior to production of the experimental socket to ensure that the experimental socket was a total contact socket and was comfortable. These inserts allowed all transducers to be placed within the socket wall so that the sensing surface was flush with the inside surface of the socket. The sites of transducer location were determined in a similar manner for all experimental sockets for each individual subject, with the centre of the PT acting as a reference point. The cast was marked longitudinally at 45° segments using an 'indexing head' and the transducer sites were located on these lines at 50 mm intervals. The sockets were manufactured without a soft liner, i.e. they were the so-called 'hard' sockets.

The normal pressure and shear stresses were recorded simultaneously using two banks of 16-channel strain gauge amplifiers, model DAQN-Bridge (Dewetron, Austria), with PCMCIA LabView DaqCard 700 (National Instruments, Austin, TX, USA) for data acquisition and a Dell Inspiron PIII computer for data storage with LabView version 6.1 as the analysing software. Amplifier gains of up to 10 000 were possible, but in these studies gains of 5000 were used. The bridge voltage for the normal force channel was set to 6 V and shear force channels to 3 V. The sampling rate used was 50 Hz.

4 PERFORMANCE OF CUSTOM-MADE TRANSDUCERS

4.1 Calibration

Calibration of the assembled transducers was carried out using dead weights. Calibration rigs were developed to ensure accurate application of loads in both normal and shear directions. To calibrate in the normal force direction (F_y) the PT transducer and the BEST were fixed to a specially designed calibration jig incorporating a linear ball bearing to minimize friction (Fig. 8(a)). Single ball bearings were used to achieve pure axial loading, without introducing bending effects, and loading and unloading was only

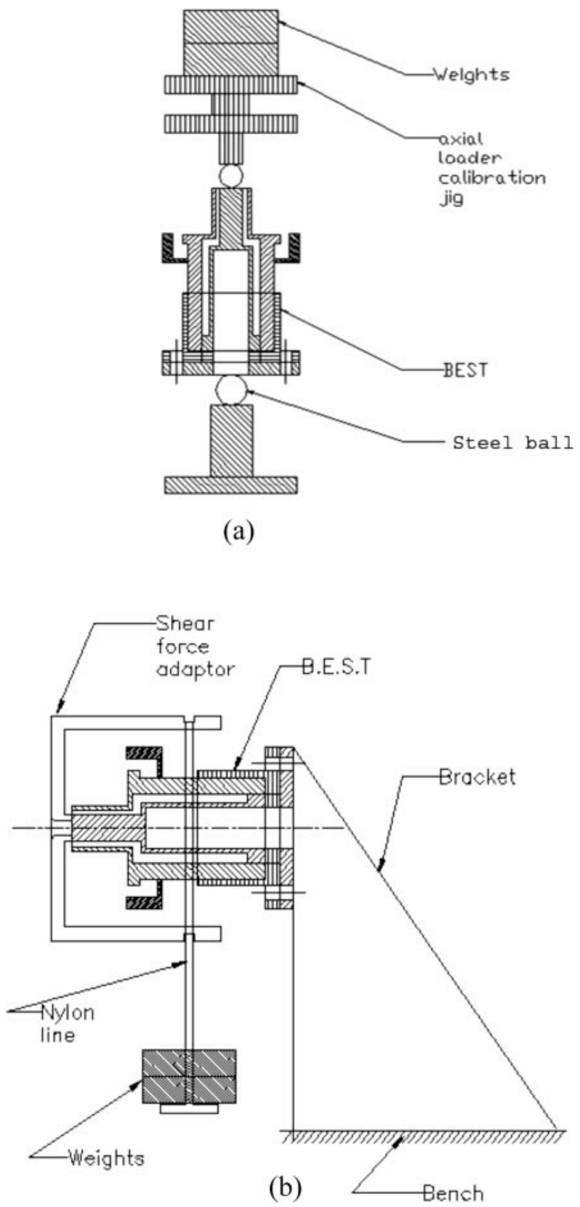


Fig. 8 Calibration arrangement of the transducers: (a) normal force channel; (b) shear force channel

applied in the compressive direction corresponding to a pressure of up to 400 kPa in ten steps (Fig. 9(a)). The signal outputs for all three channels were recorded simultaneously. For shear force calibration (F_x and F_z), a specially designed adapter was used. The transducer was positioned in a bench bracket which held it rigidly in a horizontal attitude, and was loaded using weights hung from the adapter (Fig. 8(b)). Thus, the force was applied directly over the shear strain gauges in the transducers, eliminating any bending moments. Known loads were then applied in five steps to correspond to a stress of up to 200 kPa and then unloaded to zero load while the signal outputs for all three channels were recorded simultaneously (Figs 9(b) and (c)). The calibration procedure was repeated ten times. On completion of the calibration of the positive F_z force channel, the transducer was rotated through 90° and the positive F_x force channel was then calibrated. Two further rotations of 90° enabled $-F_x$ and $-F_z$ force channels to be calibrated.

The linear mathematical approach was used to fulfil the accuracy required for the particular application. The calibration linear approach described simply assumes that each signal can be expressed as a linear combination of the components of the applied load.

From the data, a 3×3 matrix was generated, using the following method: the output signal, S_i ($I=1,2,3$), i.e. the output voltage that was measured is a direct function of the input signal, F_j ($j=1,2,3$), i.e. the applied load. In the present study each signal can be expressed mathematically as a combination of the components of the applied load

$$S_i = \sum_{j=1}^{3} M_{ij} F_j \tag{1}$$

where M_{ij} are proportionally constants relating the output to the input signals.

This expression may be expanded to

$$\begin{aligned} SF_x &= M_{11}F_x + M_{12}F_y + M_{13}F_z \\ SF_y &= M_{21}F_x + M_{22}F_y + M_{23}F_z \\ SF_z &= M_{31}F_x + M_{32}F_y + M_{33}F_z \end{aligned} \tag{2}$$

These equations state that each output signal will in general contain components corresponding to the forces F_x, F_y, and F_z. M_{11}, M_{22}, M_{33} are the principal sensitivity coefficients. All other coefficients, referred to as cross-talk coefficients, can be attributed to slight inaccuracies in the manufacture of the transducers and slight misalignments in the positioning of the strain gauges.

In matrix form, equations (2) can be expressed as

$$\mathbf{S} = \mathbf{M} \cdot \mathbf{F} \tag{3}$$

where

$$\mathbf{S} = \begin{bmatrix} SF_x \\ SF_y \\ SF_z \end{bmatrix}$$

a column matrix (3×1) of the output signal voltages;

$$\mathbf{M} = \begin{bmatrix} M_{11} & M_{12} & M_{13} \\ M_{21} & M_{22} & M_{23} \\ M_{31} & M_{32} & M_{33} \end{bmatrix}$$

a square matrix (3×3) containing calibration factors M_{ij} determined from the calibration data;

$$\mathbf{F} = \begin{bmatrix} F_x \\ F_y \\ F_z \end{bmatrix}$$

a column matrix (3×1) of the input applied loads. Equation (3) can be written as

$$\begin{bmatrix} SF_x \\ SF_y \\ SF_z \end{bmatrix} = \begin{bmatrix} M_{11} & M_{12} & M_{13} \\ M_{21} & M_{22} & M_{23} \\ M_{31} & M_{32} & M_{33} \end{bmatrix} \begin{bmatrix} F_x \\ F_y \\ F_z \end{bmatrix}$$

When a pure shear force, F_x, is applied to the transducer, the element M_{11} of matrix $\mathbf{M}$ representing the principal sensitivity coefficient and the two cross-talk M_{21}, M_{31} calibration factors can be determined. This is repeated for F_y and F_z to obtain the other calibration coefficients.

Re-arranging equation (3)

$$\mathbf{F} = \mathbf{M}^{-1} \cdot \mathbf{S} \tag{4}$$

where $\mathbf{M}^{-1}$ is the inverse of the calibration matrix $\mathbf{M}$, which is a more convenient equation during subject testing, where the output signals are recorded and the loads are to be determined.

To evaluate any cross-talk effects on the transducers' outputs due to the small bending moment arising from the presence of shear force at the interface, additional calibration tests were undertaken. In these tests, loads were applied on the transducers to simulate longitudinal and tangential shear force. The results of these tests showed that cross-talks due to these bending effects are negligible.

5 SUBJECTS AND TESTING PROTOCOLS

Ten trans-tibial amputees participated in the study on a voluntary basis, and gave written informed consent (Table 1). All were male and had undergone amputation at least 5 years prior to the study. Ethical approval was granted from the ethical committees of the University of Strathclyde and the Southern General Hospital NHS Trust, Glasgow, UK.

A TEC ProLink (TEC Interface Systems, Waite Park, Minnesota, USA) suspension sleeve was used as a suspension system and a one-way valve was sited at the distal posterior region of the socket. Each subject wore their experimental socket with stump socks, but no liner was used. Alignment was performed by one prosthetist until the subject's gait was agreed to be optimal by both subject and prosthetist. All data collection events for each experimental socket were performed on the same day with the amputee wearing the socket at all times. Static tests were conducted at the beginning of each session, during which the subject was asked to achieve full weight bearing on the prosthetic leg, followed by equal weight bearing on both legs. Data were collected approximately 5 s after the subject had achieved a stable position. Dynamic tests were conducted at the subject's self-selected walking speed, and the subject was asked to walk a distance of 7 m. At least three trials were recorded for each test. Only 12 of the 20 sites could be monitored by the data collection system simultaneously and therefore trials were repeated after relocating the transducers to ensure collection of data from all sites. Sites that were not monitored were sealed with blank transducer plugs to ensure that the pressure difference across the socket wall was maintained.

6 RESULTS

Five BESTs and one PT transducer were evaluated for hysteresis error. Hysteresis error is the maximum difference between output readings for the same point being recorded, and is obtained while increasing from zero to FSO and decreasing from FSO to

Table 1 Summary and characteristics of the ten male test subjects (SD, standard deviation; PVD, peripheral vascular disease)

Subject no.	Age	Height (m)	Mass (kg)	Body mass index, BMI (kg/m^2)	Reason for amputation	Amputated side	Stump length (mm)
1	77	1.78	95	29.98	Trauma	Right	177
2	73	1.91	80	21.93	Peripheral vascular disease	Right	125
3	58	1.69	72	25.21	Trauma	Left	145
4	53	1.74	95	31.38	Peripheral vascular disease	Left	123
5	56	1.78	76	23.99	Peripheral vascular disease	Left	150
6	61	1.78	108	34.09	Peripheral vascular disease	Left	120
7	49	1.85	133	38.86	Trauma	Right	145
8	34	1.75	67	21.88	Trauma	Right	130
9	47	1.8	89	27.47	Other	Left	148
10	61	1.68	64	22.68	Peripheral vascular disease	Right	130
Average ± SD	56.90 ± 12.47	1.78 ± 0.07	87.90 ± 21.09	27.75 ± 5.74			139.30 ± 17.30

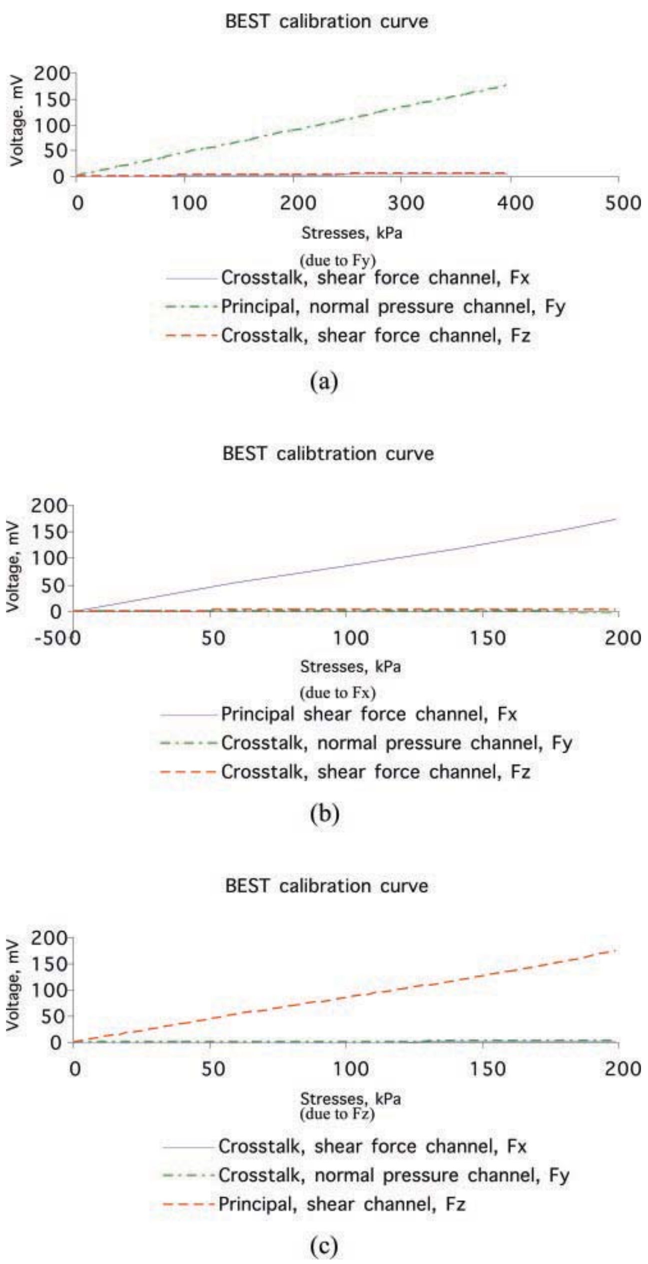


Fig. 9 Sample calibration curves for BESTs: (a) calibration in the normal force direction, F_y; (b) calibration in the tangential shear direction, F_x; (c) calibration in the longitudinal shear direction, F_z

zero. The deviation is expressed as a percentage of FSO. In this study, hysteresis was investigated during calibration by increasing and decreasing the loads applied to the transducers (Fig. 9). Average hysteresis error from the five BESTs was found to be 2.03 per cent FSO, for 200 kPa shear stress and 1.65 per cent FSO, for 400 kPa normal pressure. The hysteresis error from the PT transducer was found to be 1.53 per cent FSO, 200 kPa for the shear stress and 1.85 per cent FSO, 400 kPa for the normal stress. The hysteresis error obtained from this study falls within the range reported by Appoldt *et al.* [**3**], Sanders and Daly [**12**], and Williams *et al.* [**13**]. A summary of transducer hysteresis performance from this and other published studies is shown in Table 2.

Table 2 Comparison of reported hysteresis error from four studies

Researchers		Hysteresis error: Normal direction	Hysteresis error: Shear direction
Appoldt and Bennett [**3**]		3% FSO	Not reported
Williams *et al.* [**13**]		2% FSO	8% FSO
Sanders and Daly [**12**]		3% FSO	0.3% FSO
Current study	PT transducer	1.85% FSO	1.53% FSO
	BEST	1.65% FSO	2.03% FSO

Overall accuracy is defined as the tolerance within which a measurement can be repeated. For overall accuracy, under laboratory bench conditions, the average for the five BESTs was found within 99.56 per cent FSO for the shear direction, and within 99.64 per cent FSO for the normal direction; for the PT transducer the average was found within 99.65 per cent FSO for the shear direction and within 99.58 per cent FSO for the normal direction. During the amputee walking trials the transducers showed 92–97 per cent repeatability.

The data in Fig. 10 show sample results for one subject at a posterior-proximal site of the PTB socket using the BEST and the Entran-based device. The subjects loaded their prosthetic leg for approximately 63 per cent of each gait cycle. The results from the custom-made transducers did not differ significantly when compared with those obtained from the commercial load cell. Peak pressure results at the PT bar measured by the PT transducer during loading, from one trans-tibial subject, are as follows: tendon bar in the neutral position 220 kPa, indented by + 2 mm 284 kPa, indented by + 4 mm 417 kPa, and relieved from the neutral position − 2 mm 108 kPa. The results also show the repeatability during gait for all types of transducers (Figs 10(a), (b), and (c)). Average stresses distribution measured using the BEST from ten subjects for the PTB sockets are shown in Table 3.

7 DISCUSSION

The design and use of transducers for measuring, simultaneously, both normal and shear stresses at the stump–prosthesis socket interface have been described. The successful design specifications were determined from consideration of the work of previous researchers. This resulted in a transducer with a sensing flat surface of 5.6 mm in diameter, small enough to minimize interference distortion with the

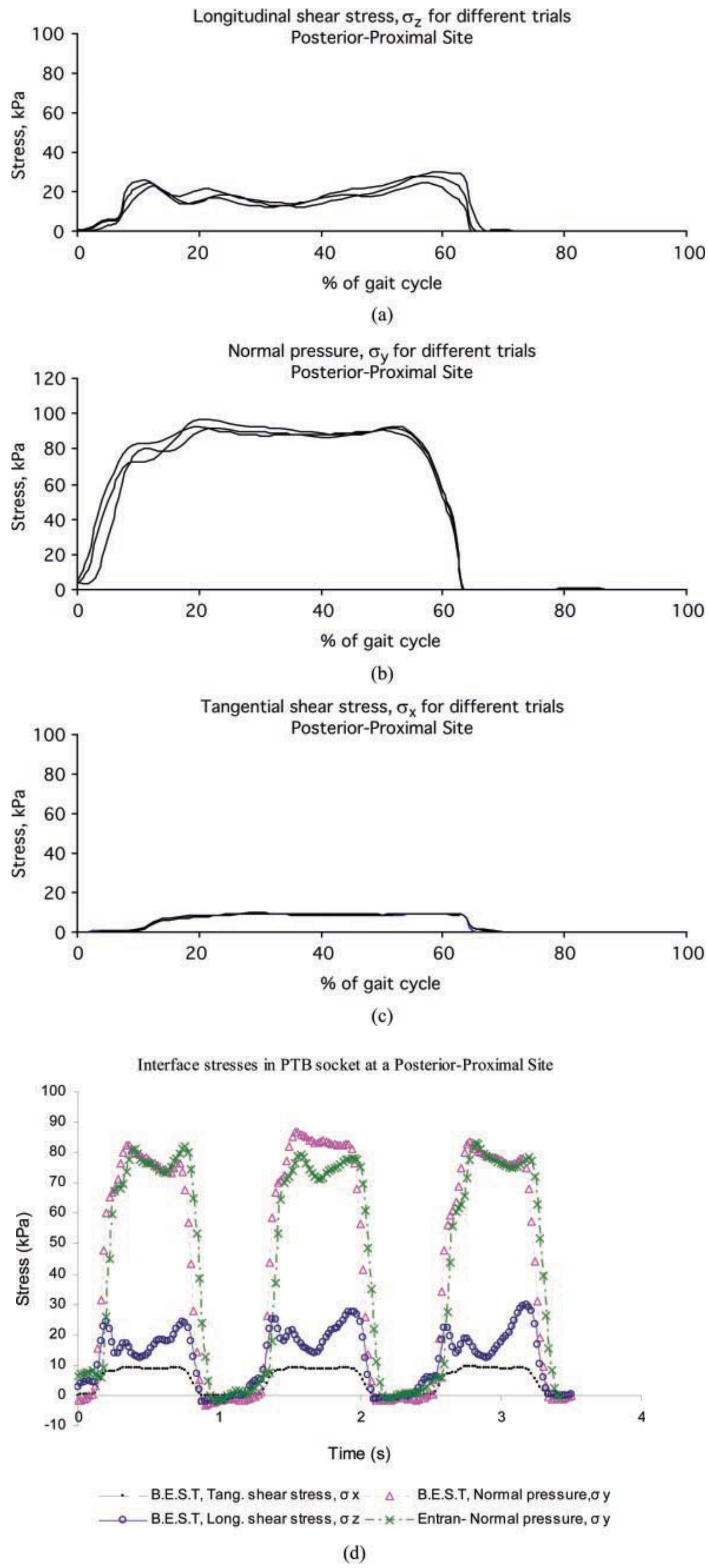


Fig. 10 (a), (b), (c) Stresses for single steps for three different trials from subject no. 3 during normal level walking. (d) Interface stresses for PTB sockets at posterior-proximal sites for three consecutive walking cycles (comparison between the BEST and Entran-based device)

Table 3 Average pressure and shear stresses recorded on the subjects tested wearing PTB sockets

	PTB socket walls			
	Anterior	Medial	Posterior	Lateral
Pressure (kPa), average ± 1 SD	115 ± 72	52 ± 27	61 ± 16	59 ± 31
Resultant shear stresses (kPa), average ± 1 SD	33 ± 10	26 ± 7	26 ± 6	26 ± 9

shape of the socket wall. In addition the 'laminating in' of metal inserts into the socket wall during socket manufacture ensured that the sensing surface of the transducer was flush with the inner surface wall of the socket. The small size of the transducers and the technique employed for positioning avoids protrusion into the soft tissues of the stump and minimizes measurement errors. The total mass of each experimental prosthesis, with all transducers and blanking plugs in situ, was approximately 3.0 kg. A TEC ProLink (TEC Interface Systems, Minnesota, USA) suspension sleeve in conjunction with a simple one-way valve was used for suspension. This method of suspension was found to eliminate successfully the pistoning action between the stump and the socket during walking. Hysteresis was not found to be significant. The methodology proved sound, and yielded repeatable results.

8 CONCLUSIONS

Transducers capable of measuring three forces, namely normal force and two perpendicular shear forces simultaneously, were designed, developed, and constructed using strain gauge technology. They proved to be robust and to provide repeatable results when utilized to measure stump–socket interface stresses. These transducers will help to provide an insight into the complex biomechanical interactions at the stump–socket interface, which may lead to the design of more functional and comfortable prosthetic sockets, minimizing the risk of tissue injury. These transducers are therefore suggested as a useful tool in the field of rehabilitation and may prove to be of value in other areas of research such as the limb/body–orthosis interfaces, the seat–buttock interface, and perhaps any general hard–soft interface, provided proper, carefully structured inserts are incorporated to allow the correct installation of the transducers.

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REFERENCES

1 **NASDAB.** *The amputee stastitical database for the United Kingdom*, 2003 (National Amputee Stastitical Database, Edinburgh, Scotland).

2 **Branemark, R., Branemark, P.-I., Rydevik, B.,** and **Myers, R.** Osseointegration in skeletal reconstruction and rehabilitation: a review. *J. Rehabil. Res. Dev.*, 2001, **38**(2), 175–181.

3 **Appoldt, F. A.** and **Bennett, L.** A preliminary report on dynamic socket pressures. *Bull. Prosthet. Res.*, 1967, **10**(8), 20–55.

4 **Appoldt, F. A., Bennett, L.,** and **Contini, R.** Tangential pressure measurement in above-knee suction sockets. *Bull. Prosthet. Res.*, 1970, **10**(13), 70–86.

5 **Burgess, E. M.** and **Moore, A. J.** A study of interface pressures in the below-knee prosthesis (physiological suspension: an interim report). *Bull. Prosthet. Res.*, 1977, **10**(28), 58–70.

6 **Katz, K., Susak, Z., Seliktar, R.,** and **Najenson, T.** End-bearing characteristics of patellar-tendon-bearing prostheses – a preliminary report. *Bull. Prosthet. Res.*, 1979, **10**(32), 55–68.

7 **Krouskop, T. A., Brown, J., Goode, B.,** and **Winningham, D.** Interface pressures in above-knee sockets. *Arch. Phys. Med. Rehabil.*, 1987, **68**, 713–714.

8 **Mizrahi, J., Susak, Z., Bahar, A., Seliktar, R.,** and **Najenson, T.** Biomechanical evaluation of an adjustable patellar tendon bearing prosthesis. *Scand. J. Rehabil. Med.*, 1985, **12**, 117–123.

9 **Pearson, J. R., Holmgren, G., March, L.,** and **Oberg, K.** Pressures in critical regions of the below-knee patellar-tendon-bearing prosthesis. *Bull. Prosthet. Res.*, 1973, **10**(19), 52–76.

10 **Van-Pijkeren, T., Naeff, M.,** and **Kwee, H. H.** A new method for the measurement of normal pressure between amputation residual limb and socket. *Bull. Prosthet. Res.*, 1980, **10**(33), 31–34.

11 **Rae, J. W.** and **Cockrell, J. L.** Interface pressure and stress distribution in prosthetic fitting. *Bull. Prosthet. Res.*, 1971, **10**(15), 64–111.

12 **Sanders, J. E.** and **Daly, C. H.** Measurement of stresses in three orthogonal directions at the residual limb–prosthetic socket interface. *IEE Trans., Rehabil. Engng*, 1993, **1**(2), 79–85.

13 **Williams, R. B., Porter, D., Roberts, V. C.,** and **Regan, J. F.** Triaxial force transducer for investigating stresses at the stump/socket interface. *Med. Biol. Engng Comput.*, 1992, **30**, 89–96.

14 **Winarski, D. J.** and **Pearson, J. R.** Least-squares matrix correlations between stump stresses and prosthesis

loads for below-knee amputees. *J. Biomech. Engng*, 1987, **109**, 238–246.

15 **Mak, A. F. T., Zhang, M.,** and **Boone, D. A.** State-of-the-art research in lower limb prosthetic biomechanics socket interface: a review. *J. Rehabil. Res. Dev.*, 2001, **38**(2), 161–174.

16 **Sanders, J., Daly, C.,** and **Burgess, E.** Interface shear stresses during ambulation with a below-knee prosthetic limb. *J. Rehabil. Res. Dev.*, 1992, **29**(4), 1–8.

17 **Sanders, J.** Interface mechanics in external prosthetics: review of interface stress measurement techniques. *Med. Biol. Engng Comput.*, 1995, **33**, 509–516.

18 **Sanders, J. E., Bell, D. M., Okumura, R. M.,** and **Draille, A. J.** Effects of alignment changes on stance phase pressures and shear stresses on transtibial amputees: measurement from 13 sites. *IEEE Trans., Rehabil. Engng*, 1998, **6**(1), 21–31.

19 **Sanders, J.** and **Daly, C.** Interface pressure and shear stresses: sagittal plane angular alignment effects in three transtibial amputee case studies. *Prosthetics Orthotics Int.*, 1999, **23**, 21–29.

20 **Tappin, J. W., Pollard, J. P.,** and **Beckett, E. A.** Method of measuring shearing forces on the side of the foot. *Clin. Phys. Physiol. Measmt*, 1980, **1**, 83–85.

21 **Lee, V. S. P., Solomonidis, S. E.,** and **Spence, W. D.** Stump–socket interface pressure as an aid to socket design in prostheses for transfemoral amputees – a preliminary study. *Proc. IMechE, Part H: J. Engineering in Medicine*, 1997, **211**(H2), 167–180. DOI: 10.1243/0954411971534287.

22 **Barbenel, J. C.** and **Sockalingam, S.** Device for measuring soft tissue interface pressure. *J. Biomed. Engng*, 1990, **12**(6), 519–522.