A treadmill ergometer for three-dimensional ground reaction forces measurement during walking

Alain Belli[a,*], Phong Bui[a,b], Antoine Berger[a], André Geyssant[a], Jean-René Lacour[b]

[a]*Laboratoire de Physiologie – GIP Excercice, Université de Saint-Etienne, France*
[b]*Laboratoire de Physiologie – GIP Excercice, Université Lyon I, Lyon, France*

Abstract

This report describes new treadmill ergometer designed to measure the vertical and horizontal ground reaction forces produced by the left and right legs during walking. It was validated by static and dynamic tests. Non-linearity was from 0.2% (left vertical force) to 1.4% (right antero-posterior force). The resonance frequency was from 219 (right vertical direction) to 58 Hz (left medio-lateral direction). A calibration "leg", an air jack in series with a strain gauge, was developed and used to produce force signals comparable to those obtained during human locomotion. The mean differences between the force measured by the calibration leg and treadmill ergometer at $5 \, \mathrm{km \, h^{-1}}$ were 3.7 N (0.7%) for the left side and 6.5 N (1.2%) for the right. Measurements obtained during human walking showed that the treadmill ergometer has considerable potential for analysing human gait. © 2000 Elsevier Science Ltd. All rights reserved.

1. Introduction

Ground-mounted force platforms have been used in many studies on humans to measure ground reaction forces (GRF) during walking and running (e.g. Bates et al., 1983; Chao et al., 1983; Nilsson and Thorstenson, 1989). However, force platforms are expensive, which limits the number of successive ground contacts they can measure. The short measurement distance also makes it difficult to obtain the same constant velocity in different trials. Lastly, the variability and asymmetry of kinematic and dynamic step parameters (Bates et al., 1979, 1983; Belli et al., 1995a) indicates the need for tools capable of measuring the GRF for many successive steps (Belli et al., 1995a) in order to obtain stable and representative value of step parameters.

Treadmills are often used to simulate overground human locomotion in the laboratory. Metabolic measurements can be made during long running bouts, but mechanical measurements are generally limited to treadmill velocity. Therefore attempts have been made to built treadmills equipped for GRF data acquisition.

Kram and Powell (1989), and more recently Fewster and Smith (1995), suggested incorporating a force platform under the moving belt of the treadmill. But, this provides accurate measurements only of vertical force because the horizontal force data are modified by friction forces between the moving belt and the fixed force platform. Horizontal GRF are important for determining the braking and propulsive phases of contact and can be used to calculate the horizontal kinetic energy changes of centre of mass of the subject (Cavagna, 1975). Both vertical and horizontal forces are also needed to determine the amplitude and the orientation of the total ground reaction vector, which are essential for calculating joint moments.

Both horizontal and vertical GRF can be measured by mounting the entire treadmill on a standard force platform or on specific force transducers. Although this system has been described (Johnson et al., 1993), the high mass and low stiffness values of conventional treadmill frames rendered accurate measurement of horizontal forces very difficult.

The present report describes and validates a new type of treadmill allowing accurate measurements of both vertical and horizontal GRF during walking. A specific treadmill frame was made as rigid as possible and bolted firmly to the ground through crystal force transducers.

---

* Corresponding author.

A prototype has been built and introduced in a previous abstract (Belli et al., 1995b). The present paper describes a second, improved prototype, its validation procedure, and typical measurements obtained during walking.

2. Methods

2.1. The treadmill

The entire treadmill is mechanically isolated. All the treadmill components, including the motor and the belt, were mounted on a single metal frame and fixed to the ground through three-dimensional (3D) force transducers. Thus, the forces due to belt friction and belt movement, can be considered to be internal forces and are not detected by the transducers. Only external forces, the 3D forces exerted by the feet on the isolated treadmill frame, are thus measured by the transducers. It is also necessary to measure the left- and right-foot forces during double contact phase of walking. Because measurements of footprints during walking have shown that the right- and left-feet contact the ground on either side of a longitudinal line representing the middle of the walkway (Chodera and Lewell, 1973; Holden et al., 1985; Chao et al., 1983), the treadmill consisted of two identical frames, one for the left and one for the right sides of the walking area (Fig. 1). They were separated by a 7 mm gap, making thus possible to obtain separate measurements of the left and right GRF.

The main problem was keeping the natural frequency of the treadmill frames as high as possible. The natural frequency of a flexed rigid frame is given by

$$F = \frac{1}{2\Pi} \times \sqrt{\frac{k}{M}}, \qquad (1)$$

where F is the resonant frequency (Hz), k the stiffness ($\mathrm{N\,M^{-1}}$) and M the mass (kg) of the frame. It was then necessary to minimise the mass and maximise the stiffness of the frame. As the frames of standard treadmill are not designed for this, a specific frame was designed and built by HEF Tecmachine (Andrézieux-Bouthéon, France). The frame was made of usual steel (AFNOR standardisation, type TS E 235), providing stiffness and good rigid connection between components. The rectangular section tubes were filled with polyester foam. Crystal force transducers (type KI 9067, Kistler, Wintertur, Switzerland) were used because of their ability to tolerate wide range of force measurements. The stiffness of these crystal transducers was $8.0 \times 10^9 \, \mathrm{N\,m^{-1}}$ for the vertical direction and $3.5 \times 10^9 \, \mathrm{N\,m^{-1}}$ for the horizontal direction (Kistler technical data), much higher than the stiffness of strain gauge transducers. Eq. (1) indicates that a frame weighing about 250 kg induces the resonant
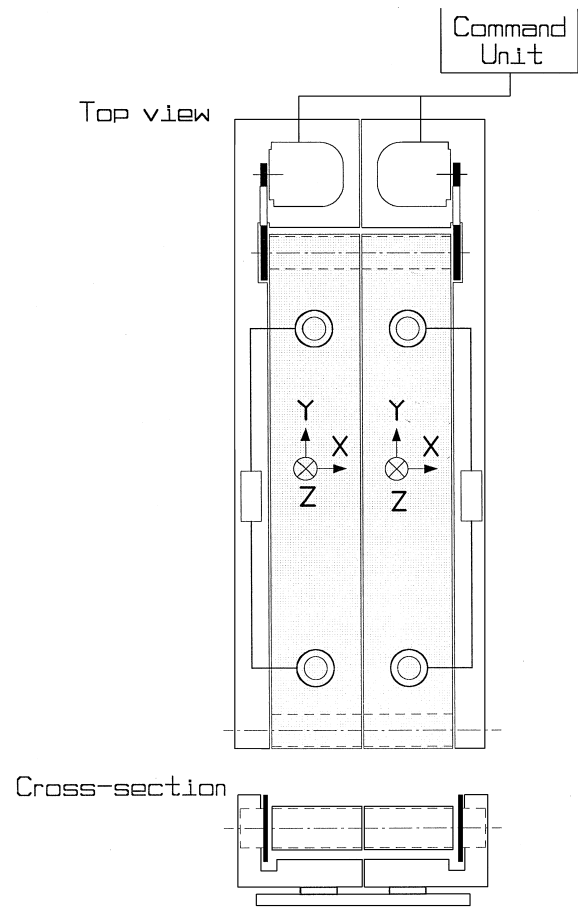

Fig. 1. Diagram of the treadmill ergometer.

frequencies of the mounted transducers to be 900 Hz (vertical) and 605 Hz (horizontal). However, as the total stiffness is probably determined by the low stiffness of the steel frame, natural frequencies of the treadmill would be lower. Because the frames will be probably flexed longitudinally, the frame length and specifically the distance between the transducers supporting the frame were minimised to keep the total stiffness high. But as comfortable walking needs a long belt, a compromise had to be found. According to Rosenrot et al. (1980) (cited by Kram and Powell, 1989) the relationship between the single foot contact distance and walking velocity of men is given by the equation $D_c = 0.665 + 0.25\,V$ where D_c is the contact distance (m) and V the walking velocity ($\mathrm{m\,s^{-1}}$). The maximal velocity of treadmill belts is $2.87\,\mathrm{m\,s^{-1}}$ ($10\,\mathrm{km\,h^{-1}}$), giving a theoretical maximal foot contact distance of 1.36 m. The longitudinal distance between force transducers was thus 1.50 m. However, comfortable walking requires that a belt longer than the contact distance; thus a 2 m long belt was used. Finally, footprint measurements (Chodera and Lewell, 1973; Holden et al., 1983; Chao et al., 1983) indicated that the each belt was 0.25 m wide.

The electro-mechanical and electronic components were selected to minimise mechanical and electrical noise. Belts were driven by 1.5 kW asynchronous motors (type LS MV 100L Leroy Somer, Saint-Etienne, France) mounted on silent blocks and monitored by a low noise controller (type 8200, Lenze, Hamein, Germany). Left and right side, the crystal force transducers were connected to processing units (type 5038A, Kistler, Wintertur, Switzerland), giving two horizontal and one vertical force signals per side.

A 16 channels, 12 bit, analog-to-digital interface card (type PC-LPM 16, National Instrument, Austin, TX, USA) in a PC was used for sampling the six force signals. Two other channels were used to monitor the instantaneous velocity of left and right treadmill belts, measured by optical transducers. Extra channels of the interface card could be used in future for other biomechanical measurements. Specific data acquisition software was written in Visual Basic 3.0 (Microsoft, USA).

2.2. Static evaluations

Tests were performed on the left and right side of the treadmill to evaluate the static characteristics of the treadmill. Vertical and horizontal force channels were sampled at 100 Hz. The linearity of vertical force measurements was determined by simply loading the left and right treadmill frames with calibrated, 20 kg masses up to maximum of 120 kg and then back to 0 kg. This test was also used to measure cross talk between vertical and horizontal forces. A hook, string, pulley and masses of 5 kg were used to determine the linearity of horizontal force measurements from 0 to ± 30 kg in the anteroposterior direction and from 0 to ± 20 kg in the median–lateral direction. Because the hook and string devices produced a residual vertical force we could not accurately measure the influence of horizontal forces on vertical force. The reproducibility of measurements was tested by loading and unloading each treadmill side 18 times by a subject (66.7 kg body mass) standing for 5 s at the centre of the frame supporting the belt. The influence of force position was tested by the same subject standing quietly on one foot at different locations (at the centre, at the four corners and at the middle of the four sides) of the right and left treadmill belts. Finally, the six force channels were sampled for 30 min with the system unloaded to evaluate the drift that usually occurs with crystal force transducers.

2.3. Dynamic evaluations

The dynamic characteristics of the treadmill were evaluated from measurements of forces produced by tapping, belt and motor movements, and by a calibration "leg". During these tests, vertical and horizontal force channels were sampled at 800 Hz. A fast Fourier transform algorithm was used to determine the amplitude frequency spectrum of the signal produced in the tapping test and during motor and belt movements. The frequency response of the treadmill was measured by tapping vertically the left and right belts of the treadmill with a plastic mallet and by calculating the frequency spectrum of the resulting force–time response. Electrical noise was measured by simply switching on the motor and setting the velocity to zero. Electrical and mechanical noises from the motor and belt movements were then measured at velocities of 2.5, 5 and 10 km h^{-1}. The treadmill response under dynamic conditions reproducing the force patterns encountered during walking as closely as possible was tested with a calibration "leg" (Fig. 2). This consisted of an air jack (type DNN 32, 12 bars max, Festo, Rosny sous Bois, France) mounted in series with a strain gauge force transducer (FGP Instrumentation, type FN 3030, 200 daN range, linearity 0.3%, Les Clayes sous Bois, France) and fixed at its upper end to horizontal axis bar oriented in the median–lateral direction of the treadmill. The axis bar was free to rotate and was kept at 0.945 m above the treadmill belt by a steel frame fixed to the ground (See Fig. 2). The lower end of the calibration leg was covered with a rubber pad and was free to move in the anteroposterior direction of the treadmill frame. The vertical tilt of the artificial leg was determined by an optical encoder (Hengsler, type RJM 600, Adlingen, Germany) measuring the rotation of the upper axis bar. During the tests, leg force and encoder signals were sampled and stored on the computer via a specially designed interface card including a strain gauge signal processor (Analog Device, type AD1B31AN) and a 12-bit AD converter (Analog Device type AD574AJN, Norwood, MA, USA) for force measurements and a 12 bit specialised counter (Hewlett Packard, type HCTL 2000, Palo Alto, CA, USA) for encoder measurements. The software synchronised and sampled data from both the calibration leg and treadmill transducers via the specific and standard interface cards.

The internal pressure of the air jack was first set at 5 bars with a bicycle pump. This pressure, corresponding to a maximal vertical force of about 650 N, was established during a preliminary test. Under pressure the calibration leg was at a maximal length of 1.06 m. The calibration leg was released over the front part of the left or right treadmill belt, which was moving at 3 or 5 km h^{-1} (0.83 or 1.38 m s^{-1}). During its backward movement, the leg shortened to 0.945 m at the vertical position and then lengthened back to 1.06 m. At the same time the leg force increased from 0 to a maximum when it was vertical and decreased back to zero.

The vertical tilt (α, in radian) and compression force (F_g, in N) of the calibration leg from the optical encoder and strain gauge measurements were used to calculate the theoretical horizontal (FL_h) and vertical (FL_v)
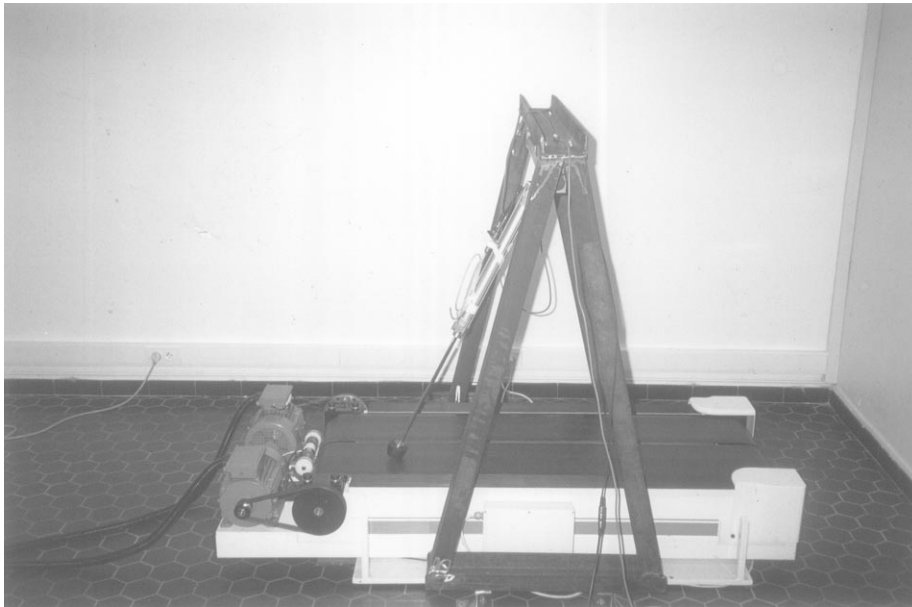
Fig. 2. The treadmill ergometer during dynamic evaluation with the artificial leg.

forces (in N) exerted on the treadmill belt as follows:

$$FL_h = F_g \cdot \sin \alpha, \quad FL_v = F_g \cdot \cos \alpha$$

FL_h and FL_v were then compared to the antero-posterior (FT_h) and vertical (FT_v) forces measured by the transducers of the treadmill by calculating the corresponding differences (ΔF_h and ΔF_v in Newton):

$$\Delta F_h = FL_h - FT_h, \quad \Delta F_v = FL_v - FT_v$$

2.4. Typical walking measurements

The vertical, anterior–posterior and median–lateral forces produced by a healthy adult man (63 kg body mass) walking on the treadmill at 4.5 km h^{-1} (1.25 m s^{-1}) were recorded. The subject was previously accustomed to treadmill walking (Wall and Charteris, 1981). Measurements were made during the last 30 s period of 2 min walking bout. The step time was defined as the duration between two successive foot contacts of the different legs. A threshold of 10 N on the vertical ground reaction force was used to define the beginning of the foot contact. Step frequency (in Hz) was the inverse of step time and step length (in m) was the step time multiplicated by the average belt velocity measured during the considered step.

2.5. Statistics

The non-linearity was defined as the maximal deviation from the least-squares linear regression force amplifier output versus load. The drift was quantified by linear regression between force and elapsed time.

3. Results

3.1. Static measurements

The maximal non-linearity (in percent of full scale of calibration) of the vertical force was $\pm 0.3\%$, that of the anterior–posterior force was $\pm 1.4\%$ and that of the median-lateral force was $\pm 0.7\%$ on the right side and they were ± 0.2, ± 0.3 and $\pm 0.4\%$ on the left side.

The reproducibility error of right vertical force was given by a standard deviation of ± 0.75 N (range 2.7 N) and that of the left vertical force was ± 0.72 N (range 2.7 N) corresponding to a relative error of 0.11% on both sides. The cross-talk produced by vertical force on anterior–posterior force was 0.9% and on median–lateral force it was 0.5% on the right side; the values were 0.8 and 0.2% on the left side.

The error due to the positions of the right and left vertical force was given by a standard deviation ± 6.6 N (range 23.4 N) and ± 6.3 N (range 22.5 N) corresponding to relative errors of 1.0% (right) and 0.96% (left). Detailed observation showed that the error due to median–lateral position was similar to the error due to anterior–posterior position. The maximal difference occurred between the anterior–lateral corner and the posterior–medial corner of the treadmill frame on both side.

The drift was 0.140 N min^{-1} (vertical), 0.002 N min^{-1} (anterior–posterior) and 0.015 N min^{-1} (median–lateral forces) for the right side, and 0.065 N min^{-1}, 0.017 N min^{-1} and 0.020 N min^{-1} for the right side. It was significantly ($P < 0.05$) related to the elapsed time except for the right anterior–posterior force ($r^2 = 0.21$, $P = 0.12$).

3.2. Dynamic measurements

In tapping test the damping time constant was lower than 40 ms in all force measurements. The fundamental resonance frequencies of the treadmill frames anterior–posterior and median–lateral directions were 219 Hz (vertical), 94 Hz (anterior–posterior) and 72 Hz (median–lateral (for the right side and 116, 60 and 58 Hz for the left side. Because the left and right parts of the treadmill were mounted, through the force transducers, on the same horizontal ground plate, the resonant frequency of the contralateral side appeared in the frequency spectrum of the tested side. But, the amplitudes of the oscillations due to the contralateral side were smaller than those of the studied side.

The output signals were not affected by switching on the motor and the electric noise measured at zero velocity was very low (standard deviations from ± 0.1 to ± 0.7 N). The noise produced by the motor and belt movements was velocity dependent. It was below ± 2.3 N at 2.5 km h^{-1}, ± 5.5 N at 5 km h^{-1} and ± 10.8 N at 10 km h^{-1}, for all the force channels and both frame sides. The first significant value of the frequency spectrum of motor noise was always obtained above 40 Hz, and main peaks were calculated at frequencies equal to or higher than 58 Hz, whatever the force channel and the velocity conditions.

3.3. Calibration leg measurements

Typical signals obtained with the calibration leg are shown in Fig. 3. The vertical and anterior–posterior force signals obtained from the artificial leg and treadmill changed similarly. The vertical force signals showed a positive peak corresponding to the impact of the calibration leg in the front of the treadmill belt. This impact also produced a maximal negative anterior–posterior force of -252 ± 10 N. The calibration leg vertical force then increased until the leg was vertical (max 663 ± 12 N) while the anterior–posterior force decreased from the maximal negative value to zero. The vertical force decreased to zero thereafter, while the anterior–posterior force increased (max. 160 ± 7 N) until just before release, and then decreased rapidly to zero. The duration of contact was 1.230 ± 0.010 s at 3 km h^{-1} and 0.735 ± 0.005 s at 5 km h^{-1}. The braking phase was always 52% of the contact phase. Values of ΔF_v and ΔF_h during contact at 3 km h^{-1} were -2.1 ± 9.9 and -2.2 ± 6.5 N for the left side and -5.1 ± 10.9 and -5.2 ± 4.6 N for the right side. The ΔF_v and ΔF_h were -3.6 ± 18.0 and -3.7 ± 8.0 N for the left side and -6.3 ± 20.9 and -6.4 ± 5.6 N for the right side at 5 km h^{-1}, respectively. These differences were due to two different types of oscillations. One oscillation, was low amplitude and high frequency (> 50 Hz), and measured only by the force transducers of the treadmill. The second characterised by

low-frequency (17–30 Hz) oscillations, was found in treadmill force transducers and, with a smaller amplitude, in the strength gauge signal of the artificial leg. This second oscillations occurred after impact and before release, and not when the calibration leg was close to vertical.

3.4. Walking measurements

A total of 27 left steps and 27 right steps were recorded during the 30 s measurement, giving a step time of 0.556 ± 0.008 s and a frequency of 1.80 ± 0.03 Hz. The subject had no problem keeping his/her left and right feet on the corresponding frames. The treadmill belt velocity was 1.250 ± 0.033 m s^{-1} (range 1.116–1.342 m s^{-1}) on the right side and 1.249 ± 0.30 m s^{-1} (range 1.140–1.337 m s^{-1}) on the left. The step length was then 0.694 ± 0.010 m. Typical force–time curves are shown in Fig. 4.

4. Discussion

4.1. Static characteristics

The non-linearity of the static measurements was compatible with the $\pm 1\%$ maximal non-linearity given by the manufacturer of the crystal transducers, except for the 1.4% non-linearity of the anterior–posterior force of the right frame. Although the dimensions of the frames were accurately measured and the force transducers were fitted according to the specifications of the manufacturer, the non-linearity of the right side force channels could not be reduced. Nevertheless, these data were comparable with the non-linearity of standard force platform measurements. The reproducibility (0.1% non-reproducibility) was also high and the error due to the position (1%) was comparable with that measured in treadmill-mounted force platforms (Kram and Powell, 1989). The crosstalks (From 0.2 to 0.9%) due to the left and right vertical forces were in agreement with transducers specifications (crosstalk $\leqslant \pm 1\%$). As previously mentioned, it was not possible to accurately measure the crosstalk due to horizontal forces on vertical force during the calibration procedure. However, horizontal forces are smaller than vertical forces in walking, so that the relative cross-talk errors caused by horizontal forces should be smaller than those caused by vertical forces.

The measured drift was much lower than the maximal possible drift of 0.4 N s^{-1} calculated from crystal transducers and change amplifier specifications. It is therefore, not necessary to reset the treadmill transducers for short measurements (i.e. 3–5 min). Longer measurements, such as those used in fatigue studies, may require resetting of the crystal transducers during the swing phase of the leg using a specific electronic and software system.
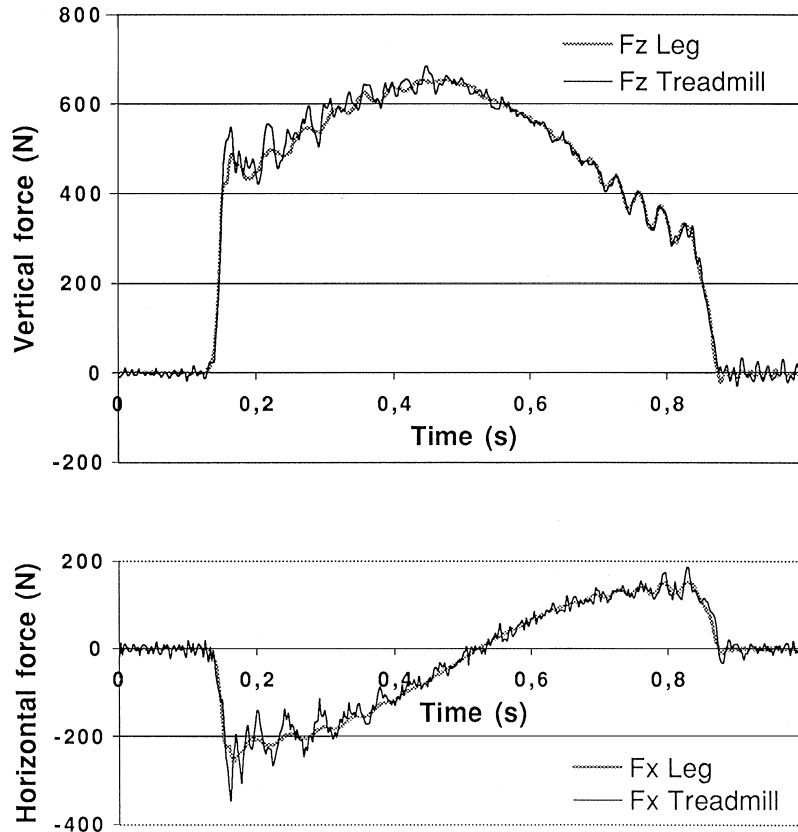
Calibration leg - 1.4 m/s - Left side

Fig. 3. Typical raw (non-filtered) data obtained during dynamic evaluation with the artificial leg.

4.2. Dynamic characteristics

The resonant frequencies were lower than those expected from transducers characteristics and treadmill mass (900 Hz (vertical) and 605 Hz (horizontal), respectively, see Section 2) showing that the stiffness of the frames plays a major role in the dynamic characteristics of the ergometer. However, it was difficult to further increase the resonance frequencies of mechanical structures, because this required adjusting many separate components. This is illustrated by the different resonance frequencies on the left and right apparently identical sides. Thus the resonance frequency of every new frame must be established before use. Nevertheless, the minimal resonance frequencies (116 Hz (vertical) and 58 Hz (horizontal) of the present ergometer are satisfactory in respect to the frequency spectrum of ground reaction forces of walking (Winter et al., 1974).

Electromagnetic and mechanical noises caused by belt and motor movements were small below 5 km h^{-1} (maximally $\pm$ 5.5 N), whatever the force channel or the treadmill side. It was slightly better than the motor and belt noise ($\pm$ 20 N) found by Kram and Powell (1989) on

vertical force measurements. Moreover, this noise was high frequency (60 Hz), an thus readily removed with a digital low-pass filter (Winter et al., 1974).

4.3. Calibration leg test

The force–time relationship obtained were comparable to typical curves measured on a force plate during human locomotion (Nilsson and Thorstensson, 1989), especially for the GRF of running. However, the vertical force trough usually encountered in GRF of walking when the leg is vertical was not observed in the present data (see Fig. 4). Nevertheless the duration of contact, braking and pushing phases were similar to those obtained on force plates at walking velocities of 3 and 5 km/h (Chao et al., 1983; Nilsson et al., 1985). Force amplitudes were also compatible with data obtained by a 60 kg human subject. Therefore, the artificial leg method provides ground reaction force patterns for calibrating the treadmill ergometer under conditions similar to real human locomotion.

The force differences between the treadmill and artificial leg measurements (from $\pm$ 4.6 to $\pm$ 20.9 N) were greater than, but still comparable to, the errors measured
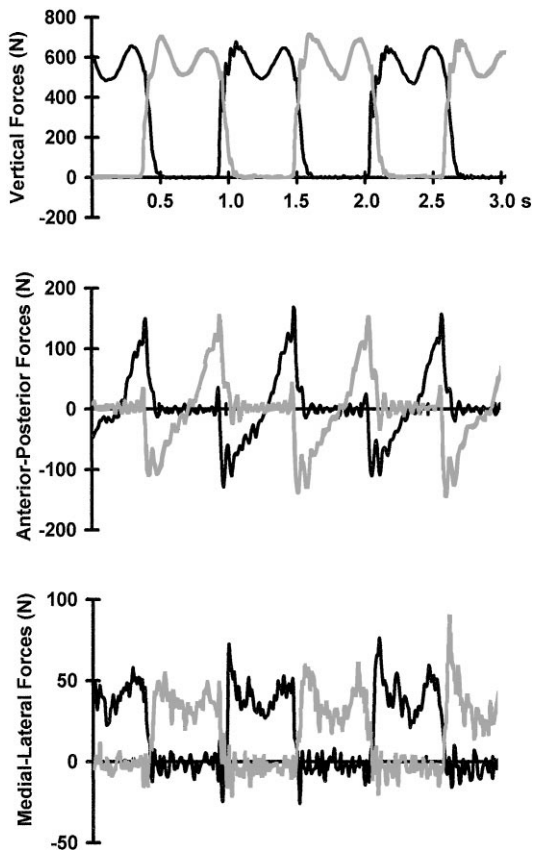
Fig. 4. Typical raw (non-filtered) data obtained during walking. Solid line and shaded line correspond to right and left ground reaction forces, respective. Time scale is the same for all measurements.

in static validation and the noise induced by belt and motor movements. These differences were due to two types of oscillations. The first, high-frequency (> 50 Hz) oscillations, was consistent with the resonance frequencies of the treadmill frame. This type of oscillations is readily filtered out. The second, low-frequency (17–30 Hz) oscillations did not correspond to the resonant frequencies of the treadmill frame, but oscillations with the same frequency were measured on treadmill belt velocities. Therefore, these force oscillations were due to changes in the velocity of the treadmill belt, probably due to friction between the moving belt and the treadmill frame and to the compliance of the belt. This is supported by the fact that these oscillations were present in both treadmill and artificial leg force data.

4.4. Walking test

The step time, step frequency and step length were comparable to those obtained at comparable velocities during standard treadmill and ground walking (Wall and Charteris, 1981, Nilsson and Thorstensson, 1989). The pattern and amplitude of force-time curves (Fig. 4) were also similar to the data obtained by using ground-moun-

ted forces plates (Nilsson and Thorstensson, 1989). The main advantage of such treadmill measurements is the capacity to quickly record and analyse forces from a large number of steps at a fixed velocity. The variations between and asymmetry of steps (Bates et al., 1979, 1983; Belli et al., 1995a) means that measuring the GRF of many successive steps can reduce the errors of kinematic and dynamic measurements (Belli et al., 1995a) and provide a better understanding of human locomotion.

This paper was not intended to analyse the forces obtained during treadmill walking, only to show the potential of the treadmill ergometer. The present data will therefore not be discussed further. However, detailed analysis of data showed that low frequency oscillations, similar to those mentioned in the artificial leg test, were also present in the force and velocity data collected during treadmill walking. The same phenomenon is also present during running on a treadmill mounted force platform (see Fig. 5 in Kram and Powell, 1989). Lastly the heel contact produced a sudden decrease in the treadmill velocity signal in line with data obtained in treadmill running (Belli et al., 1995; Cavagh and Kram, 1989). Therefore, future treadmill design should specify and reduce the velocity oscillations of the belts. Nevertheless, the measurement possibilities provided by the treadmill ergometer could be useful to further describe and analyse the locomotion of healthy and infirm subjects.

References

Bates, B.T., Osternig, L.R., Moson, B.R., James, S.L., 1979. Functional variability of lower extremely during the support phase of running. Medicine in Science in Sports 11, 328–331.

Bates, B.T., Osternig, L.R., Sawhill, J.A., James, S.L., 1983. An assessment of subject variability, subject-shoe interaction, and the evaluation of running shoes using ground reaction force data. Journal of Biomechanics 16, 181–191.

Belli, A., Lacour, J.R., Komi, P.V., Candau, R., Denis, C., 1995a. Mechanical step variability during treadmill running. European Journal of Applied Physiology 70, 510–517.

Belli, A., Bui, P., Berger, A., Lacour, J.R., 1995b. A treadmill for measurements of ground reaction forces during walking. In: Proceedings of the XVth Congress of the International Society of Biomechanics, July 2–6, Jyväskylä, Finland, pp. 100–101.

Cavagna, G.A., 1975. Force platforms as ergometers. Journal of Applied Physiology 19, 249–256.

Cavanagh, P.R., Kram, R., 1989. Stride length in distance running: velocity, body dimensions, and added mass effects. Medicine and Science in Sports 21, 467–479.

Chao, E.Y., Laughman, R.K., Schneider, E., Stauffer, R.N., 1983. Normative data of knee joint motion and ground reaction forces in adult level walking. Journal of Biomechanics 16, 219–233.

Chodera, J.D., Levell, R.W., 1973. Footprint patterns during walking. In: Kenedi, R.M. (Ed.), Perspectives in Biomechanical Engineering. University Park press, Baltimore, MD, USA, pp. 81–90.

Fewster, J.B., Smith, G.A., 1995. Development of a treadmill for measuring vertical gound reaction forces and center of pressure during gait. In: Proceeding of the XVth Congress of the International Society of Biomechanics, July 2–6, Jyväskylä, Finland.

Holden, J.P., Cavanagh, P.R., Williams, K.R., Bednarski, K.N., 1985. Foot angles during walking and running. In: Winter, D.A., Norman, R.W., Wells, R.P., Hayes, K.C., Patla, A.E. (Eds.), Biomechanics IXa. Human Kinetic Publisher, Champain, IL, USA.

Kram, R., Powell, A.J., 1989. A treadmill mounted force platform. Journal of Applied Physiology 67, 1692–1698.

Nilsson, J., Thorstensson, A., Halbertsma, J., 1985. Change in leg movements and muscle activity with speed of locomotion and mode of progression in humans. Acta Physiologica Scandinavica 123, 457–475.

Nilsson, J., Thorstensson, A., 1989. Ground reaction forces at different speed of human walking and running. Acta Physiologica Scandinavica 136, 217–227.

Wall, J.C., Charteris, J., 1981. A kinematic study of long-term habituation to treadmill walking. Ergonomics 24, 531–542.

Winter, D.A., Sidwall, H.G., Hobson, D.A., 1974. Measurement and reduction of noise in kinematics of locomotion. Journal of biomechanics 7, 157–169.